The Biomechanics of the Dynamic Defense Mechanism

By

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Submitted to the University of Hertfordshire in partial fulfilment of the requirements of the degree of PhD

July 2013

Acknowledgements

First and foremost, I would like to thank my principal supervisor, and friend, Mitch. Your continuous help, encouragement and motivation helped me through what seemed like the ‘never ending’ years. I would also like to thank Professor Tim Watson for his rigorous critique of the methods used in this project, which without, my knowledge would be inferior.

To the ‘super techs’, Neil Willmore, Camilla Holland and Hassan Khalil, who without their constant help, support and crude sense of humours I could not have completed this project in such good spirit.

To the kind volunteers who agreed to take part in my studies, I could not have done it without you.

And finally, to mum and dad for all their continued love and support throughout not only this PhD, but my entire University career; I could not have done it without you both. You have helped me to build a future for myself, for which I will be eternally grateful.

Abstract

Context: It has been suggested that muscle fatigue can lead to injury, however, research investigating this phenomenon in functional ankle instability (FAI) subjects is lacking. **Aim:** The purpose of this thesis was to research postural sway and muscular latency in FAI subjects and healthy controls, both before and immediately after localised and globalised fatigue protocols. **Subjects:** All subjects used in this project were males, between the ages of 18 and 25 years, and participated in regular ($\geq$2 x week) aerobic exercise. Subjects were categorised into healthy subjects, or subjects with a history of FAI using the FAI questionnaire. **Methods:** Neuromuscular control was analysed in FAI subjects and healthy controls through measures of muscular latency and postural sway. These measures were repeated both before and immediately after localised and globalised fatigue protocols. **Results:** The induction of localised and globalised fatigue had no effect on muscle latency in the FAI or healthy subjects. However, postural sway was significantly increased in the FAI subjects, following localised and globalised fatigue, with globalised fatigue also significantly increasing postural sway in the healthy subjects. The globalised football-specific fatigue protocol caused the greatest deficits in the FAI subjects, but also the healthy controls. **Conclusions:** In terms of muscle latency individuals that participate in sports, as well as sports clinicians and coaches, should not be concerned about the theorised relationship between the onset of fatigue and an increased injury risk at the ankle. However, in terms of postural sway the globalised football-specific fatigue protocol caused the greatest deficits. This highlights that the fatigued individual may be at greater risk of musculoskeletal injury during prolonged exercise that involves multiple joints, such as a football match.

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Abbreviations

AMTI	Advanced Mechanical Technology Inc.
ASIS	Anterior Superior Iliac Spine
AVR	Average Rectified
COM	Centre of Mass
COP	Centre of Pressure
DA	Dominant Ankle
EMG	Electromyography
GM	Gluteus Medius
ICC	Intraclass Correlation Coefficient
NDA	Non-Dominant Ankle
NS	Non Stabilised Setup
PL	Peroneus Longus
RMS	Root Mean Squared
SA	Stable Ankle
SD	Standard Deviation
SEBT	Star Excursion Balance Test
SEM	Standard Error of Measurement
SENIAM	Standard EMG for a Non-Invasive Assessment of Muscles
SS	Stabilised Setup
TA	Tibialis Anterior
UA	Unstable Ankle

Chapter One

Introduction

Up to 302,000 patients attend accident and emergency departments with ankle sprains each year (Bridgman, Clement, Downing, Walley, Phair & Maffulli, 2003). The lateral ligament complex is the most commonly injured structure in the ankle joint (Wolfe, Uhl, Mattacola & McCluskey, 2001), representing up to 95% of all ankle sprains (Messina, Farney & DeLee, 1999). The mechanism of injury for an ankle sprain is usually forced talocrural joint plantarflexion and subtalar joint inversion (Mitchell, Dyson, Hale & Abraham, 2008a). Recurrent sprains have been reported in over 70% of patients who had previously sustained an inversion ankle sprain (Kent-Braun, 1999; Yeung, Chan, So & Yuan, 1994). Recurrent sprains, residual disability, a feeling of "giving way", and a sensation of joint weakness characterise functional ankle instability (FAI), a condition that often arises secondary to inversion trauma (Beckman & Buchanan, 1995; Fernandes, Allison & Hopper, 2000; Konradsen, Olesen & Hansen, 1998; Konradsen & Ravn, 1991). Due to the significant amount of time lost from sport, work and leisure-time activities, research on the factors that contribute to FAI is warranted.

Muscle latency of the peroneus longus and tibialis anterior has commonly been assessed in individuals suffering from FAI (Ebig, Lephart, Burdett, Miller & Pincivero, 1997; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990; Mitchell et al., 2008a; Vaes, Duquet & Van Gheluwe, 1999) due to the protective function of these muscles to resist inversion and plantarflexion, respectively. However, there is very limited research on the muscle latency of the gluteus medius muscle (Beckman & Buchanan, 1995). During the gait cycle the gluteus medius muscle provides stability to the hip in the frontal plane (Friel, McLean, Myers & Caceres, 2006). Weakness in a stabilising muscle, such as the gluteus medius, may produce deviations in joint motion, a subsequent loss of stability and may contribute towards a repeated injury at the ankle

(Friel et al., 2006; Riemann, 2002). Further investigation in to the role of the gluteus medius muscle is therefore necessary.

Impaired postural control with increased amplitude and speed of centre of pressure (COP) movements has often been reported in patients with functional ankle instability (Friden, Zatterstrom, Lindstrand & Moritz, 1989; Goldie, Evans & Bach, 1994; Hale, Hertel & Olmsted-Kramer, 2007; Harkins, Mattacola, Uhl, Malone & McCrory, 2005; Tropp, Ekstrand & Gillquist, 1984). Postural sway analysis times often differ between studies, with some being as long as 30 seconds (Fu & Hui-Chan, 2005; Leanderson, Bergqvist, Rolf, Westblad, Wigelius-Roovers & Wredmark, 1999; McGuine, Greene, Best & Leverson, 2000; Trojian & McKeag, 2006). No explanation is given by these authors for their balance time chosen, and often the long duration of balancing time is not specific to a real sporting situation. Wilkinson and Allison (1989) identified 200 ms as the average fastest reaction time in 20-29 year olds; therefore, anything prior to 200 ms would be beyond human conscious control. Analysis of the subconscious time period may identify postural sway deficits that are sometimes not present in FAI subjects when analyzing a conscious time period.

Some authors have suggested that fatigue plays a significant role in the occurrence of ankle injuries (Gribble & Hertel, 2004a; Huston, Sandrey, Lively & Kotsko, 2005; Ochsendorf, Mattacola & Arnold, 2000; Pasquet, Carpentier, Duchateau & Hainaut, 2000). Anecdotally, it has been reported that most of these injuries occur at the end of an activity when the participant is fatigued (Hawkins, Hulse, Wilkinson, Hodson & Gibson, 2001). In terms of muscular latency, research has found that isokinetic fatigue has led to increased (delayed) muscle latencies in healthy subjects (Cools, Witvrouw,

Declercq, Danneels & Cambier, 2002). If fatigue has a detrimental effect on muscle latency, this could potentially lead to an increased risk of injury. While several studies have evaluated muscle latencies in healthy versus FAI subjects (Beckman & Buchanan, 1995; Ebig et al., 1997; Johnson & Johnson, 1993; Konradsen et al., 1998; Konradsen & Ravn, 1991), a better understanding of the musculature responses to an inversion-plantarflexion stress in a fatigued state may help to clear up discrepancies in the literature, and identify if fatigue is a risk factor that may lead to an ankle sprain in healthy subjects, or lead to repeated sprains in FAI subjects.

There is also evidence to support a relationship between fatigue and impaired static postural control (Gribble & Hertel, 2004a; Johnston, Howard, Cawley & Losse, 1998; Lundin, Feuerbach & Grabiner, 1993; Miller & Bird, 1976). Nelson and Johnson (1973) studied the effects of both localised and globalised fatigue on postural stability. Both the global and local fatigue models indicated a decline in static balance, but the generalised mode of fatigue exhibited a greater amount of sway within subjects. Often the methods of assessing postural stability are static, alongside methods of inducing fatigue which are not particularly sports related. Therefore, there is a clear demand for further research into the effects of more sports specific fatigue protocols, such as those employed by Drust, Cable and Reilly (2000), on dynamic postural stability tasks.

1.1 Main Aims

The main aims of this thesis were:

- To evaluate muscle latency in FAI subject's compared to healthy controls

- To evaluate single limb postural sway in FAI subject's compared to healthy controls.

- To research muscle latency in FAI subject's compared to healthy controls, both before and immediately after localised and globalised fatigue protocols.

- To research single limb postural sway in FAI subject's compared to healthy controls, both before and immediately after localised and globalised fatigue protocols.

1.2 Objectives

The main objectives of this thesis were to:

- Measure muscle latency in FAI subjects compared to healthy controls using electromyography.

- Measure postural sway in FAI subjects compared to healthy controls using a force platform.

- Measure muscle latency in FAI subjects compared to healthy controls both before and immediately after localised ankle and hip isokinetic protocols and a globalised football-specific fatigue protocol, using electromyography.

- Measure postural sway in FAI subjects compared to healthy controls both before and immediately after localised ankle and hip isokinetic protocols and a globalised football-specific fatigue protocol, using a force platform.

1.3 Hypotheses

Below are the main hypotheses referring to the thesis as a whole; specific hypotheses are identified within the individual studies.

H_1 - FAI subjects will have significantly increased (delayed) muscle latencies in comparison to the healthy controls.

H_2 - FAI subjects will have significantly increased postural sway in comparison to the healthy controls.

H_3 - The fatigue protocols will further increase the effect of delayed muscle latencies in the FAI subjects, in comparison to the healthy controls.

H_4 - The fatigue protocols will further increase the effect of greater postural sway in the FAI subjects, in comparison to the healthy controls.

1.4 Contributions to the Literature

Pilot study seven and pilot study nine from this thesis have both been published in the International Journal of Sports Medicine.

Gautrey, C. N., Watson, T. & Mitchell, A. (2013). The effect of isokinetic testing speed on the reliability of muscle fatigue indicators during a hip abduction-adduction fatigue protocol*. International Journal of Sports Medicine, 34*, 646-653.

Gautrey, C. N., Watson, T. & Mitchell, A. (2013). The effect of velocity on load range during isokinetic hip abduction-adduction exercise. *International Journal of Sports Medicine, 34,* 623-630.

Several other papers will be submitted for publication in the near future.

Chapter Two

Review of Literature

2.1 Ankle Sprain Epidemiology

Approximately 5,000 patients suffering from ankle sprains are admitted to accident and emergency every day in the United Kingdom (Heyworth, 2003). Lateral ankle sprains have often been reported as the most common injury in sport (Barrett & Bilisko, 1995; Orteza, Vogelbach & Denegar, 1992; Robbins, Waked & Rappel, 1995), but also occur among other physically active individuals (Ross & Guskiewicz, 2004). It has been reported that between 10% (Barker, Beynnon & Renström, 1997; Smith & Reischl, 1986) and 30% (DeLoes & Goldie, 1988) of all injuries sustained in sport are to the ankle complex. The lateral ligament complex is the most frequently injured structure in the ankle joint (Wolfe et al., 2001), representing from 45% (Liu & Jason, 1994) to 95% of all ankle sprains (Messina et al., 1999). The incidence of FAI, exhibiting residual symptoms such as feelings of instability, giving way, pain or re-injury, has varied from 10% (Forestier & Toschi, 2005) up to 80% in ankle sprain sufferers (Smith & Reischl, 1986). The amount of time lost due to ankle sprains ranged from 16% (Liu & Jason, 1994) to 25% of total playing time (Ashton-Miller, Ottaviani, Hutchinson & Wojtys, 1996; Mack, 1982). It has been suggested that ankle sprain injuries are usually sustained in sports involving running (Barrett & Bilisko, 1995), cutting (Barrett & Bilisko, 1995), jumping (Callaghan, 1997) and contact with other players (Garrick & Requa, 1989; Kuwada, 1995). This may explain the high incidence of ankle sprains in sports such as football (Hawkins et al., 2001).

It has been estimated that between 33% (Leanderson et al., 1999) and 55% (McKay, Goldie, Payne & Oakes, 2001) of individuals suffering from ankle sprains do not seek injury treatment from a health care professional (McKay et al., 2001; Smith & Reischl,

1986). Thus, the incidence of ankle sprains may often be underestimated (Ekstrand & Gillquist, 1983; Hertel, 2002). The main sufferers of ankle sprains are young active males, and injury may lead to a loss of working hours (Brooks, Potter & Rainey, 1981; DeLoes, 1990). Due to the high number of injuries, even a small decrease in the incidence of recurrent sprains would mean great economical savings, less playing time lost from sport, and an improvement in the everyday life of FAI sufferers (Leanderson et al., 1999).

2.2 Dominant versus Non-Dominant Limb

It has been shown repeatedly that ankle sprains tend to affect the dominant leg of a sports person (Gribble, Hertel & Denegar, 2007; Woods, Hawkins, Hulse & Hodson, 2002). Ashton-Miller et al. (1996) stated that the dominant limb is involved more than twice as often as the non-dominant limb. However, there are many different definitions of the 'dominant limb' in the literature which renders the studies difficult to compare. Bressel, Yonker, Kras and Heath (2007) defined the dominant limb as the preferred leg for kicking a ball, whereas, Gribble, Hertel et al. (2007) defined the dominant limb as the limb the subject would choose to stand on whilst kicking a ball. The justification for using the balancing leg as the dominant limb was that in terms of postural sway the balancing leg would be the dominant limb. Youdas, Loder, Moldenhauer, Paulsen and Hollman (2006) complicated the literature further by defining the 'dominant limb' as the preferred leg to kick a ball, but then used the non-dominant leg as the stance leg during testing. This may indicate that Youdas et al. (2006) deem the non-dominant limb to actually be the dominant limb in terms of postural sway. Results of studies referring to dominant

and non-dominant limbs therefore have to be interpreted with caution (Delahunt, 2007a).

2.3 Mechanism of Injury

The mechanism of injury for a lateral ankle sprain is talocrural joint plantarflexion and subtalar joint inversion of the ankle (Willems, Witvrouw, Delbaere, Cock & Clercq, 2002; Woods, Hawkins, Hulse & Hodson, 2003). Two different types of ankle sprains commonly occur in sport (Jacobs, Uhl, Mattacola, Shapiro & Rayens, 2007). Those which arise from a direct force such as landing on an opponent's foot, uneven terrain or a forceful kick to the foot from an opponent (Jacobs et al., 2007), and those which arise from a more indirect mechanism such as a rapid change of direction or a sudden stop (Papadopoulos, Nicolopoulos, Anderson, Curran & Athanasopoulos, 2005; Stasinopoulos, 2004). The anterior talofibular ligament (ATFL) is injured first, followed by injury to the calcaneofibular ligament (CFL) and posterior talofibular ligament (PTFL) (Puffer, 2001) (Figure 2.1). This stress can damage not only the ligaments, but can also lead to peroneal muscle strains, dislocated peroneal tendons, syndesmosis sprains, and damage to the nerves and mechanoreceptors that are located around the lateral aspect of the ankle (Docherty, Arnold & Hurwitz, 2006; Puffer, 2001).

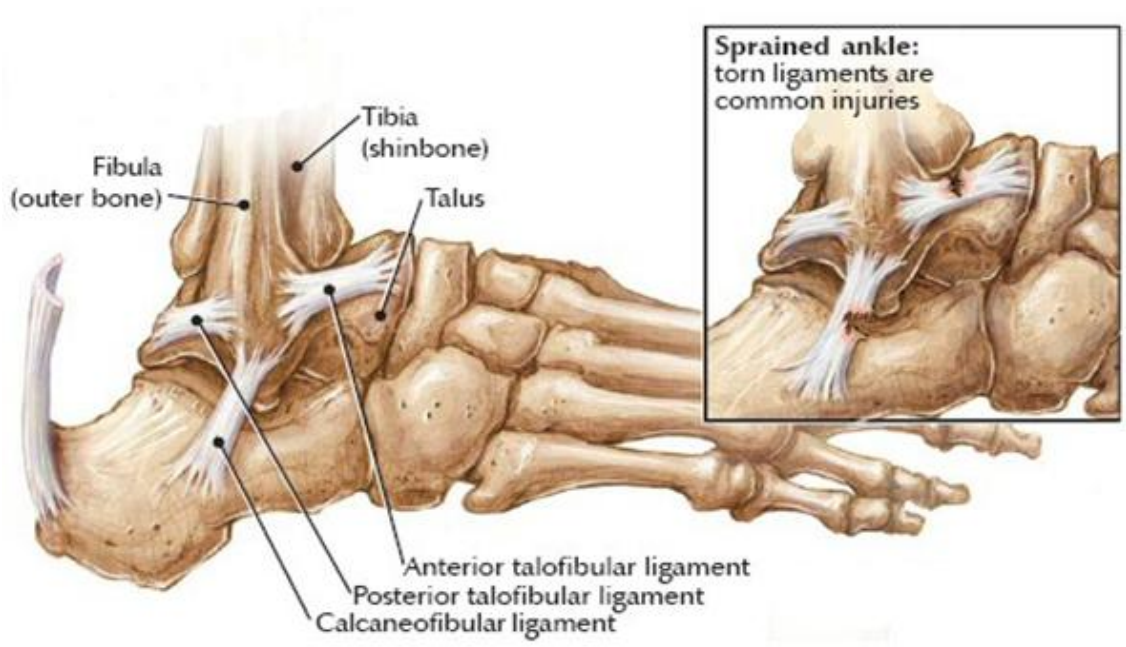


Figure 2.1. Lateral View of the Ankle Showing the Anterior Talofibular Ligament, Calcaneofibular Ligament and Posterior Talofibular Ligament (Marieb, Wilhelm & Mallatt, 2012).

2.4 Chronic Ankle Instability

The mechanism of recurrent ankle injury is not thought to be different than that of initial acute ankle sprains; however, adverse changes that occur after primary injury are believed to predispose individuals to recurrent sprains (Hertel, 2002). Two theories of the cause of chronic ankle instability (CAI) have traditionally been postulated: mechanical instability and functional instability (Riemann, 2002; Riemann & Lephart, 2002). These two terms, however, are probably not mutually exclusive entities but more likely form a continuum of pathologic contributions to CAI (Hertel, 2002).

2.4.1 Mechanical Instability

Mechanical instability of the ankle complex occurs as a result of anatomic changes after initial ankle sprains, which lead to insufficiencies that predispose the ankle to further episodes of instability (McGuine et al., 2000). These changes include pathologic laxity, decreased dorsiflexion range of motion, synovial inflammation and impingement, and the development of degenerative joint disease (Hertel, 2002). Mulligan (1995) suggested that individuals with FAI may have an anteriorly and inferiorly displaced distal fibula which means the ATFL may be more slack in its resting position. Thus, the talus can go through a greater range of motion before the ATFL becomes taut and this may predispose individuals with CAI to recurrent episodes of instability (Mulligan, 1995).

2.4.1.1 Dorsiflexion Range of Motion

Diminished dorsiflexion following a lateral ankle sprain is thought to contribute to FAI (Hertel, 2000). Inflexibility of the triceps surae prevents the ankle from reaching full dorsiflexion and as a result the ankle is held in a more plantarflexed position throughout the gait cycle (Delahunt, Monaghan & Caulfield, 2006a). The talocrural joint is in its closed packed position in full dorsiflexion, thus the talus is able to invert and internally rotate more when it is not in full dorsiflexion at heel strike (Hertel, 2000). Therefore, this excess motion may predispose individuals with diminished dorsiflexion to recurrent FAI (Delahunt et al., 2006a; Hertel, 2000).

Leanderson, Eriksson and Nemeth (1993) studied a population of professional basketball players with bilateral FAI and found that they were shown to have a mean of

3.6° of passive dorsiflexion, while healthy control individuals had a mean of 17.9° of dorsiflexion. In agreement, it has been found that a group of dancers with FAI showed significantly less dorsiflexion range of motion than their uninjured counterparts (Wiesler, Hunter & Martin, 1996). It has also been reported that people with inflexible ankles (34° dorsiflexion range of motion) have nearly five times the risk of ankle sprain of people with average flexibility (45° dorsiflexion range of motion) (Pope, Herbert & Kirwan, 1998).

2.4.2 Functional Instability

2.4.2.1 Mechanoreceptors

Various mechanoreceptors are present in joint capsules, ligaments, muscles, and skin around the ankle (Freeman, Dean & Hanham, 1965; Richie, 2001). In order for the nervous system to properly control skeletal muscle movements, it must receive continuous sensory feedback from the contracting muscle (Powers & Howley, 2004). This sensory feedback includes (1) information concerning the tension developed by a muscle and (2) an account of the muscle length (Powers & Howley, 2004). Golgi tendon organs provide the central nervous system with feedback concerning the tension developed by a muscle, while the muscle spindle provides sensory information concerning the relative muscle length (Powers & Howley, 2004) (Figure 2.2).

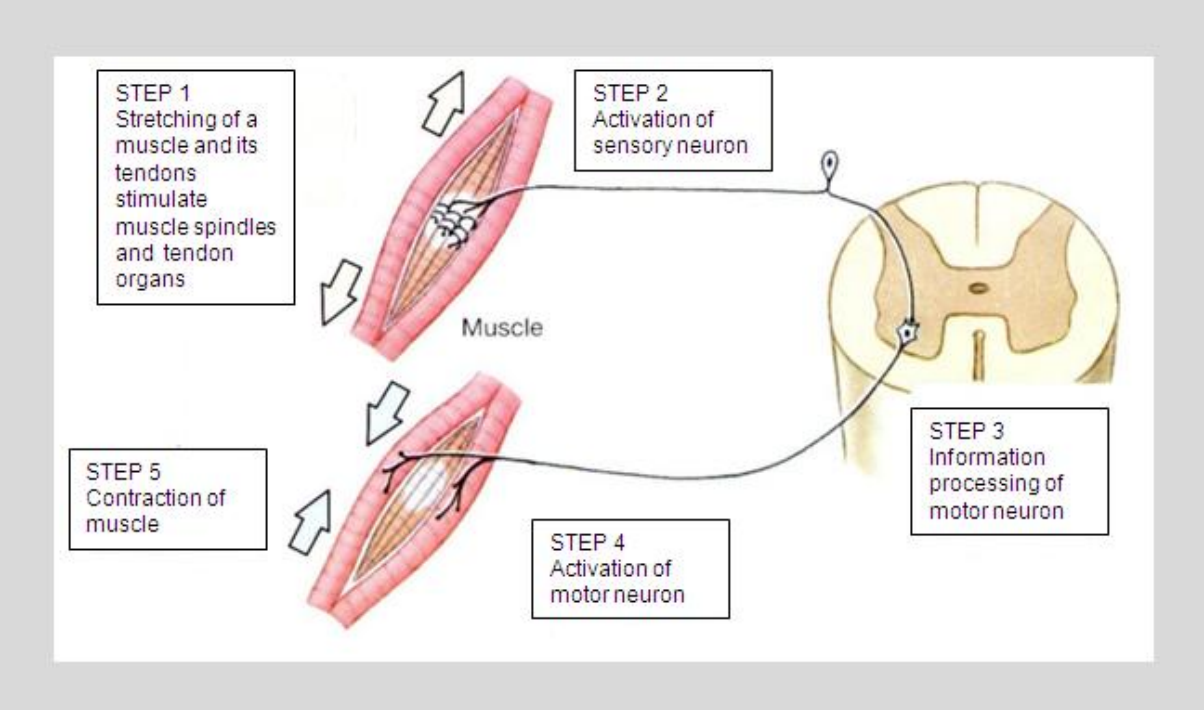


Figure 2.2. Protective Mechanism of Muscle Spindles and Golgi Tendon Organs to Prevent Damage of the Muscle via the Spinal Reflex Arc (Marieb et al., 2012).

2.4.2.2 Muscle Spindles

Muscle spindles are found in large numbers in most human locomotor muscles, and muscles that require the finest degree of control, such as the muscles of the hands (Powers & Howley, 2004). Muscle spindles contain 2 types of sensory nerve endings (Palastanga, Field & Soames, 2002). The primary endings respond to dynamic changes in muscle length. The second type of sensory ending is called the secondary ending, and it does not respond to rapid changes in muscle length, but provides the central nervous system with continuous information concerning static muscle length (Powers & Howley, 2004). The function of the muscle spindle is to assist in the regulation of movement and to help maintain posture (Palastanga et al., 2002; Powers & Howley, 2004). This is accomplished by the muscle spindles ability to detect changes in the

length of skeletal muscle fibres and cause the central nervous system to respond to these changes (Powers & Howley, 2004).

2.4.2.3 Golgi Tendon Organs

The Golgi tendon organs continuously monitor the tension produced by a muscle contraction (Palastanga et al., 2002). Golgi tendon organs are located within the tendon. In essence, the Golgi tendon organs act as 'safety devices' that help to prevent excessive force during muscle contraction (Palastanga et al., 2002). When activated, Golgi tendon organs send information to the spinal cord via sensory neurons, which in turn excite inhibitory neurons (Powers & Howley, 2004). An inhibitory disynaptic reflex helps prevent excessive muscle contractions and provides a finer control over skeletal movements (Powers & Howley, 2004).

2.4.2.4 Muscle Latency Deficits

During a joint perturbation, reflexive muscle activity occurs in response to stimulation of mechanoreceptors within ligaments and muscles (Hogervorst & Brand, 1998; Sainburg, Poizner & Ghez, 1993), presumably to reduce the magnitude of joint movement (Lynch, Eklund, Gottlieb, Renström & Beynnon, 1996). The time between a perturbation and reflexive muscle activation is known as the latency period (Ebig et al., 1997; Lynch et al., 1996; Nawoczenski, Owen, Ecker, Altman & Epler, 1985), which is essentially the duration of a muscles stretch reflex. It has been stated that a deficiency of the muscle activation in response to a sudden unexpected perturbation could compromise joint stability (Delahunt, 2007a).

Articular deafferentiation which was proposed by Freeman et al. (1965) stated that the basic mechanism of ankle instability following ankle injury develops due to the lesion of mechanoreceptors in the joint capsule and ligaments surrounding the ankle. According to this theory, dynamic stability of the ankle joint is dependent on the ability of the evertors (peronei) to react quickly to sudden inversion perturbations, to develop sufficient tension to prevent injurious ranges of ankle motion, and thus prevent a lateral ligament sprain (Freeman et al., 1965). It has been suggested that an increase of the response time of the peronei to sudden inversion may have highly significant consequences in terms of risk of injury to the lateral ligaments of the ankle (Wilkerson & Nitz, 1994).

Isakov, Mizrahi, Solzi, Susak and Lotem (1986) used a tilting platform to induce sudden unexpected inversion while concomitantly recording peroneal electromyographic (EMG) activity. The results found no significant differences in peroneal reaction time when comparing a healthy control group to a group of subjects with FAI (Isakov et al., 1986). However, disagreements have been stated in the literature as Konradsen and Ravn (1990) used a similar testing method to Isakov et al. (1986) and found that the FAI group exhibited significantly longer peroneus longus and peroneus brevis reaction times (82 ms and 84 ms, respectively) to the inversion stress when compared to a healthy control group (65 ms and 69 ms, respectively) (Konradsen & Ravn, 1990).

Karlsson and Andreasson (1992) compared the reaction times of the peronei to a sudden inversion stress and found that the involved limbs of individuals with unilateral FAI demonstrated significantly longer peroneus longus (84.5 ms versus 68.8 ms) and peroneus brevis (81.6 ms versus 69.2 ms) reaction times when compared to the

uninvolved healthy contra lateral limb (Karlsson & Andreasson, 1992). However, there are disagreements in the literature as Ebig et al. (1997) examined the EMG response time of the peroneal and tibialis anterior muscles in response to a sudden plantarflexion/inversion stress in subjects with unilateral FAI. The results of this study indicated no significant differences between the injured and uninjured ankle in subjects with unilateral FAI for the reaction time of the peroneal and tibialis anterior muscles (Ebig et al., 1997).

Studies which have found a delay in the peroneal reaction time (Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990) are in agreement with the finding that slower motor nerve conduction velocities of the peroneal nerve were shown after inversion trauma (Kleinrensink, Stoeckart & Meulstee, 1994). The principle evertor muscles (peroneus longus and peroneus brevis) are innervated by the superficial peroneal nerve, whereas, the invertor muscles (tibialis anterior) are innervated by the deep peroneal nerve (Cingel, Kleinrensink, Uitterlinden, Rooijens, Mulder, Aufdemkampe et al., 2006). The superficial peroneal nerve rather than the deep peroneal nerve is more likely to be affected by inversion trauma, because of its position in respect to the inversion-eversion axis (Cingel et al., 2006). Therefore, delay of neuromuscular response can be expected in the muscles innervated by the superficial peroneal nerve (Cingel et al., 2006).

The ability of the peroneal musculature to provide dynamic ankle protection against sudden unanticipated ankle inversion has been questioned (Ashton-Miller et al., 1996; Konradsen, Voigt & Hojsgaard, 1997). After the delay due to neural latencies, which typically range from 85 to 90 ms until myoelectric activity is first observed, there is an

additional delay because muscle contractile mechanics dictate that a further 90 ms is required by a muscle to develop contractile force to even half maximal levels (Ottaviani, Ashton-Miller, Kothari & Wojtys, 1996). Konradsen et al. (1997) has shown that unlimited subtalar inversion from a standing position would put the lateral ligament complex at risk of sprain after approximately 100 ms. Thus, Konradsen et al. (1997) concluded that the ankle musculature cannot react fast enough to protect an ankle from injury in the case of sudden unexpected inversion stress. It has been suggested that due to the time delays mentioned above (85 and 90 ms), that the evertor musculature must be activated prior to the onset of the external forces during ground contact to provide dynamic ankle stability (Ashton-Miller et al., 1996; Konradsen et al., 1997). In support of this theory, McKinley and Pedotti (1992) suggested that greater preparatory muscle activity, as measured by EMG, would provide a better pre-programmed dynamic defense mechanism, thus minimising dynamic postural stability scores. Specifically, subjects with greater and earlier co-contraction of lower leg muscle before landing from a jump displayed lower time-to-stabilisation scores (McKinley & Pedotti, 1992).

2.4.2.5 Studies Refuting Articular Deafferentiation

More recent studies suggest that the theory of articular deafferentiation may not be the main physiological mechanism underlying the development of FAI (Konradsen, Ravn & Sorensen, 1993). Konradsen et al. (1993) investigated the peroneal reflex reaction time to sudden ankle inversion before and after regional block of the ankle and foot with local anaesthetic. The anaesthesia totally blocked the afferent input from mechanoreceptors in the ligaments and capsule of the ankle (Konradsen et al., 1993). The peroneal reaction time to sudden ankle inversion was not altered (80 ms before and 83 ms under

anaesthesia). Articular deafferentiation made no difference to response time. Konradsen et al. (1993) concluded that afferent input from the active calf musculature is responsible for dynamic ankle protection against sudden ankle inversion stress.

Riemann, Myers, Stone and Lephart (2004) examined the effect of an experimentally induced anaesthesia of the anterior talofibular ligament and calcaneofibular ligament on postural stability during a single leg stance as well as during a single leg step down landing task. Results failed to demonstrate a difference between the control and the experimental conditions (Riemann et al., 2004). Thus, the authors concluded that articular deafferentiation may not be the process by which subjects develop functional instability of the ankle joint, and that other factors such as central motor programming may be more important (Riemann et al., 2004).

2.4.2.6 Postural Sway Deficits

Mechanoreceptors are responsible for providing afferent information regarding joint movement and position (Delahunt, 2007a). Several authors have suggested that damage to these mechanoreceptors following a lateral ankle sprain, may interrupt the flow of these afferent impulses into the central nervous system, and therefore lead to balance deficits, and contribute to the development of FAI (Freeman, 1965a; Freeman, 1965b; Freeman et al., 1965).

Another explanation is that following an inversion stress to the ankle joint, the mechanoreceptors located within the ligaments and joint capsule may become stretched (Docherty, Arnold et al., 2006). This potentially means that if the

mechanoreceptors become permanently lengthened, the protective control mechanism from the muscles and nerves to prevent inversion occurring will also become delayed, and so there is a higher possibility of an inversion sprain occurring (Ross, Guskiewicz & Yu, 2005).

Garn and Newton (1988) found that there was a higher incidence of balance deficits noted whilst FAI subjects stood on their injured leg. In a study by Lentell, Katzman and Walters (1990) 15 (45%) subjects demonstrated no differences in balance from one limb to the other. The remaining 18 (55%) subjects did demonstrate notable balance deficits from one side to the other. In 17 of the 18 subjects the deficit was found when standing on their injured limb. Mulloy-Forkin, Koczur, Battle and Newton (1996) also found that 63% of gymnasts with FAI showed balance deficits during an eyes closed single leg stance task when standing on their injured limb.

Tropp, Odenrick & Gillquist (1985) used stabilometry to try and objectively quantify the association between balance deficits and FAI. Centre of pressure (COP) excursions were studied during a single leg stance with eyes open. The study reported that soccer players with a history of FAI showed significantly higher COP excursions when compared to a healthy control group (Tropp et al., 1985). In a subsequent study, Tropp (1986) reported that there were no significant differences in COP excursions between the affected and unaffected legs of soccer players with unilateral FAI. However, a comparison of both legs of the FAI group with a healthy non injured control group revealed significantly higher COP excursion values (Tropp, 1986).

In contrast to these results several studies have failed to show differences in balance performances between FAI subjects and healthy controls (Delahunt, 2007b). Baier and Hopf (1998) studied the COP excursions in a group of athletes with FAI compared to healthy controls. The authors did not find any significant differences in balance performance between the two groups (Baier & Hopf, 1998). Several other studies have also failed to show balance deficits in FAI subjects compared to healthy subjects during a single leg stance (Bernier, Perrin & Rijke, 1997; Delahunt, 2007b; Kinsella & Harrison, 1998).

When looking at the above studies one must be aware that the one legged balance test is a relatively static task (Delahunt, 2007b). It has been suggested that most joint receptors are only active near the end of the range of motion, and a more dynamic method may be necessary for neural discharge of joint mechanoreceptors (Wilkerson & Nitz, 1994). Various authors have endeavoured to address this issue (Delahunt, 2007b). Olmstead, Carcia, Hertel and Shultz (2002) used the star excursion balance test (SEBT), to detect balance deficits in subjects with ankle instability. They found that the ankle instability subjects demonstrated significantly decreased reaching distances when balancing on their injured limb when compared to their healthy limb, and when compared to a healthy control group. Olmstead et al. (2002) concluded that static tests such as the single leg balance test may not be sensitive enough to detect motor control deficits related to balance performance, and that dynamic tests like the SEBT provide a means of identifying functional deficits related to balance performance in subjects with ankle instability. It has been suggested that static conditions such as single leg balance tests may fail to elicit postural control deficits due to the ease of the testing procedure (Riemann, Guskiewicz & Shields, 1999).

2.4.2.7 Strength Deficits

In the literature a common cause of FAI involves weakness of the peroneal muscles (Delahunt, 2007b; Willems et al., 2002). The peronei act as the primary evertors of the ankle, and hence weakness of this muscle group may impair the ability to dynamically control inversion stresses, thus rendering the ankle vulnerable to inversion sprain (Delahunt, 2007b). However, some of the evidence supporting weakness of the peroneal muscles involves manual muscle testing, which is a crude and subjective form of muscle strength evaluation (Munn, Beard, Refshauge & Lee, 2003). Strength training of the peronei forms a central component of treatment programs for FAI (Caulfield, 2000), thus indicating that peronei weakness is regarded by many clinicians as a significant factor in the development of functional problems following lateral ligament ankle sprains (Delahunt, 2007b). Supporting this view, Willems et al. (2002) used an isokinetic dynamometer to determine peak torque and peak torque/body weight for concentric and eccentric eversion-inversion movements of the ankle. The authors found that subjects suffering from FAI showed significant weakness in evertor muscle strength, compared to a group of healthy controls (Willems et al., 2002).

Ryan (1994) failed to show the presence of evertor strength deficits in a group of subjects with unilateral FAI. Interestingly, however, there was a decrease in the peak torque of the ankle invertors (tibialis anterior, tibialis posterior, extensor hallucis longus, flexor hallucis longus, flexor digitorum longus) on the injured side when compared to the non injured side (Ryan, 1994). Wilkerson, Pinerola and Caturano (1997) also demonstrated significant invertor deficits for both peak torque and average power using the isokinetic dynamometer at speeds of 30°/s and 120°/s.

There is research to suggest that the presence of eccentric invertor strength deficits may play a role in the development of residual symptoms following lateral ligament sprains. The presence of invertor strength deficits has been suggested to occur due to selective inhibition (Ryan, 1994). The process of selective inhibition was described by Swearingen and Dehne (1964), who postulated that decreased stress tolerance of an injured joint triggers reflexive mechanisms which inhibit muscles that are capable of increasing tensile stress on damaged ligaments. Thus, the invertors of the ankle may be inhibited due to their ability to initiate movement in the direction of the injury (Swearingen & Dehne, 1964).

2.4.2.8 Joint Position Sense

Jerosch and Bischof (1996) reported that subjects with a history of recurrent ankle sprains exhibited a deficit in active replication of joint position sense in the inversion range of motion; while Boyle and Negus (1998) observed a deficit in passive joint position replication in the plantarflexion-inversion range of motion. Correct positioning of the foot is very important in gait and sports. Hitting the ground in an overly inverted position could result in spraining the ankle. It has been suggested that subjects with FAI who exhibit a deficit in the replication of active and/or passive joint position sense may have inappropriate foot positioning (Willems et al., 2002). Because of the altered afferent input, these subjects may be more susceptible to ankle reinjury (Willems et al., 2002).

2.4.2.9 Altered Arthrokinematics and Arthrokinetics

It has been suggested that inappropriate positioning of the foot and ankle complex could be a contributing factor to the development of FAI (Tropp, 2002). Functional ankle instability subjects were found to have a more inverted position of the ankle joint prior to and immediately following heel strike compared to a non injured healthy control group (Monaghan, Delahunt & Caulfield, 2006). It has been suggested that these differences in inversion between the FAI and the healthy control group could leave the FAI group vulnerable to inversion sprain (Delahunt, 2007a).

2.5 Functional Ankle Instability Determination

Many researchers have investigated patients with FAI, however, very few have determined this instability by the use of validated questionnaires. The use of questionnaires may create a more objective method of identifying patients suffering from FAI. However, with this in mind, many researchers still appear to develop their own criteria to determine FAI, which often makes it difficult to compare studies as subjects may vary considerably.

Some questionnaires have used graded scales to determine the level of disability caused by FAI. Wikstrom, Bishop, Inamdar and Hass (2010) used the Ankle Joint Functional Assessment Tool (AJFAT), which is a 48 point scale that can be used to indicate self-assessed instability of the involved limb. Hale and Hertel (2005) used the Functional Ankle Disability Index (FADI) and the FADI Sport. These are 104 and 32 point scales, respectively, in which lower scores represent greater instability. Hiller,

Refshauge, Bundy, Herbert & Kilbreath (2006) devised the Cumberland Ankle Instability Tool (CAIT), where self reported scores were based on the patients graded response, and were summated to generate a total score. A reported problem with using graded scale questionnaires is that a variety of disabilities may be identified, with some individuals suffering certain symptoms that others do not.

With this is mind, Hubbard and Kaminiski (2002) developed the Functional Ankle Instability Questionnaire (FAIQ), which included two parts; part 1 was an ankle instability questionnaire where subjects had to specifically answer yes to questions 3, 5, 6, 7 and 9, and no to questions 4, 8 and 10, part 2 was a clinical examination of ankle stability, which included the anterior drawer and talar tilt tests to help rule out mechanical instability of the ankle joint. The FAIQ was shown to be a valid method which would, if used by several different examiners, determine which subjects suffered from FAI and which subjects did not (Kaminski, Buckley, Powers, Hubbard & Ortiz, 2003).

In addition to these questionnaires some authors appear to develop their own criteria to determine FAI (Demeritt, Shultz, Docherty, Gansneder & Perrin, 2002; Mazaheri, Negahban, Salavati, Sanjari & Parnianpour, 2010). These authors have listed inclusion criteria such as “more than one repeated injury on the same ankle”, or “the perception of the ankle giving way”, and “no present participation in a rehabilitation programme”. These criteria often miss out vital signs and symptoms associated with FAI, and therefore it has to be questioned whether the subjects used in these studies actually suffered from FAI.

2.6 Muscle Latency during Tilt Platform Perturbations

The latency of muscular activity to involuntary perturbation is used to assess the efficiency of the spinal reflex pathway (Lephart, Pincivero & Rozzi, 1998). The role of this neuromuscular pathway during sudden unexpected movements has been evaluated by EMG analysis of muscle response times to involuntary perturbations (Bressel et al., 2007). The time between a perturbation and reflexive muscle activation is known as the latency period (Ebig et al., 1997; Lynch et al., 1996; Nawoczenski et al., 1985), which is essentially the duration of a muscle's stretch reflex (Kernozek, Durall, Friske & Mussallem, 2008). It has been stated that a deficiency of the muscle activation in response to a sudden unexpected perturbation could compromise joint stability (Delahunt, 2007a).

In addition to the latency period is the electromechanical delay (EMD), the delay between muscle activation and the production of tension at the muscle's skeletal attachments (Alexander & Bennet-Clarke, 1977). This lag occurs because time is required for the action potentials propagation along the sarcolemma, the excitation-contraction coupling process, and the removal of slack in the elastic elements (Alexander & Bennet-Clarke, 1977; Cavanagh & Komi, 1979). If the combined muscle latency and EMD are shorter than the time it takes for the ankle joint to reach its physiological motion limits, the muscles may help to decelerate ankle joint movement and reduce ligamentous sprain (Kernozek et al., 2008).

2.6.1 Effect of Age on Latency Times

Wilkinson and Allison (1989) looked at the regression of age upon reaction time in 5,325 subjects. It was found that 20-29 year olds had the fastest reaction times compared to other age groups. The average fastest reaction time in the 20-29 year olds was approximately 200 ms (Wilkinson & Allison, 1989). It was found that reaction time was fastest in the 20's, declining rapidly below that age and more gradually above it, such that the 20's were significantly faster than the teens and under 10's, but when compared to the older age groups they were only significantly faster than the decades 50 and above (Wilkinson & Allison, 1989).

2.6.2 Measurements of Muscle Latency Using a Tilt Platform

It has often been criticised that exact mechanisms of injury are difficult to recreate in the laboratory. Injuries rarely occur with a person standing at rest. However, to make comparisons there has to be standardisation (Lynch et al., 1996). Standing at rest with equal body weight distribution is the safest and most reproducible posture available in the laboratory. Tilt magnitude presents another variable; what may represent severe inversion trauma for one individual may be unstressful for another. Again, to make comparisons standardisation is necessary (Lynch et al., 1996).

The degrees of inversion movement of the tilt platform have varied greatly from 15° to 50° (Akhbari, Takamjani, Salavati & Sanjari, 2007; Eechaute, Vaes, Duquet & Gheluwe, 2007; Gruneberg, Nieuwenhuijzen & Duysens, 2003; Hiller, Refshauge, Herbert & Kilbreath, 2007; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990;

Papadopoulos, Nicolopoulos, Baldoukas, Anderson & Athanasopoulos, 2005). The degrees of plantarflexion on the tilt platform have also varied from 20° to 42° (Akhbari et al., 2007; Tohyama, Yasuda, Beynnon & Renström, 2006). One study did not include the plantarflexion movement on the tilt platform and only studied inversion (Akhbari et al., 2007). However, results should be interpreted with caution as it has been stated that a combination of plantarflexion and inversion should be used to replicate the true mechanism of injury of a lateral ankle sprain (Akhbari et al., 2007).

Additionally, the distribution of body weight on the tilt platform may influence the angular velocity of the tilt perturbation (Benesch, Putz, Rosenbaum & Becker, 2000). Some studies asked the subject to evenly distribute their weight across both feet (Fritschy, de Reynier & Blanc, 1988; Konradsen & Ravn, 1990; Lynch et al., 1996). One study even used a pair of scales to equally distribute body weight (Kernozek et al., 2008). Other studies asked subjects to place a higher percentage of their body weight onto a particular foot (usually the foot being tested) (Benesch et al., 2000; Isakov et al., 1986; Morey-Klapsing, Arampatzis & Bruggemann, 2004). Due to the differences in methodology, comparison of the results should be made with caution (Delahunt, 2007a).

2.7 Postural Control

Postural control or balance can be defined statically as the ability to maintain a base of support with minimal movement (Winter, Patla & Frank, 1990). Bressel et al. (2007) defined dynamic postural control as the ability to perform a task while maintaining a stable position. Whereas, Wikstrom, Tillman, Chmielewski & Cauraugh (2007) defined dynamic postural stability as maintaining balance while transitioning from a dynamic to a

static state. In order to maintain postural control, the body is in a state of continuous movement, adjusting to keep the centre of gravity over the base of support (Olmsted, Carcia, Hertel & Shultz, 2002; Ross et al., 2005). With respect to postural control afferent information arises from vestibular, visual and somatosensory sources (Maurer, Mergner & Peterka, 2006). Next, the afferent information gathered from these sources must be integrated and processed in the central nervous system to determine the necessary motor commands (Fukuoka, Nagata, Ishida & Minamitani, 2001). The motor commands are then executed by the muscles of the trunk and extremities in order to maintain postural stability (Docherty, Arnold et al., 2006; Riemann, 2002).

Postural sway deficits have been identified frequently in individuals suffering from FAI (Hertel, 2002; Leanderson et al., 1999). McHugh, Tyler, Tetro, Mullaney and Nicholas (2006) found that subjects who demonstrated high postural sway scores had nearly seven times as many ankle sprains as subjects who had low postural sway scores. Postural sway deficits appear to be present in some individuals prior to injury and may actually predispose these individuals to injury (McGuine et al., 2000). However, there are inconsistencies in the literature as a study by Gribble, Radel and Armstrong (2006) failed to show postural sway differences in individuals with FAI.

2.7.1 Differences between Centre of Mass and Centre of Pressure

For many individuals in the applied and clinical areas, the terms centre of mass (COM) and centre of pressure (COP) are often misinterpreted or interchanged (Winter, 2005). The COM of the body is the net location of the centre of mass in three-dimensional space. The location of the COM in the vertical direction is sometimes called the centre

of gravity (COG). The trajectory of this vertical line from the COM to the ground allows us to compare the trajectories of the COM and COP. The trajectory of the COP is totally independent of the COM, and it is the location of the vertical ground reaction force vector from a single force platform, assuming that all body contact points are on the platform. The vertical ground reaction force is a weighted average of the location of all downward (action) forces acting on the force plate (Winter, 2005). These forces depend on the foot placement and the motor control of the ankle musculature. Thus, the COP is the neuromuscular response to the imbalances of the body's COM. Winter (2005) states that the major misuse of the COP comes from researchers who refer to the COP as "sway", thereby inferring it to be the kinematic measure of COM. Postural "sway" is actually proportional to the acceleration of the COP. However, it should be noted that the COP excursion distance was used throughout this thesis as an indirect measure of postural sway.

2.7.2 Measuring Postural Control

The majority of studies to date have identified postural sway deficits in participants with FAI using stabilometric devices such as force platforms. These range from the New Balance Master (McGuine et al., 2000), to the Bertec strain gauge (Hertel, Buckley & Denegar, 2001; Ross & Guskiewicz, 2004), to the Smart EquiTest System (Fu & Hui-Chan, 2005), to the AMTI force platform (Ekdahl, Jarnlo & Andersson,1989) and the Kistler force platform (Mitchell, Dyson, Hale & Abraham, 2008b). These devices have been shown to have a very high sensitivity in detecting even the smallest changes in postural sway (Docherty, Gansneder, Arnold & Hurwitz, 2006; McHugh et al., 2006).

Some studies have used more clinical measures such as a single leg balance test (Docherty, Valovich-McLeod & Shultz, 2006; Hertel et al., 2001). Trojian and McKeag (2006) reported that the single leg balance test could be used to identify athletes with an increased risk of ankle sprains; however, Hertel et al. (2001) showed that results from the single leg balance test can lack sensitivity when evaluating small changes in postural sway.

Olmsted et al. (2002) attempted to evaluate participants with FAI using the SEBT. This is an objective measure of lower extremity maximal reach that is performed while maintaining a single leg balance with the contra-lateral limb. Their findings stated that subjects with FAI had deficits in postural control. However, once again this method has been criticised for lacking sensitivity when recording small changes in postural sway (McHugh et al., 2006).

2.7.3 Static and Dynamic Measures of Postural Control

It has been shown in much of the literature that to challenge the postural control system testing should be performed in a weight bearing position (Hume & Gerrard, 1998; Masharawi, Carmeli, Masharawi & Trott, 2003; Willems et al., 2002). This also creates a more functional test, as ankle sprains are most likely to occur in a weight bearing position during sport (Willems et al., 2002). However, there are many disagreements in the literature as to the best method to test postural sway (Hertel et al., 2001; Ross et al., 2005).

The single leg stance position that was often used in the literature to challenge postural control varies greatly. Docherty, Valovich-McLeod et al. (2006) instructed participants to stand on one leg and place their hands on their hips. Hertel et al. (2001) instructed participants to place their arms folded across their chest, the non stance leg was held in approximately 30° of hip flexion and 45° of knee flexion and was not allowed to touch the stance leg. However, the above authors fail to rationalise the use of these stance positions. Ross and Guskiewicz (2004) instructed subjects to remain as motionless as possible but did not control for arm position, trunk flexion or lower extremity flexion during the stance. If postural stability is compromised during sporting activity the athlete will adopt the most appropriate position for them to remain balanced, so a more realistic situation is recreated during testing, which enhances ecological validity (Olmsted et al., 2002; Ross et al., 2005).

Many authors have used the single leg balance test, and have stated that this is a repeatable measure which is sufficient enough to challenge the postural sway system (Baier & Hopf, 1998; Trojian & McKeag, 2006). However, this method has been criticised by many authors who have stated that static measures of postural stability cannot be used to detect functional ankle deficits (Hertel et al., 2001; Riemann, 2002; Ross & Guskiewicz, 2004). Maintaining a single leg stance places relatively small strength demands on the lower extremity musculature, and range of motion requirements are lower than when performing more dynamic tasks (Olmsted et al., 2002). So even if a previously injured athlete shows normal postural sway on their injured ankle during a single leg balance, these athletes may still be predisposed to recurrent episodes of ankle instability during more dynamic activities (Hertel et al., 2001).

More functional tests have therefore been adopted to test postural sway such as a drop jump onto the force plate (Ross & Guskiewicz, 2004). This has been shown to demonstrate a more dynamic and functional component that can be related to sport as it requires the athlete to land, decelerate, and quickly stabilise (Olmsted et al., 2002; Ross & Guskiewicz, 2004). These tests have been shown to be more sensitive in detecting functional deficits in the lower extremity during dynamic activities and may be more useful in predicting the risk of individual athletes for recurrent ankle sprains (Hertel et al., 2001; Ross et al., 2005).

Several authors have found that postural stability decreases after lateral ankle ligament injury (Freeman, 1965a; Tropp & Odenrick, 1988; Zatterstrom, Frieden & Lindstrand, 1994). However, this finding is controversial in the literature and there is no relationship between static (e.g. COP measurements) and dynamic (e.g. SEBT) measurements of postural sway (Riemann, 2002). A possible reason for this can be found in the type of mechanoreceptor that these protocols stimulate. Centre of pressure scores, measured in a static leg stance, are dependent on not only visual and vestibular information but information from the slow adapting mechanoreceptors as well (Wikstrom, Tillman, Chmielewski & Borsa, 2006). However, dynamic joint stability tests are functional and stimulate the fast adapting mechanoreceptors of the lower extremity, thus testing the sensitivity of different mechanoreceptors (Wikstrom et al., 2006).

2.7.4 Centre of Pressure Measurements

Functional instability is a subjective measure of FAI (Riemann, 2002). Two very common dependent measures of postural control include the length of the path of the

COP and the velocity of COP excursions (Evans, Hertel & Sebastianelli, 2004). Shorter length of COP displacement and slower velocity of COP excursions are associated with better postural control (Riemann, 2002). It has commonly been found that FAI subjects have decreased postural stability compared with healthy controls (Hertel, 2002; Leanderson et al., 1999; McHugh et al., 2006), these FAI subjects frequently have shown greater COP displacements and faster velocity of COP excursions (Evans et al., 2004; Hertel et al., 2001).

2.7.5 Ankle and Hip Strategies

When balancing in a single limb stance the foot pronates and supinates in an effort to keep the body's centre of gravity above the base of support, which is referred to as the ankle strategy (Hertel, 2002). When responding to larger postural displacements, the primary action of most people occurs at the hip resulting in active trunk rotation, or the so-called 'hip strategy' (Nasher & McCollum, 1985). The choice of a postural strategy to disturbance was found to depend on the available appropriate sensory information (Nasher, Shupert, Horak & Black, 1989). Individuals with FAI have been shown to use more of a hip strategy to maintain unilateral stance (MacKinnon & Winter, 1993).

Friel et al. (2006) found weaker hip abductors in the involved limb of people with FAI. In support of this finding Bullock-Saxton, Janda and Bullock (1994) postulated that altered sensation in one joint can lead to muscle function changes in another, more proximal joint. If the firing, recruitment and strength of the hip abductor muscles in people with FAI have been altered because of the distal injury, the frontal plane stability normally

supplied by this muscle is lacking, and the risk for repeated injury increases (Bullock-Saxton et al., 1994).

During gait the hip abductors initiate a lateral pelvic tilt during early double support in response to the lateral displacement of the mass of the head, arms and trunk (Mackinnon & Winter, 1993). For the remainder of stance, the hip abductors work to control the lateral pelvic tilt (MacKinnon & Winter, 1993). Hence, in the presence of hip abductor muscle weakness, the position of the foot at initial contact may be more adducted than normal. MacKinnon and Winter (1993) found that the body uses several strategies to control body balance, and both distal and proximal components contribute to the fine tuning of the centre of mass location as it relates to the support limb. In addition to hip abductor weaknesses, increases in subtalar inversion were associated with decreased hip abduction (MacKinnon & Winter, 1993).

The finding of central adaptations in subjects with FAI was confirmed by the work of Gauffin, Tropp and Odenrick (1988) and Tropp and Odenrick (1988), who demonstrated an impaired postural control not only at the injured side but also at the non injured side in these subjects. Van Deun, Staes, Stappaerts, Janssens, Levin & Peers (2007) found that during the transition from a double leg stance position to a single leg stance position there was a later onset of the hip and hamstring muscles in subjects with FAI compared to healthy control subjects. The authors concluded that impairments in muscle activation are not only present in structures around the injured ankle but also exist around other joint complexes. The authors concluded that one possible explanation is that the central nervous system decreases the reliance on proprioceptive information from one location where this source of information is confounded, and

increases the reliance on input from other locations that provide reliable information for maintaining postural balance. This has been defined as sensory re-weighting (Van Deun et al., 2007). Van Deun et al. (2007) stated that one way to compensate for the impairments at the ankle could be to increase the reliance on proprioceptive information from the knees and hips. The observed strategy may arise from learned changes to patterns of movement control, as a result of previous injury, or could have existed before, leading to injury and re-injury (Van Deun et al., 2007).

2.8 Peripheral versus Central Control Mechanisms

Many studies, which have looked at subjects suffering from FAI have identified unilateral differences of increased postural sway on the injured side (Harkins et al., 2005; Olmsted et al., 2002). Hale et al. (2007) found that subjects with FAI demonstrated deficits in postural control and during the SEBT on the injured limb but not the contra-lateral healthy limb. These findings conform with Mitchell et al. (2008a) who used a tilt platform to look at muscle reaction times in subjects with unilateral FAI. When the functionally unstable ankle was used as the support limb, the reaction times of extensor digitorum longus ($P = 0.032$) and tibialis anterior ($P = 0.017$) were significantly slower than when the healthy limb was used as the support limb. The above findings would support the peripheral control mechanism for postural control, as only the injured side was affected (Olmsted et al., 2002).

However, many studies have identified bilateral deficits in individuals with unilateral FAI (Caulfield & Garrett, 2002; Hertel, Buckley & Denegar, 2001). Tropp et al. (1984) found that subjects with FAI did not differ in unilateral stance abilities on the injured versus the

uninjured ankles. Though a comparison of both limbs in the subjects with FAI with a healthy control group revealed significantly higher centre of pressure excursions. This result immediately offers two interpretations: (1) the patients with functionally unstable ankles may have a predisposition to FAI, as evidenced by the decreased performance on the contra-lateral healthy limb; and (2) FAI affects the postural control system at a level that is high enough to influence stability during stance on either extremity (Tropp et al., 1984).

The finding of bilateral changes in subjects with unilateral ankle instability has been shown in many studies that used external controls to compare subjects with unilateral FAI (Konradsen & Ravn, 1991; Lofvenberg, Karrholm, Sundelin & Ahlgren, 1995; Mitchell et al., 2008a; Tropp & Odenrick, 1988). In both a postural measure (Konradsen & Ravn, 1991; Tropp & Odenrick, 1988) and muscle latencies (Konradsen & Ravn, 1991; Lofvenberg et al., 1995), there was no difference between legs of the subjects with unilateral FAI, but a difference was shown between these subjects and healthy control subjects. Ebig et al. (1997) also found no significant differences between the injured and uninjured ankles in subjects with unilateral FAI for muscle latencies of the peroneal and tibialis anterior muscles. These findings indicate that a central processing problem may exist in people with FAI (Hiller et al., 2007). This supports the theory of motor programme control where receptors from the paired lower limb joints provide afferent information, and damage to one joint and its receptors results in insufficient information reaching higher centres, and therefore high quality movement is jeopardised (Gauffin et al., 1988; Hogervorst & Brand, 1998; Waddington & Adams, 1999). It therefore, appears that there is not a spectrum of disability worsening from uninjured

through unilateral FAI to bilateral FAI, but a difference between uninjured subjects and those with either unilateral FAI or bilateral FAI (Hiller et al., 2007).

2.9 Neuromuscular Fatigue

Some authors have suggested that fatigue plays a significant role in the occurrence of ankle injuries (Gribble & Hertel, 2004a; Huston et al., 2005; Ochsendorf et al., 2000; Pasquet et al., 2000). Fatigue is defined as any exercise-induced reduction in force generating capacity of a muscle (Bigland-Ritchie & Woods, 1984). Anecdotally, many injuries occur during the latter stages of activity when fatigue is present (Hawkins et al., 2001). Whether the onset of fatigue occurs centrally or peripherally, several researchers have documented decreases in the neuromuscular feedback system of the joint around which the fatigued muscles are located (Gribble & Hertel, 2004a; Harkins et al., 2005; Yaggie & McGregor, 2002; Yeung, Au & Chow, 1999).

2.9.1 Isokinetic Dynamometry to Elicit Localised Fatigue

The isokinetic dynamometer is often the choice of method for localised muscular fatigue studies (Bellew & Fenter, 2006; Gribble & Hertel, 2004a; Salavati, Moghadam, Ebrahimi & Arab, 2007; Wikstrom, Powers & Tillman, 2004; Yaggie & McGregor, 2002), as although this fatigue protocol may not be sports specific, it is a way of standardising the speed and movement for each subject. It is generally agreed in the literature that isokinetic testing is preferred over isometric testing, as it provides a more realistic sporting movement (Yaggie & Armstrong, 2004). Following the completion of isokinetic fatigue protocols (usually 50% of the maximal voluntary contraction), numerous authors

have reported deficits such as: impairments in postural sway measures (Gribble & Hertel, 2004b; Salavati et al., 2007; Yaggie & McGregor, 2002), a reduction in peak torque of various muscles (Carcia, Martin & Drouin, 2008) and a decrease in peroneal reflex amplitude (Jackson, Gutierrez & Kaminski, 2009; Wilson & Madigan, 2007). The majority of studies focus on healthy subjects (Bellew & Fenter, 2006; Gribble & Hertel, 2004b; Salavati et al., 2007; Wikstrom et al., 2004; Yaggie & McGregor, 2002), with very limited research on FAI sufferers (Gribble, Hertel, Denegar & Buckley, 2004).

2.9.2 Effect of Localised Fatigue on Muscle Latency

Jackson et al. (2009) hypothesised that isokinetic fatigue would cause an increase (delay) in muscle latencies. However, the results of Jackson et al. (2009) found that isokinetic fatigue lead to a significant decrease (improvement) in muscle latency in the peroneus longus and peroneus brevis muscles. Jackson et al. (2009) found no Group x Test interactions, and therefore put their results down to a possible learning effect, in which all subjects became more comfortable on the tilt perturbation device throughout the testing, which resulted in a facilitation of the reflex and therefore an improvement in muscle latency. In contrast, Cools et al. (2002) studied muscle latencies in the deltoid and trapezius muscles during a sudden downward falling movement of the arm. Their results found that following isokinetic fatigue there was a significant increase (delay) in muscle latencies in all muscles tested (Cools et al., 2002).

Jackson et al. (2009) also investigated the effect of isokinetic fatigue on other EMG parameters and found a decrease in reflex amplitude of the muscles in response to a tilt perturbation following fatigue. This result suggests that fatigue may impair reflex

amplitude, and therefore may impair an individual's ability to correct for an unexpected ankle inversion in a fatigued state. This may support the anecdotal evidence that many injuries occur later in the competition, when the athlete is fatigued (Hawkins et al., 2001).

2.9.3 Effect of Localised Fatigue on Postural Control

Gribble and Hertel (2004a) found significantly increased COP excursion velocity following hip and ankle isokinetic fatigue protocols. They also reported that the hip fatigue protocol produced higher COP excursion velocities compared to the ankle fatigue protocol (Gribble & Hertel, 2004a). Similar to these findings Miller and Bird (1976) found that fatigue to the proximal musculature of the hip and knee produced greater deficits in postural control compared to fatigue of the ankle musculature. The results from these studies show that maintenance of upright stance in a fatigued state may rely more on proximal neuromuscular control than on the previously accepted ankle strategy of distal muscle recruitment in maintaining postural control.

Gribble and Hertel (2004a) explained that the muscles controlling the hip have larger cross sectional areas compared to muscles surrounding the ankle. It is inherent that the larger, more proximal musculature has the ability to create stronger contractions but with potential of less efficiency of corrective contractions during single-leg stance compared to the ankle (Gribble & Hertel, 2004a). During a fatigued state, it is possible that efficiency of compensatory muscle firing about the hip during a single-leg stance is reduced such that maintenance of single-stance is substantially impaired.

Few researchers have investigated the effect that FAI and fatigue have on postural control collectively, especially dynamic postural control tasks. Gribble et al. (2004) used the SEBT as a measure of dynamic postural control and found that FAI subjects displayed smaller reach distance values and knee flexion angles for all reach directions compared with the uninjured side and the healthy group. The effect of fatigue also amplified this trend (Gribble et al., 2004).

2.9.4 Football Specific Protocol to Elicit Globalised Fatigue

Muscular fatigue is usually evident in the course of a football match, especially towards the end of play (Hawkins et al., 2001). Reilly and Thomas (1976) stated that a fatigue effect was noticeable in the second half of the game as reflected by a drop in the work rate. Bangsbo, Norregaard and Thorso (1991) also reported that a 5% greater distance was covered in the first half. The Drust protocol is a football-specific intermittent exercise protocol which is commonly used to provide a fatiguing exercise estimated to be the equivalent in intensity to playing a game of football (Rahnama, Lees & Reilly, 2006). Following the completion of this protocol, several authors have shown deficits such as; a reduction in peak torque of the extensors and flexors of the knee (Nummela, Heath & Paavolainen, 2008; Rahnama, Reilly & Lees, 2002; Rahnama, Reilly, Lees & Graham-Smith, 2003), impairments in soccer kick performance (Kellis, Katis & Vrabas, 2006), an increased varus alignment of the knee (Greig & Siegler, 2009) and a reduction in sprint velocity (Nummela et al., 2008) all which may have implications for increased injury incidence.

2.9.5 Effect of Globalised Fatigue on Muscle Latency

At present there are no studies investigating the effect of a globalised fatigue protocol on muscle latency, however, other EMG parameters have been investigated. Rahnama et al (2006) studied the effect of intermittent football specific exercise on EMG activity at various running speeds. Rahnama et al. (2006) found that there was a decrease in muscle activity as a result of fatigue. The authors concluded that this reduced activity was likely to be associated with decreased strength during prolonged exercise (Rahnama et al., 2006). However, in contrast to these findings, Greig, McNaughton and Lovell (2006) found an increase in integrated EMG activity as a function of exercise duration. The authors stated that the biceps femoris was required to produce greater muscular output to achieve the same standardised workload over time. With these discrepancies present in the literature, other parameters such as muscular latency should be investigated.

2.9.6 Effect of Globalised Fatigue on Postural Control

The effect of many globalised fatigue protocols on postural control have been investigated in the literature. Nelson and Johnson (1973) observed the affects of both local and general fatigue protocols on balance, and found that generalised fatigue impaired balance to a greater extent than local fatigue. In agreement, Yaggie and Armstrong (2004) found that the Wingate exercise test, as a means of generalised fatigue, increased postural sway and transiently degraded postural control in healthy males. Fox, Docherty, Schrader and Applegate (2008) found that both an aerobic yo-yo test, and an anaerobic maximal sprints test negatively affected postural control as

measured by the Balance Error Scoring System (BESS). Nardone, Tarantola, Galante & Schieppati (1998) observed postural sway deviations following a 25 minute treadmill run, and found that sway measures were still elevated after 13 minutes of recovery but had returned to baseline after 23 minutes. In agreement, Bove, Faelli, Tacchino, Lofrano, Cogo and Ruggeri (2007) found that following maximal treadmill exercise there was a significant increase in body sway. However, the aerobic physical exercise prescribed is often not specific to a 'real' sporting situation. The effect of more sports specific protocols on postural control, such as those employed by Drust et al. (2000) are warranted.

2.9.7 Theorised Mechanisms of Neuromuscular Fatigue

Many mechanisms of fatigue have been proposed over the years (Hunter & Enoka, 2003, Kanehisa, Yata, Ikegawa & Fukunaga, 1995; Sahlin, Tonkonogi & Soderlund, 1998; Singh, Nussbaum, Lin & Madigan, 2005; Taylor, Butler & Gandevia, 2000; Westerblad & Allen, 2002). These are commonly separated into peripheral fatigue mechanisms and central fatigue mechanisms. Peripheral fatigue is fatigue occurring within the local motor unit, whereas, central fatigue is fatigue occurring proximal to the motor unit (Kent-Braun, 1999). A commonly discussed factor that affects peripheral fatigue is energy supply and the accumulation of metabolites (Sahlin et al., 1998), particularly lactic acid, which impairs a muscle's ability to produce force (Spagenburg, Ward & Williams, 1998). In addition to this, other metabolites accumulate such as inorganic phosphate, and these are likely to impair the release and reuptake of Ca^{2+} from the sarcoplasmic reticulum (Westerblad & Allen, 2002). Other commonly discussed peripheral factors include muscle fibre type distribution (Tesch, Sjodin, Thorstensson &

Karlsson, 1978; Thorstensson & Karlsson, 1976), muscle strength before fatigue (Hunter & Enoka, 2003, Kanehisa et al., 1995), and the length of the muscle (Fitch & McComas, 1985).

Some of the possible mechanisms of central fatigue include: 1) decreased cortical drive to the motor neuron pool (Taylor et al., 2000), 2) decreased muscle spindle excitability due to the loss of K^+ ions (Singh et al., 2005), 3) an increase in the threshold for muscle spindle discharge which causes a change in co-activation of the alpha and gamma motor neurons as well as a desensitisation of muscle, joint and cutaneous receptors (proprioceptive system) (Kernell, 1969), and 4) desensitisation of the motor neurons, which delays motor unit firing, and may eventually impair neuromuscular control (Kernell, 1969).

In support of central fatigue mechanisms, Freeman et al. (1965) theorized that damage to joint receptors leads to delays in afferent conduction to recruit corrective muscle contractions from efferent signals in response to perturbation, altering joint stability. Deficits in postural control among FAI sufferers have helped to support this theory (Freeman, 1965b; Freeman et al., 1965; Goldie et al., 1994; Tropp et al., 1984; Tropp et al., 1985). Contemporary theory points to the disruption of muscle spindle activity after joint injury as possibly contributing to deficits in neuromuscular control among FAI sufferers (Khin, Ishii, Sakane & Hayashi, 1999). Fatigue increases the threshold of muscle spindle discharge, which disrupts the afferent feedback, subsequently altering joint awareness (Rozzi, Yuktanandana, Pincivero & Lephart, 2000). Deficits in postural control after induced fatigue have helped confirm this theory (Gribble & Hertel, 2004a; Johnston et al., 1998; Lundin et al., 1993; Mattacola, Uhl, McCrory & Malone, 2001).

2.10 Conclusion

Ankle sprains have been shown to be one of the most common injuries in sport. Many 'first time' ankle sprain sufferers go on to experience recurring symptoms, such as episodes of giving way, pain, reduced range of motion, instability and weakness. Postural sway deficits and delayed muscular latencies have been frequently identified in individuals suffering from FAI. These deficiencies could compromise joint stability, and possibly lead to repeated ankle sprains. It has also been suggested that fatigue plays a significant role in the occurrence of ankle injuries. This is supported by anecdotal evidence that many injuries occur during the latter stages of activity when fatigue is present. Fatigue has been shown to have a negative effect on both postural sway and muscular latency. However, additional research is warranted to investigate the combined effects of FAI and fatigue.

Chapter Three

Muscle Latency Reliability Studies

3.1 Introduction to Chapter

This chapter includes Pilot Study One and Pilot Study Two. These pilot studies were undertaken to establish the reliability of the EMG analysis procedure to determine muscle latency that will be used in Study One and Study Three of this thesis.

3.2 Pilot Study One: Reliability of the Determination of Onset of Muscle Contraction Using Electromyography; Sampling Rate, Analysis Method and Smoothing Level.

3.2.1 Abstract

Purpose: To evaluate the relative and absolute reliability for the determination of the onset of muscle contraction, when using different sampling rates (1000, 2500 and 5000 Hz), analysis methods (RMS or average rectified) and smoothing levels (2, 5 or 10 ms), in healthy and FAI subjects following a tilt perturbation. **Aim:** To identify the most reliable combination of parameters to determine EMG onset. **Method:** Ten males suffering from unilateral FAI and 10 male healthy controls were subjected to six inversion and plantarflexion tilt perturbations, three on each leg. Electromyographic signals were recorded for the peroneus longus, tibialis anterior and gluteus medius muscles of both limbs. **Results:** The results highlighted that the most reliable combination that produced ICC's above 0.80 (range: 0.80 to 0.91) and low SEM variance (range: 2.25 to 2.51%) across all conditions was 1000 Hz/RMS/2 ms. **Conclusion:** Previous to the results of the present study, there was little agreement regarding the most appropriate method of recording and analysing the EMG trace to

determine the onset of muscle contraction. Some studies have provided explanations for why they have used a certain sampling rate, or analysis method, however, research was lacking that investigated the reliability when these parameters were combined.

3.2.2 Introduction

Several researchers have used EMG methods to study functionally unstable subjects versus healthy subjects in response to a tilt perturbation (Brunt, Andersen, Huntsman, Reinhart, Thorell & Sterling, 1992; Isakov et al., 1986; Johnson and Johnson, 1993; Konradsen and Ravn, 1990; Mitchell et al., 2008a). During an ankle plantarflexion and inversion perturbation, the ankle dorsiflexor and evertor muscles will be reflexively activated to decelerate the plantarflexion and inversion movements (Ebig et al., 1997; Lynch et al., 1996; Nawoczenski et al., 1985). The time between the perturbation and reflexive muscle activation is known as the latency period (Ebig et al., 1997; Lynch et al., 1996), which is essentially the duration of a muscle's stretch reflex (Kernozek et al., 2008). This onset of muscle contraction is one of the most common EMG parameters evaluated; however, no standard method of recording and analysing the EMG trace is used in the literature. In order for comparisons to be easily made between studies, a standard and reliable method for recording and processing the EMG signal must be determined.

When looking at the sampling rate of the EMG the rates vary in the literature. The sampling rates ranged from 500 Hz (Gruneberg et al., 2003), to 1000 Hz (Akhbari et al., 2007; Benesch et al., 2000; Eechaute et al., 2007; Kernozek et al., 2008), to 1024 Hz (Lynch et al., 1996), and up to 2000 Hz (Delahunt et al., 2006b; Hodges and Bui, 1996).

The majority of studies opt for a sampling rate of 1000 Hz, however, there is usually no justification for why this rate is chosen. The Nyquist Sampling Theorem states that the sampling frequency should be at least twice the highest frequency contained in the signal. For example, if the highest bandwidth frequency is 450 Hz, the sampling frequency must be at least 900 Hz. In terms of reliability no study at present has investigated the sampling rates of the EMG on relative and absolute reliability. In the present study it can be hypothesised that when the EMG is sampled at 1000 Hz, the least variation will be detected due to the decreased sampling rate, and therefore the most reliable results will be found.

DeLuca (1997) addressed some of the issues surrounding the processing of the EMG signal. Two analysis methods are commonly used: the root-mean-squared (RMS) and the average rectified method (AVR). Both are appropriate and provide useful measurements of signal amplitude (DeLuca, 1997; DeLuca & Merletti, 1988). For the EMG signals detected during voluntary movement, for example a tilt perturbation, the RMS value may be more appropriate as it represents the signal power, and thus has a clear physical meaning, whereas, the AVR value is a measure of the area under the signal and therefore does not have a specific physical meaning (DeLuca, 1997). The majority of studies have opted for the RMS method (Brunt et al., 1992; DeLuca, 1997; Ebig et al., 1997; Kernozek et al., 2008; Mitchell et al., 2008a), with very few choosing the AVR method (DeLuca, 1997). However, many studies do not mention the type of signal processing they used, or it appears that they have directly analysed the raw EMG trace (Benesch et al., 2000; Eechaute et al., 2007; Lynch et al., 1996).

The final part of analysing the EMG signal involves the level of smoothing used on the trace. Smoothing is used to remove some of the 'noise' from the trace, such as movement or wire artifact. It has been suggested that over smoothing above 25 to 30 ms is not recommended when the focus of the EMG is time related, as over smoothing above this level may introduce detectable delays. Unfortunately, no studies could be found in the literature that actually reported the level of smoothing that they undertook on their processed signal (DeLuca, 1997; Kernozek et al., 2008; Mitchell et al., 2008a). If under smoothing of the signal occurs this may induce a type I error, and lead to a detection of the onset of muscle contraction before it has actually occurred, whereas, over smoothing of the signal may lead to delays in the detection of muscle onset determination, known as a type II error. It can be seen from the literature that currently there is little agreement regarding the most appropriate method of recording and analysing the EMG trace to determine the onset of muscle contraction.

The aim of this study was therefore to evaluate the relative and absolute reliability for the determination of the onset of muscle contraction, when using different sampling rates (1000, 2500 and 5000 Hz), analysis methods (RMS or AVR) and smoothing levels (2, 5 or 10 ms), in healthy and FAI subjects following a tilt perturbation. The study aimed to identify the most reliable combination of parameters to determine EMG onset.

3.2.3 Method

3.2.3.1 Subjects

Twenty male subjects were recruited for this study; ten subjects suffered from functional ankle instability (age = 20.20 ± 4.35 years, height = 179.90 ± 5.55 cm, and mass = 79.60 ± 11.28 kg) and ten subjects acted as healthy controls (age = 21.04 ± 3.36 years, height = 181.11 ± 6.75 cm, and mass = 78.78 ± 11.05 kg). Institutional ethical approval was granted for this study. All subjects read the subject briefing document (Appendix One) and provided written informed consent (Appendix Two) before participation.

Friel et al. (2006) defined FAI as the subjective feeling of ankle instability or recurrent, symptomatic ankle sprains (or both) due to proprioceptive and neuromuscular deficits. The subjects with a history of FAI in the present study had not suffered an ankle sprain for a minimum of 3 months, so were currently deemed healthy but with a history of FAI. Inclusion criteria consisted of males, aged 18-25 years, who participated in semi-professional football (two training sessions and one match per week) and who were right leg dominant. The dominant leg was defined as the preferred kicking leg and in the unilateral FAI group the right ankle was the unstable ankle.

Subjects were excluded from the study if they were under the influence of alcohol or any other psycho-active substance, if they had a cold, flu, inner ear or sinus infection in the last two weeks, if they suffered from any musculo-skeletal injuries, knee or hip injuries, fractures to the lower limbs, visual impairments, vestibular deficits, or signs of injury such as pain and/or swelling in their ankles. Subjects were also excluded if they had

ever been told by a doctor that they should not exercise, if they did not participate in regular (≥2 x week) aerobic exercise, and if they did not feel fully fit and eager to act as a subject (Appendix Three).

All suspected FAI subjects were required to fill out the FAI questionnaire (Appendix Four). Developed by Hubbard and Kaminiski (2002) this validated questionnaire required subjects to answer “yes” to questions 3, 5, 6, 7 and 9, and “no” to questions 4, 8 and 10 to be included in the study as an FAI subject. Following satisfactory completion of the questionnaire, both of the subject’s ankles were examined to rule out mechanical instability via the anterior drawer and talar tilt tests. The validity of these tests have been established (Bahr, Pena, Shine, Lew, Lindquist, Tyrdal et al., 1997; Docherty & Rybak-Webb, 2009), however, Bahr et al. (1997) suggested that the sensitivity of these tests were improved when the anterior drawer test was performed with the ankle in plantar-flexion, and the talar tilt test was performed with the ankle in dorsi-flexion. The subject’s uninjured ankle acted as the control.

3.2.3.2 Experimental Design

Subject’s age, mass and height were recorded. Three muscle sites on each lower extremity were prepared for EMG set up by shaving and cleaning the area with an alcohol wipe (Seton Healthcare Group plc, UK). Electromyographic activity was recorded using the DataLINK data acquisition system (Biometrics Ltd, UK) with pre-amplified stainless steel surface electrodes (Biometrics Ltd, SX230-1000, gain x1000, bandwidth 20-450 Hz, noise < 5 μV, input impedance > 10^{15} Ω), which reduced the influence of wire movement artifact. The EMG signal and digitals were sampled at three

different rates a) 1000 Hz, b) 2500 Hz, and c) 5000 Hz. The electrodes were applied to the peroneus longus, tibialis anterior and gluteus medius muscles, with an inter-electrode distance of 2 cm. The SENIAM guidelines were followed for electrode placement (Appendix Five). Surface electrodes were applied to both legs of the subject, and the subject unit box where the electrodes were plugged into, was attached around the waist of the subject. The electrode wires were secured down using Velcro bands; one around the upper thigh and one around the lower thigh.

Correct placement of the electrodes was verified by manual resistance testing, which activated the specific muscle group required. Manual muscle testing was performed on a clinical assessment couch. For testing of the peroneus longus and tibialis anterior muscles the subject was asked to lie in the supine position with their feet hanging off the end of the couch. For testing of the peroneus longus the subject was required to plantar flex and evert their foot, whilst the investigator applied the opposite force of dorsiflexion and inversion. For the tibialis anterior the subject was required to dorsi flex and invert the foot, whilst the investigator applied the opposite force of plantarflexion and eversion. For the gluteus medius muscle the subject was required to lie on their side with the test leg facing upwards. The subject was then asked to abduct the thigh at the hip joint, whilst the investigator applied the opposite force of adduction. Conformation of correct electrode placement was seen by observing the oscillations of the EMG trace on the DataLINK software (Version 5.02 Biometrics Ltd, UK) on the computer screen (Figure 3.1).

The subject then completed a five minute cycle on a Monark cycle ergometer (Monark, Varberg, Sweden) at 50 rpm with a resistance of 50 Watts. The tilt platform was

purpose built, and was used in the study by Mitchell et al. (2008a). The tilt platform was designed to simulate the lateral ankle sprain mechanism of talocrural joint plantarflexion and subtalar joint inversion. The tilt movement consisted of two components from a neutral standing position: 30° of inversion in the frontal plane and 20° of plantarflexion in the sagittal plane. The platform was constructed with two moveable plates so that either foot could be tilted independently, thus removing any anticipatory effect (Figure 3.2). A digital sensor was attached to the hinge of the tilt platform, so the tilt onset could be recorded in a separate channel of the DataLINK software.

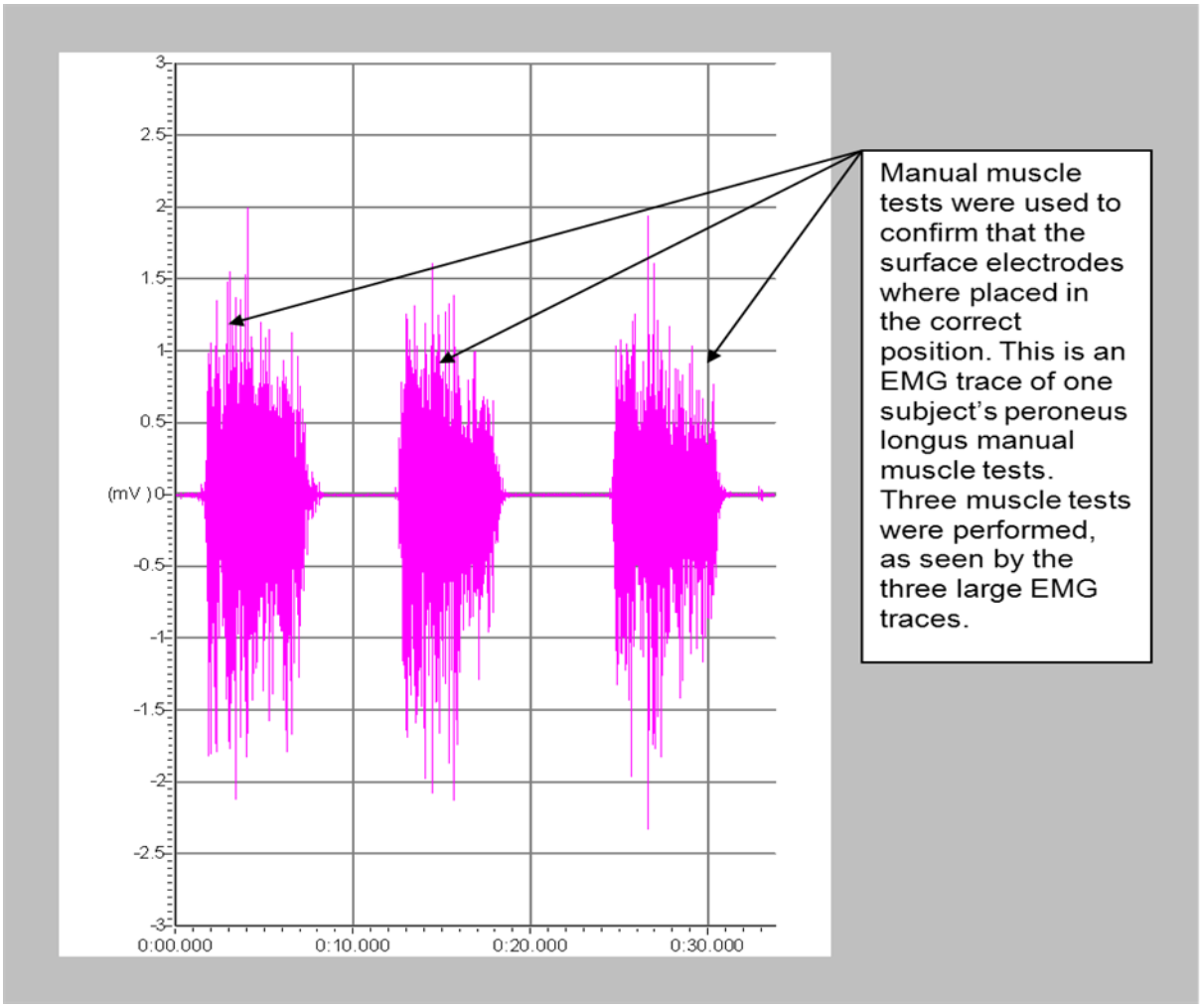


Figure 3.1. Confirmation of Correct Electrode Placement

The subjects were asked to stand on the tilt platform, with their eyes open and were told that at some point in the next 30 seconds one side of the tilt platform would tilt. Subjects had a 2 minute rest period between tilt trials. This procedure was then repeated randomly on each leg (i.e. the unstable ankle (UA) and stable ankle (SA) of the FAI group and the dominant ankle (DA) and non-dominant ankle (NDA) of the control group) a total of three times and averages of these were used for analysis. After performing the procedure the surface electrodes were removed from the subject's lower limbs. The subject then performed a five minute cool down on the cycle ergometer at 50 rpm with a resistance of 50 Watts. To assess test-retest reliability the subjects were required to return to the laboratory 7 days later to repeat the above procedure.

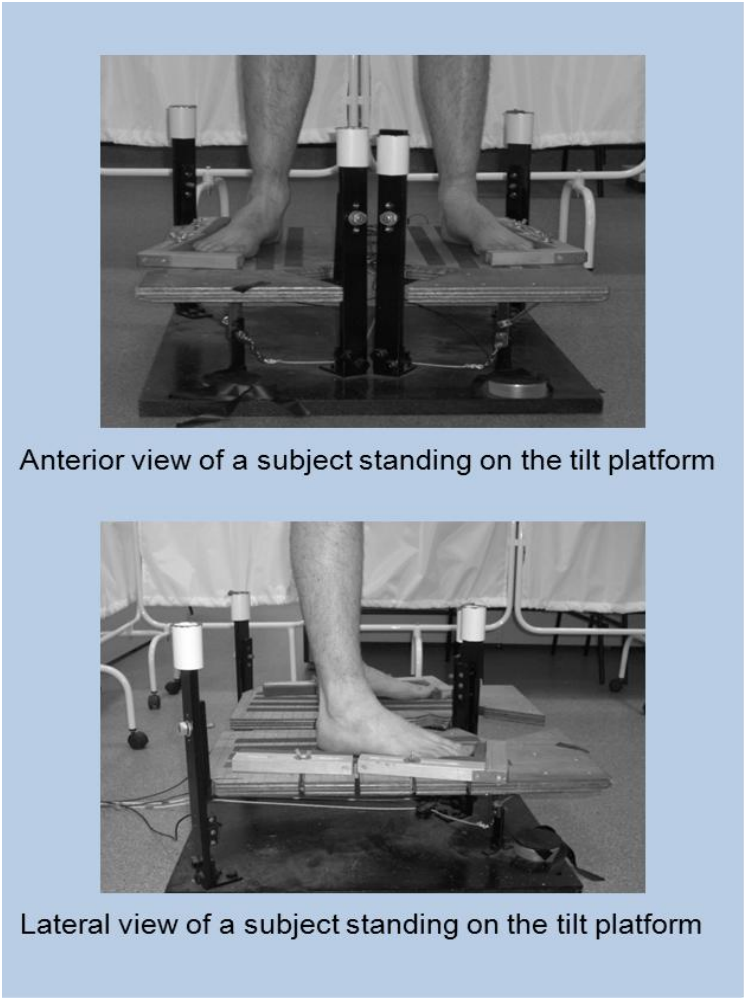


Figure 3.2. Tilt Platform Used to Induce the Plantarflexion and Inversion Perturbation

3.2.3.3 Data Analysis

The Biometrics DataLINK software was used to record the EMG data. The data was analysed using DataLOG (Version 7.00 Biometrics Ltd, UK). Each EMG trace was processed using i) RMS method and ii) AVR method. Following this the traces were smoothed by a) 2 ms, b) 5 ms and c) 10 ms (Figure 3.3), using the sliding window technique. This technique works by choosing a selected time frame (e.g. 10 ms); the sliding window then averages the data over 10 ms intervals (i.e. between 1 and 10 ms, then 2 and 11 ms, then 3 and 12 ms, and so on), until the entire trace has been analysed. After these conversions, the data was exported to Excel (2003), where macros were created to assist the analysis of the data.

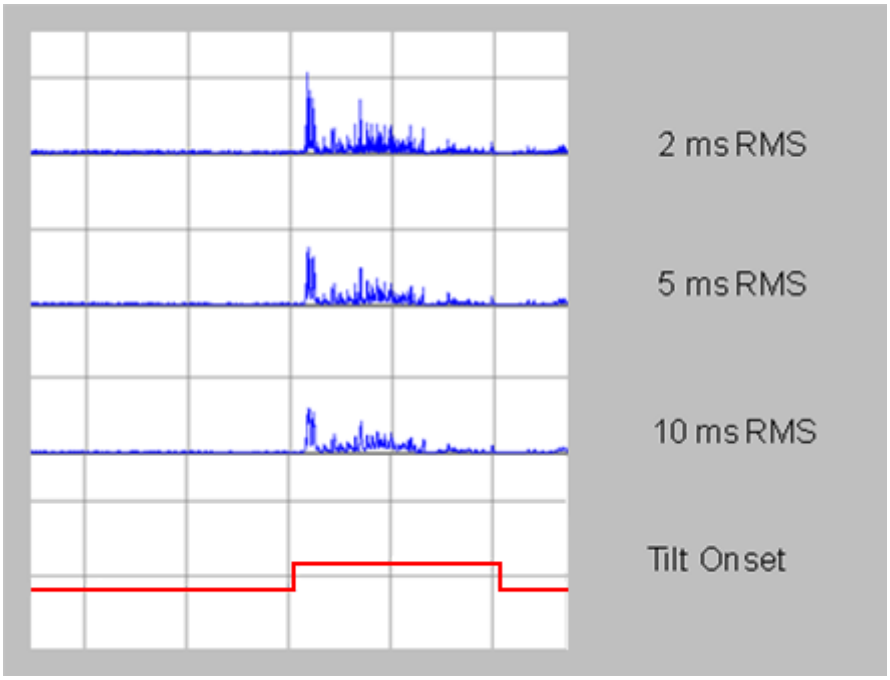


Figure 3.3. Visual Changes to EMG Trace with Different Levels of RMS Smoothing

Firstly, the mean baseline value for each muscle was calculated by highlighting a 50 ms window where the EMG trace was flat – indicating that the subject was standing still at

this point. After the mean was calculated the standard deviation (SD) was calculated using the same 50 ms window. Following this, the standard deviation was tripled to get the 3 SD value. Finally, this 3 SD value was added to the mean. The Excel macro was then instructed to find the point at which the EMG trace went above the mean plus 3 SD. If this burst of muscle activity lasted for 50 ms or greater the muscle was determined as 'on'. From this point the muscle latency was found by reading off the adjacent time (ms) value from the start of the muscle burst. The muscle latency was then calculated by subtracting the muscle onset time from the tilt onset time (Appendix Six).

The tilt onset time was found by using a digital sensor, which was attached to the hinge of the tilt platform. When the platform was tilted, a digital signal was sent to the Biometrics DataLINK software and recorded. When the EMG data was exported to Excel the digital signal was also exported. The point of tilt onset was identified when there was a voltage state change in the column which represented the digital. The corresponding time (ms) value was then read off when this occurred, and was used to calculate muscle latency (Figure 3.4) (Appendix Six).

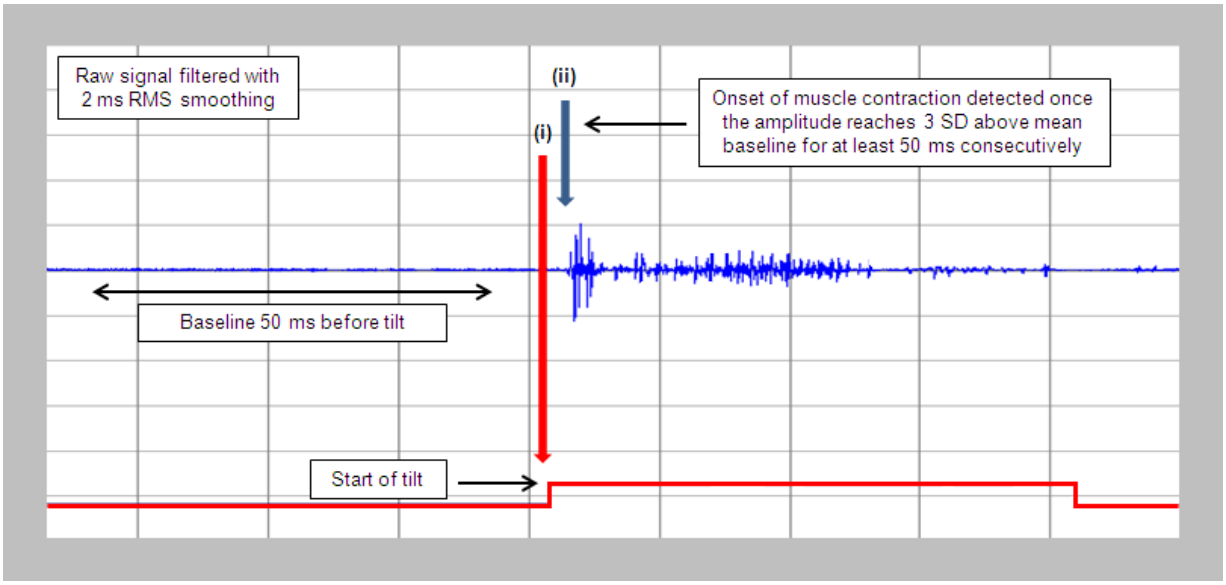


Figure 3.4. Onset Detection Using Tested Parameters. Muscle latency is calculated by subtracting the time at (ii) (onset of muscle contraction) by the time at (i) (onset of tilt mechanism).

3.2.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. Systematic bias, which refers to a difference in measurements in a particular direction between repeated tests, was assessed with 18 (muscle latencies for the 18 different combinations) 4 x 2 (ankle [UA, SA, DA or NDA] x time [first week testing or second week testing]) mixed factorial analysis of variance (ANOVA) for each of the three muscles tested (peroneus longus, tibialis anterior and gluteus medius) in the tilt and support limb. The within-subject factor was time of test, and the between-subject factor was ankle tested. Sphericity was verified for all data being compared by the Mauchly test. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of

homogeneity of intercorrelations. The Multivariate Test box (Wilk's Lambada value) was studied for two-way interactions and then main effects, to identify differences for the within-subject factor (time) ($P<0.05$). The Test of Between-Subject Effects box was observed to identify differences for the between-subject factor (ankle) ($P<0.05$). Tukey's post-hoc test was used to determine exactly where the significant findings occurred for the between-subject factor. Due to multiple comparisons being made between groups, a Bonferroni adjustment was carried out, and the alpha level was set at $P<0.0125$.

Relative reliability was assessed by calculating the Intraclass Correlation Coefficent ($ICC_{(2,1)}$). Weir (2005) stated that the 2-way model addressed both systematic and random error, and therefore was most appropriate. An ICC above 0.90 was considered very high, between 0.70 and 0.89 as high, between 0.50 and 0.69 as moderate, and below 0.49 as low (Munro, 1997). Currier (1990) suggested that an ICC value > 0.80 was acceptable for clinical work. From the ICC value the Standard Error of Measurement (SEM) was calculated, which represented absolute reliability. It has been argued that the 2-way model ICC should be used when calculating the SEM as the systematic and random errors are considered separately (Hopkins, 2000; Weir, 2005). The SEM was calculated using the following formula:

$$SEM = SD\sqrt{1 - ICC}$$

Where SEM = standard error of measurement, SD = the standard deviation of the sample, and ICC = the calculated intraclass correlation coefficient. The SEM was then converted to a percentage of the muscle latency value. This was done in order to allow

clinical usage of this measure (Bosquet, Maquet, Forthomme, Nowak, Lehance & Crosier, 2010).

The muscle latency data from this pilot study was also used to calculate power, and therefore predict a suitable sample size to be used in Study One and Study Three of this thesis (Appendix Seven).

3.2.4 Results

The relative reliability results show that the 18 combinations of analysis presented with a range of reliability values, from poor through to excellent (ICC range: 0.44 to 0.91) (Appendix Eight). However, the main trends that have become apparent from the relative reliability results are that when the sampling rate was set to 1000 Hz this produced the highest relative reliability values (ICC range: 0.63 to 0.91). When the sampling rate was increased to 2500 and 5000 Hz the relative reliability decreased slightly (2500 Hz ICC range: 0.59 to 0.84, 5000 Hz ICC range: 0.44 to 0.76). When the RMS method of analysis was used this produced slightly higher reliability values than the AVR method (RMS ICC range: 0.53 to 0.91, AVR ICC range: 0.44 to 0.81). The 2 ms smoothing level produced the highest relative reliability values (ICC range: 0.51 to 0.91). When the level of smoothing was increased to 5 ms and 10 ms the relative reliability decreased (5 ms ICC range: 0.48 to 0.84, 10 ms ICC range: 0.44 to 0.80). As previously mentioned, Currier (1990) suggested that an ICC above 0.80 was acceptable for clinical work. There was only one combination that produced ICC's above this value throughout each condition; 1000 Hz/RMS/2 ms (sampling rate/analysis method/smoothing level) (Table 3.1). Therefore, from the relative reliability results this

combination would be the most appropriate for determining the onset of muscle contraction using EMG.

The absolute reliability results show similar trends to the relative reliability results (Appendix Eight). When the sampling rate was set to 1000 Hz this produced the lowest SEM values (range: 2.25 to 2.72%). When the sampling rate was increased to 2500 and 5000 Hz the SEM variance increased (2500 Hz range: 2.30 to 2.76%, 5000 Hz range: 2.36 to 2.84%). When the RMS method of analysis was used this produced slightly lower SEM variance, than the AVR method (RMS range: 2.25 to 2.78%, AVR range: 2.44 to 2.84%). The 2 ms smoothing level produced the lowest SEM variance (range: 2.25 to 2.76%). When the level of smoothing was increased up to 5 ms and 10 ms the SEM variance increased (5 ms range: 2.32 to 2.84%, 10 ms range: 2.40 to 2.75%). It has been stated that SEM variances below 10% are deemed acceptable. All the SEM variances were below this value so all could be considered appropriate. However, when taking into consideration both absolute and relative reliability, there was only one combination that showed SEM values below 10% and ICC results above 0.80, this was 1000 Hz/RMS/2 ms (Table 3.1). Therefore, from the absolute and relative reliability results this combination would be the most appropriate for determining the onset of muscle contraction using EMG.

In addition to the reliability results, the mixed factorial ANOVA found no significant difference in muscle latency between the first week of testing and the second week of testing in all subjects and muscles tested.

Table 3.1. Test-Retest Results for the 1000 Hz/RMS/2 ms EMG Analysis Combination

CONDITION	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
TILT LIMB												
DA	48.45(3.2)	48.40(4.5)	0.91	2.25	49.55(3.8)	49.47(3.6)	0.90	2.28	52.11(4.8)	52.12(4.6)	0.85	2.35
NDA	49.65(3.3)	49.60(4.4)	0.88	2.35	49.56(3.6)	49.46(3.5)	0.81	2.30	53.34(4.9)	53.38(4.9)	0.80	2.40
UA	55.55(3.3)	55.54(4.5)	0.82	2.45	54.93(3.8)	54.98(3.6)	0.80	2.49	59.34(4.8)	59.39(4.6)	0.81	2.51
SA	54.64(3.7)	54.65(4.5)	0.81	2.39	54.80(3.2)	54.87(3.4)	0.80	2.30	58.38(4.3)	58.41(4.1)	0.81	2.30
SUPPORT LIMB												
DA	64.16(3.1)	64.18(3.3)	0.81	2.35	65.34(3.3)	65.33(2.4)	0.80	2.30	66.48(4.3)	66.47(3.7)	0.85	2.45
NDA	65.53(3.3)	65.57(3.3)	0.80	2.39	66.13(3.7)	66.17(2.8)	0.81	2.36	67.78(4.2)	67.79(3.3)	0.83	2.47
UA	66.56(3.9)	66.57(3.7)	0.82	2.51	67.10(3.7)	67.09(2.8)	0.84	2.47	68.78(4.9)	68.75(3.9)	0.89	2.36
SA	66.11(3.0)	66.16(3.4)	0.85	2.45	66.84(3.3)	66.83(2.4)	0.80	2.30	68.28(4.4)	68.24(3.0)	0.82	2.45

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement, DA: Dominant Ankle, NDA: Non-Dominant Ankle, UA: Unstable Ankle, SA: Stable Ankle

3.2.5 Discussion

The aim of this study was to evaluate the relative and absolute reliability for the determination of the onset of muscle contraction, when using different sampling rates (1000, 2500 and 5000 Hz), analysis methods (RMS or AVR) and smoothing levels (2, 5 or 10 ms), in healthy and FAI subjects following a tilt perturbation. The study aimed to identify the most reliable combination of parameters to determine EMG onset. The results highlighted that the most reliable combination that produced ICC's above 0.80 and low SEM variance across all conditions was 1000 Hz/RMS/2 ms.

The sampling rates in the present study were set at 1000 Hz, 2500 Hz and 5000 Hz. The results found that the 1000 Hz condition produced the most reliable results (ICC range: 0.63 to 0.91, SEM range: 2.25 to 2.72%). When the sampling rate was increased up to 2500 Hz and 5000 Hz the reliability decreased (2500 Hz ICC range: 0.59 to 0.84, 2500 Hz SEM range: 2.30 to 2.76%, 5000 Hz ICC range: 0.44 to 0.76, 5000 Hz SEM range: 2.36 to 2.84%). It was hypothesised that when the EMG sampled at 1000 Hz, the least variation would be detected due to the decreased sampling rate, and therefore the most reliable results would be found. This hypothesis can therefore be formally accepted. Very rarely do authors state why they have chosen a particular level of sampling. The Nyquist Sampling Theorem states that the sampling frequency should be at least twice the highest frequency contained in the signal, or in mathematical terms:

$$f_s \geq 2f_c$$

where f_s is the sampling frequency, and f_c is the highest frequency contained in the signal. In the present study the highest sampling rate, or bandwidth frequency, was 450 Hz, therefore 1000 Hz is slightly more than twice this frequency, and would be suitable

as a sampling rate. Ives and Wigglesworth (2003) found that oversampling was unnecessary to gather typical amplitude and timing measures from the surface EMG signal. However, sampling below half the Nyquist rate is likely to result in a poor temporal and amplitude representation of the signal (Ives & Wigglesworth, 2003). The present study agrees with the results from Ives and Wigglesworth (2003) as 1000 Hz was found to be the most reliable sampling rate, and oversampling at 2500 Hz and 5000 Hz decreased reliability. Therefore, 1000 Hz would be recommended for use in future studies when using the same equipment and protocol as the present study.

Results of the present study showed that when the RMS method of analysis was used this produced the most reliable results (RMS ICC range: 0.53 to 0.91, RMS SEM range: 2.25 to 2.78%). When the AVR value was used the reliability results decreased (AVR ICC range: 0.44 to 0.81, AVR range: 2.44 to 2.84%). DeLuca (1997) stated that for the EMG signals detected during movement, for example a tilt perturbation, the RMS value may be more appropriate as it represents the signal power, and thus has a clear physical meaning, whereas, the AVR value is a measure of the area under the signal and therefore does not have a specific physical meaning (DeLuca, 1997). The majority of studies have opted for the RMS method (Brunt et al., 1992; DeLuca, 1997; Ebig et al., 1997; Kernozek et al., 2008), but again the authors do not often justify why this method was chosen. Lee, Ho, Rastgaar, Krebs and Hogan (2011) found that the RMS method showed good reliability during isometric voluntary contractions of the biceps brachii muscle. The results from the present study agree with Lee et al. (2011) as the RMS method was shown to be the most reliable method of signal processing and, therefore, future researchers should consider this method.

The results from the present study found that the 2 ms smoothing level produced the most reliable results (ICC range: 0.51 to 0.91, SEM range: 2.25 to 2.76%). When the smoothing level was increased up to 5 ms and 10 ms the reliability decreased (5 ms ICC range: 0.48 to 0.84, 5 ms SEM range: 2.32 to 2.84%, 10 ms ICC range: 0.44 to 0.80, 10 ms SEM range: 2.40 to 2.75%). It has been suggested that over smoothing above 25 to 30 ms is not recommended when the focus of the EMG is time related, as over smoothing above this level may introduce detectable delays. Unfortunately, no studies could be found in the literature that actually reported the level of smoothing that they undertook on their processed signal (DeLuca, 1997; Kernozek et al., 2008). The level of smoothing can influence the probability of making a type I or type II error. Type I errors occur when the muscle is identified as active when it is not, which would arise if the smoothing level was low, such as the 2 ms smoothing level. In contrast, type II errors indicate a failure to identify the EMG onset when it occurs. The frequency of a type II error is increased when the smoothing level is increased, such as the 10 ms condition. Even with these errors in mind the most reliable smoothing level was 2 ms. This value may appear to be very low, however, the Biometrics EMG data acquisition system that was used in the present study, uses pre-amplified electrodes, which removes the wire movement artifact from the EMG signal. This would mean that lower levels of smoothing could be deemed appropriate. The present study found that a smoothing level of 2 ms was the most reliable, and is therefore recommended for future studies when using the same EMG equipment and procedure as the present study.

Although high reliability coefficients (such as ICC's) have been previously reported for EMG studies, SEM values have received little attention in the literature. The SEM is a critical factor which shows the accuracy to which a protocol is reproducible. The SEM

value in the present study was expressed as a percentage in order to allow clinical usage of these measures. As demonstrated by the results of the current study, re-test values for the FAI subjects ranged from 2.30 to 2.84% to the initial test, and for healthy subjects ranged from 2.25 to 2.81% to the initial test. It should therefore, seem appropriate in future studies to attribute differences in muscle latency results to intervention or injury, should they exceed the SEM values outlined in Appendix Eight.

3.2.5.1 Clinical Implications

The relevance of the reliability findings of the present study lies predominantly in the research domain. It may be argued that the increase in reliability is marginal between some of the analysis combinations. For example, when observing the 1000 Hz/RMS/5 ms combination the ICC's range from 0.70 to 0.84, and the SEM's range from 2.32 to 2.69%, when the 1000 Hz/RMS/2 ms combination is observed the ICC's range from 0.80 to 0.91, and the SEM's range from 2.25 to 2.51%. These changes may seem trivial, but in the field of research where reliable protocols are a necessity, the 1000 Hz/RMS/2 ms combination improved the reliability of the protocol, and should be considered over the other combinations in future research.

3.2.5.2 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. The results are also only applicable if the same equipment and protocol as the present study are used.

Future studies may wish to repeat this study but using different EMG data acquisition systems, and varied protocols.

3.2.6 Conclusion

Prior to the results of the present study, there was little agreement regarding the most appropriate method of recording and analysing the EMG trace to determine the onset of muscle contraction. Some studies have provided explanations for why they have used a certain sampling rate, or analysis method, however, research was lacking that investigated the reliability when these parameters were combined. The present study found that the most reliable combination that provided ICC's above 0.80 and low levels of SEM variance across all conditions was 1000 Hz/RMS/2 ms. This combination should be considered in future research if the same equipment and protocol is used as in the present study.

3.3 Development of Research

No standard method of determining the onset of EMG was found to be used in the literature. Pilot Study One addressed the issue of sampling rates, analysis methods and smoothing levels. It was found that the most reliable combination that provided ICC's above 0.80 and low levels of SEM variance across all conditions was 1000 Hz sampling rate, RMS analysis method and 2 ms smoothing level. Therefore, this combination was deemed suitable for use in Study One and Study Three of this thesis. Pilot Study One also provided data to calculate power, and therefore predict a suitable sample size to be used in Study One and Study Three of this thesis.

3.4 Pilot Study Two: Reliability of the Determination of Onset of Muscle Contraction Using Electromyography; Baseline Time, Deviation Level and Number of Samples Exceeding Threshold.

3.4.1 Abstract

Purpose: To evaluate the relative and absolute reliability for the determination of the onset of muscle contraction, when using different baseline times (50, 100 or 500 ms), signal deviation levels (1 SD, 2 SD or 3 SD) and numbers of samples that must exceed the threshold (50 or 100 ms), in healthy and FAI subjects following a tilt perturbation. **Aim:** To identify the most reliable combination of parameters to determine EMG onset. **Method:** Ten males suffering from unilateral FAI and 10 male healthy controls were subjected to six inversion and plantarflexion tilt perturbations, three on each leg. Electromyographic signals were recorded for the peroneus longus, tibialis anterior and gluteus medius muscles of both limbs. **Results:** The results found two combinations that provided ICC's above 0.80 and low SEM variance across all conditions, these were 50 ms/3 SD/50 ms and 50 ms/3 SD/100 ms. A baseline of 50 ms and a deviation level of 3 SD proved to be the most reliable in the present study. However, the number of samples that must exceed the threshold (50 or 100 ms) did not influence reliability. **Conclusion:** Future researchers that are determining muscle onset following a tilt perturbation need to ensure they visually inspect their traces before using a computer algorithm, as erratic bursts of muscle activity in the base line will affect muscle latency and decrease reliability.

3.4.2 Introduction

The temporal characteristics of EMG recordings are parameters of neuromuscular function commonly used in the evaluation of posture and movement (Latash, Aruin & Shapiro, 1995; Lee, Buchanan & Rogers, 1987). The onset of EMG is one of the most common of these parameters evaluated; however, no standard method of determination of this parameter is used in the literature. In limb movement studies, the muscle latency of the postural muscles may be up to several hundred milliseconds (Hodges & Bui, 1996). However, studies investigating muscular latency in response to an unexpected tilt perturbation have been reported much lower, usually in the range of 50-70 ms (Konradsen & Ravn, 1990; Mitchell et al., 2008a). This indicates that in order to allow comparisons between muscles, experimental conditions, and subject groups, accuracy of EMG onset determination is crucial.

A number of studies evaluating the temporal parameters of EMG do not report the methods used for the identification of EMG onset (Belenkii, Gurfinkel & Paltsey, 1967; Oddsson & Thorstensson, 1987). Many studies where the onset determination is described have used visual evaluation of the EMG trace (Latash et al., 1995; Woollacott, Von Hosten & Rosblad, 1988), generally without reporting the criteria on which this visually determined decision is made. Several studies using visual onset have just reported it as the earliest detectable rise in EMG activity above the steady state (Crenna & Frigo, 1987; Inglis, Horak, Shupert & Jones-Rycewicz, 1994; Woollacott et al., 1988). However, this method has often been reported as very subjective and crude, and is often reliant upon the experience of the examiner (DiFabio, 1987; Hodges & Bui, 1996).

In an attempt to increase the objectivity (DiFabio, 1987) of the evaluation of EMG onset and to reduce observer bias (Studenski, Duncan & Chandler, 1991) an increasing number of studies rely on computer analysis methods (Bullock-Saxton, 1994; Karst & Willet, 1995; Steele, 1994; Thompson & McKinley, 1995). However, the methods differ greatly in the literature, and at present there is little agreement on the most appropriate method. Although each of the computer algorithms used differing criteria to determine EMG onset, no studies have evaluated the reliability of each.

When looking at the mean baseline to determine when a muscle is at rest the methods and times used vary significantly. Some authors have used the baseline as the time immediately prior to the stimulus (DiFabio, 1987; Nashner, Shumway-Cook & Marin, 1983; Neafsey, Hull & Buchwald, 1978), others have used the baseline as the time during quiet standing (Chanaud & Macpherson, 1991; Lee et al., 1987; Studenski et al., 1991) and many authors fail to report how they determined their baseline EMG level (Bullock-Saxton, Janda & Bullock, 1993; Greenisen, Vroomen & Vroomen, 1979; Happee, 1992; Thompson & McKinley, 1995). The times selected to determine the mean baseline also vary greatly from 15 ms (Karst & Willet, 1995), to 50 ms (DiFabio, 1987), to 100 ms (Chanaud & Macpherson, 1991; Studenski et al., 1991), to 240 ms (Steele, 1994) and as high as 500 ms (Bullock-Saxton et al., 1994; Lee et al., 1987; Neafsey et al., 1978). Many authors again fail to report their baseline measurement time (Bullock-Saxton et al., 1994; Greenisen et al., 1979; Happee, 1992; Thompson and McKinley, 1995).

When determining EMG onset studies that use computer algorithms have used a certain number of SD’s above the level recorded at baseline. However, the authors

often fail to mention how or why this level of deviation is chosen. Most authors typically report changes of 1-3 SD which influences the probability of making a type I or type II error. Type I errors occur when the muscle is identified as active when it is not, which arises if the threshold is low, such as the 1 SD condition. In contrast, type II errors indicate a failure to identify the EMG onset when it occurs. The frequency of a type II error is increased in the conservative 3 SD condition. In the literature, the levels of deviation have ranged from 1 SD (Karst & Willet, 1995; Steele, 1994), to 1.5 SD (Nashner et al., 1983), to 2 SD (Lee et al., 1987; Neafsey et al., 1978), to 2.5 SD (Chanaud & Macpherson, 1991), to 3 SD (DiFabio, 1987) and even as high as 10 SD (Kernozek et al, 2008). Authors have also used other deviation methods such as 15% of the maximal contraction (Bullock-Saxton et al., 1993) or 5% of the peak magnitude of the burst (Bullock-Saxton et al., 1994).

When observing the number of samples that must exceed a given threshold for the muscle to be determined as ‘on’, very few studies have actually reported this value (Hodges & Bui, 1996). This would suggest that most authors just use the first rise in muscle activity following the tilt perturbation as their onset of muscle activity. Hodges and Bui (1996) found that that the number of milliseconds (ms) for which the mean must exceed the threshold did influence the accuracy of onset determination. Hodges and Bui (1996) found that when the sample was short the chances of identifying an erratic burst of activity as the onset was increased (type I error). In contrast, when the sample was long the chance of ignoring a short EMG burst was increased (type II error) (Hodges & Bui, 1996). Hodges and Bui (1996) found that when comparing their computer based methods to visual determination methods the short sample width (10 ms) consistently identified the EMG onset prior to the visually derived value, the long sample width (50

ms) delayed the identification of onset of EMG, and the medium (25 ms) sample closely approximated the visually derived values. It can be seen from the literature that currently there is little agreement regarding the most appropriate method to determine EMG onset.

The aim of this study was therefore to evaluate the relative and absolute reliability for the determination of the onset of muscle contraction, when using different baseline times (50, 100 or 500 ms), different signal deviation levels (1 SD, 2 SD or 3 SD) and different numbers of samples that must exceed the threshold (50 or 100 ms), in healthy and FAI subjects following a tilt perturbation. The study aimed to identify the most reliable combination of parameters to determine EMG onset. Identification of a computer algorithm that accurately identifies the onset of EMG will provide a method to increase the objectivity of EMG onset determination.

3.4.3 Method

3.4.3.1 Subjects

The same subjects were used as in Pilot Study One (Section 3.2.3.1)

3.4.3.2 Experimental Design

The same experimental design was used as in Pilot Study One (Section 3.2.3.2), apart from the EMG signal and digitals sampled at 1000 Hz.

3.4.3.3 Data Analysis

The same data analysis procedure was used as in Pilot Study One (Section 3.2.3.3); apart from the EMG trace was processed using the RMS method and was smoothed by 2 ms. In addition, each EMG trace was analysed using the 18 different combinations.

3.4.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. Systematic bias, which refers to a difference in measurements in a particular direction between repeated tests, was assessed with 18 (muscle latencies for the 18 different combinations) 4 x 2 (ankle [UA, SA, DA or NDA] x time [first week testing or second week testing]) mixed factorial analysis of variance (ANOVA) for each of the three muscles tested (peroneus longus, tibialis anterior and gluteus medius) in the tilt and support limb. The within-subject factor was time of test, and the between-subject factor was ankle tested. Sphericity was verified for all data being compared by the Mauchly test. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of homogeneity of intercorrelations. The Multivariate Test box (Wilk's Lambada value) was studied for two-way interactions and then main effects, to identify differences for the within-subject factor (time) ($P<0.05$). The Test of Between-Subject Effects box was observed to identify differences for the between-subject factor (ankle) ($P<0.05$). Tukey's post-hoc test was used to determine exactly where the significant findings occurred for

the between-subject factor. Due to multiple comparisons being made between groups, a Bonferroni adjustment was carried out, and the alpha level was set at $P<0.0125$.

Relative reliability was assessed by calculating the $ICC_{(2,1)}$. From the ICC value the SEM was calculated, which represented absolute reliability (Section 3.2.3.4, paragraph 2).

The muscle latency data from this pilot study was also used to calculate power, and therefore predict a suitable sample size to be used in Study One and Study Three of this thesis (Appendix Nine).

3.4.4 Results

The relative reliability results show that all 18 combinations of analysis presented with moderate to excellent reliability (ICC range: 0.61 to 0.95) (Appendix Ten). However, the main trends that have become apparent from the relative reliability results are when the baseline time was set at 50 ms this produced the highest relative reliability values (ICC range: 0.95 to 0.70). When the baseline time was increased to 100 ms and 500 ms the relative reliability decreased (100 ms ICC range: 0.90 to 0.67, 500 ms ICC range: 0.88 to 0.61). When the deviation level was set at 3 SD this produced the highest relative reliability values (ICC range: 0.68 to 0.95). It can be seen that the deviation level of 1 SD and 2 SD produced lower ICC values (1 SD ICC range: 0.63 to 0.85, 2 SD ICC range: 0.61 to 0.83). However, it can be seen that relative reliability was not affected by the number of samples needed to exceed the threshold. When observing all the combinations (Appendix Ten), there was no difference in muscle latency results when

either 50 ms or 100 ms was used, therefore either would be acceptable as the ICC values were the same. As previously mentioned, Currier (1990) suggested that an ICC above 0.80 was acceptable for clinical work. There were only two combinations that produced ICC's above this value throughout each condition: 50 ms/3 SD/50 ms and 50 ms/3 SD/100 ms (baseline time/deviation level/samples exceeding threshold) (Table 3.2). Therefore, it would seem that either of these combinations would be appropriate for determining the onset of muscle contraction using EMG.

The absolute reliability results show similar trends to the relative reliability results (Appendix Ten). When the baseline was set at 50 ms this produced the lowest SEM values (range: 2.20 to 2.86%). When the baseline time was increased to 100 ms and 500 ms the SEM variance increased (100 ms range: 2.30 to 3.03%, 500 ms range: 2.41 to 3.15%). When the deviation level was set at 3 SD this produced the lowest SEM variance (range: 2.20 to 2.98%). When the level of deviation was 1 SD or 2 SD the SEM variance increased (1 SD range: 2.30 to 3.15%, 2 SD range: 2.32 to 3.12%). However, the absolute reliability was not affected by the number of samples needed to exceed the threshold. When observing all the combinations (Appendix Ten), there was no difference in muscle latency results when either 50 ms or 100 ms was used, therefore either would be acceptable as the SEM variance was identical. It has been stated that SEM variances below 10% are deemed acceptable. All the SEM variances were below this value so all could be considered appropriate. However, when taking into consideration both absolute and relative reliability, there were only two combinations that showed SEM values below 10% and ICC results above 0.80, these were 50 ms/3 SD/50 ms and 50 ms/3 SD/100 ms (Table 3.2). Therefore, either of these combinations would be appropriate for determining the onset of muscle contraction using EMG.

In addition to the reliability results, the mixed factorial ANOVA found no significant difference in muscle latency between the first week of testing and the second week of testing in all subjects and muscles tested.

Table 3.2. Test-Retest Results for the 50 ms/3 SD/50 ms and 50 ms/3 SD/100 ms EMG Analysis Combinations

CONDITION	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
TILT LIMB												
DA	49.51(3.5)	49.60(3.3)	0.95	2.20	49.70(3.7)	49.69(3.1)	0.93	2.22	52.78(3.9)	52.81(3.0)	0.90	2.55
NDA	50.55(3.5)	50.61(3.3)	0.88	2.25	49.75(2.3)	49.67(3.1)	0.90	2.21	53.78(3.9)	53.81(3.0)	0.88	2.54
UA	55.50(3.3)	55.54(3.3)	0.90	2.35	53.70(1.9)	53.87(3.1)	0.88	2.42	59.79(3.9)	59.83(3.0)	0.88	2.52
SA	54.91(3.3)	54.95(3.0)	0.88	2.30	53.85(3.2)	53.86(2.2)	0.87	2.40	58.80(3.9)	58.83(3.0)	0.89	2.51
SUPPORT LIMB												
DA	64.68(3.1)	64.70(3.2)	0.85	2.35	65.45(3.7)	65.49(3.0)	0.82	2.40	66.75(2.9)	66.80(2.7)	0.85	2.65
NDA	65.68(3.1)	65.70(3.5)	0.86	2.36	66.45(3.6)	66.46(3.2)	0.88	2.47	67.75(2.6)	67.80(2.7)	0.87	2.65
UA	66.67(3.1)	66.70(3.6)	0.86	2.34	67.45(3.7)	67.47(3.0)	0.85	2.45	68.78(2.9)	68.80(3.8)	0.88	2.68
SA	66.29(3.1)	66.30(3.5)	0.82	2.33	66.93(3.0)	66.95(3.2)	0.83	2.46	68.38(2.9)	68.40(2.8)	0.83	2.63

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement, DA: Dominant Ankle, NDA: Non-Dominant Ankle, UA: Unstable Ankle, SA: Stable Ankle

3.4.5 Discussion

The aim of this study was to evaluate the relative and absolute reliability for the determination of the onset of muscle contraction, when using different baseline times (50, 100 or 500 ms), different signal deviation levels (1 SD, 2 SD or 3 SD) and different numbers of samples that must exceed the threshold (50 or 100 ms). The study aimed to identify the most reliable combination of parameters to determine EMG onset. The results highlighted two combinations that provided ICC's above 0.80 and low levels of SEM variance across all conditions, these were 50 ms/3 SD/50 ms and 50 ms/3 SD/100 ms.

The mean baseline to determine when a muscle was at rest was set at 50, 100 and 500 ms in the present study. The results found that the 50 ms baseline time produced the most reliable results (ICC range: 0.95 to 0.70, SEM range: 2.20 to 2.86%). When the baseline time was increased to 100 ms and 500 ms the reliability decreased (100 ms ICC range: 0.90 to 0.67, 100 ms SEM range: 2.30 to 3.03%, 500 ms ICC range: 0.88 to 0.61, 500 ms SEM range: 2.41-3.15%). The baseline time was taken prior to the tilt perturbation when the subject was in quiet stance. When visually observing the EMG traces, many of the traces did not have a perfectly flat baseline, and often showed short bursts of muscle activity where the subject may have twitched or accidently moved. In the present study when the short 50 ms baseline time was used it seemed that many of these short bursts were excluded, however, as the baseline time increased up to 100 and 500 ms more erratic bursts would be included in the baseline, which contributed towards higher baseline EMG levels, therefore delayed identification of muscle onset (type II error), and in addition to this lower reliability was observed. The few studies in

the literature that have mentioned their baseline measurement time have also selected 50 ms, which would agree with the findings of the present study, however, the authors give no justification as to why this time period was chosen (DiFabio, 1987; Hodges & Bui, 1996). The results from the present study highlight that 50 ms was the most reliable baseline time, however, future studies using computer algorithms to determine muscle onset must visually inspect their traces before using a computer algorithm (DiFabio, 1987; Hodges & Bui, 1996), as erratic bursts of muscle activity in the baseline may affect muscle latency and decrease reliability.

Results of the present study showed that when the deviation level was set at 3 SD this produced the highest reliability values (ICC range: 0.68 to 0.95, SEM range: 2.20 to 2.98%). It can be seen that the deviation level of 1 SD and 2 SD produced lower reliability values (1 SD ICC range: 0.63 to 0.85, 1 SD SEM range: 2.30 to 3.15%, 2 SD ICC range: 0.61 to 0.83, 2 SD SEM range: 2.32 to 3.12%). Little consensus exists for the deviation level for computer based EMG onset determination in the literature. Statistically based EMG onset determination methods typically report changes of 1-3 SD which influences the probability of making a type I or type II error. Type I errors occur when the muscle is identified as active when it is not, which arises if the threshold is low, such as the 1 SD condition. In contrast, type II errors indicate a failure to identify the EMG onset when it occurs. The frequency of a type II error is increased in the conservative 3 SD condition.

The results from the present study showed that EMG onset was slower with the 3 SD condition in comparison to the 1 and 2 SD conditions, however, the 3 SD condition produced the most reliable combinations. Some studies have used much larger

standard deviations, even as high as 10 SD (Kernozek et al., 2008), however, the issue of incurring a type II error was not discussed in their results. Hodges and Bui (1996) did discuss type I and type II errors, but they still found the criteria that produced the most accurate identification of the onset of EMG was 25 ms/3 SD/50 Hz. Hodges and Bui (1996) found that although there was delayed EMG onset determination with the 3 SD condition relative to the other deviation levels (1 SD and 2 SD), individual parameter combinations involving each deviation level were not different from the visually derived values, therefore previously reported methods should not be disputed on the basis of the deviation level.

In the present study if the subject twitched or accidently moved before the tilt perturbation (which was apparent in some EMG traces), when the deviation level was set low (1 SD and 2 SD) the computer algorithm picked up these random bursts as the muscle being ‘on’. The Excel spreadsheet would then have to be visually inspected to ensure that the determination that was used for analysis actually occurred after the tilt perturbation, and was therefore the true onset of muscle activity. When the deviation level was set at a higher level (3 SD) there were less random bursts picked up, which was less problematic when determining muscle onset.

The results from the present study found that the number of samples (ms) for which the mean must exceed the threshold (either 50 or 100 ms) did not influence muscle latencies, and therefore, the reliability results. The EMG traces used in the present study all showed muscles bursts lasting for more than 100 ms following the tilt platform perturbation. This meant that when the number of samples for which the mean must exceed the threshold was set at 50 ms or 100 ms, there was no difference in the onset

result. This result differs to the findings of Hodges and Bui (1996) who found that that the number of milliseconds (ms) for which the mean must exceed the threshold did influence the accuracy of onset determination. Hodges and Bui (1996) found that when the sample was short the chances of identifying an erratic burst of activity as the onset was increased; in contrast, when the sample was long the chance of ignoring a short EMG burst was increased. Hodges and Bui (1996) found that when comparing their computer based methods to visual determination methods the short sample width (10 ms) consistently identified the EMG onset prior to the visually derived value, the long sample width (50 ms) delayed the identification of onset of EMG, and the medium (25 ms) sample closely approximated the visually derived values. The present study chose 50 ms and 100 ms as the number of samples that must exceed the threshold, and no differences were found in muscle latency.

As previously stated in Pilot Study One SEM values have received little attention in the literature. The SEM is a critical factor which shows the accuracy to which a protocol is reproducible. The SEM value in the present study was expressed as a percentage in order to allow clinical usage of these measures. As demonstrated by the results of the current study, re-test values for the FAI subjects ranged from 2.30 to 3.15% to the initial test, and for healthy subjects ranged from 2.21 to 3.12% to the initial test. It would therefore, seem appropriate in future studies to attribute differences in muscle latency results to intervention or injury, should they exceed the SEM values outlined in Appendix Ten.

3.4.5.1 Clinical Implications

As previously mentioned in Pilot Study One (Section 3.2.5.1) the relevance of the reliability findings of this study lies predominantly in the research domain. It may be argued that the increase in reliability is marginal between some of the analysis combinations. For example, when observing the 50 ms/3 SD/50 ms combination the ICC's range from 0.82 to 0.95, and the SEM's range from 2.20 to 2.68%, when the 100 ms/3 SD/50 ms combination is observed the ICC's range from 0.72 to 0.90, and the SEM's range from 2.37 to 2.78%. These changes may seem minor, but in the field of research where reliable protocols are a necessity, the 50 ms/3 SD/50 ms combination improved the reliability of the protocol, and should be considered over the other combinations in future research.

3.4.5.2 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. The results are also only applicable if the same equipment and protocol as the present study are used. Future studies may wish to repeat this study but using different EMG data acquisition systems, and varied protocols.

3.4.6 Conclusion

The results highlighted two combinations that provided ICC's above 0.80 and low levels of SEM variance across all conditions, these were 50 ms/3 SD/50 ms and 50 ms/3

SD/100 ms. A baseline of 50 ms and a deviation level of 3 SD proved to be the most reliable in the present study. However, the number of samples that must exceed the threshold (50 or 100 ms) did not influence reliability. Future researchers that are determining muscle onset following a tilt perturbation need to ensure they visually inspect their traces before using a computer algorithm, as erratic bursts of muscle activity in the baseline may affect muscle latency and decrease reliability.

3.5 Development of Research

No standard method of determining the onset of EMG was found to be used in the literature. Pilot Study Two addressed the issue of baseline times, deviation levels and the number of samples exceeding the threshold. The results of Pilot Study Two highlighted the combination that provided ICC's above 0.80 and low levels of SEM variance across all conditions was 50 ms baseline, 3 SD level, 50 ms exceeding the threshold. Therefore, this combination was deemed suitable for use in Study One and Study Three of this thesis. Pilot Study Two also provided data to calculate power, and therefore predict a suitable sample size to be used in Study One and Study Three of this thesis.

Chapter Four

Muscle Latency in Healthy versus Functionally Unstable Subjects

4.1 Study One: Muscle Latencies in Healthy and Functionally Unstable Subjects During an Unexpected Plantarflexion and Inversion Tilt Perturbation

4.1.1 Abstract

Aim: To investigate muscle latency times of the peroneus longus, tibialis anterior and gluteus medius muscles in the unilateral FAI subject's UA and SA, compared to a healthy control group's DA and NDA, when acting as (i) a tilt limb, and (ii) a support limb. **Method:** Twenty males suffering from unilateral FAI and 20 male healthy controls were subjected to six inversion and plantarflexion tilt perturbations, three on each leg. Electromyographic signals were recorded for the peroneus longus, tibialis anterior and gluteus medius muscles of both limbs. **Results:** The results indicated that there was a significant (P<0.0125) delay in muscle latencies of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to the DA and NDA of the control group, when analysing the tilt limb, however there were no significant differences when analysing the support limb. **Conclusion:** Muscle latency was delayed in both the UA and SA of the FAI subjects, which would suggest a central mechanism of control, or possibly a genetic predisposition to FAI in some individuals.

4.1.2 Introduction

Lateral ankle sprains have been reported as the most common injury in sport (Barrett & Bilisko, 1995; Orteza et al., 1992; Robbins et al., 1995), but also occur among other physically active individuals (Ross & Guskiewicz, 2004). Following the initial ankle

sprain, some individuals will develop a pathological condition known as FAI, exhibiting residual symptoms such as feelings of instability, giving way, pain or re-injury.

In relation to FAI, a common area of investigation is muscle latency (Ebig et al., 1997; Isakov et al., 1986; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990; Mitchell et al., 2008a; Vaes et al., 1999). Many studies have investigated muscle latencies of the peroneus longus (Isakov et al., 1986; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990) and the tibialis anterior muscles in FAI subjects (Ebig et al., 1997; Lofvenberg et al., 1995; Mitchell et al., 2008a). These two muscles are commonly investigated as the peroneus longus is responsible for eversion of the ankle, therefore, resisting inversion during an ankle sprain mechanism. The tibialis anterior is responsible for dorsiflexion of the ankle, and therefore, resists plantarflexion during an ankle sprain. However, there is very limited research on the muscle latency of the gluteus medius muscle (Beckman & Buchanan, 1995). It has been stated that during ambulation the gluteus medius muscle provides stability to the hip in the frontal plane (Friel et al., 2006). Weakness in a stabilising muscle, such as the gluteus medius, may produce deviations in joint motion, a subsequent loss of stability and may contribute towards a repeated injury at the ankle (Friel et al., 2006; Riemann, 2002).

During a joint perturbation, reflexive muscle activity occurs in response to stimulation of mechanoreceptors within ligaments and muscles (Hogervorst & Brand, 1998; Sainburg et al., 1993), presumably to reduce the magnitude of joint movement (Lynch et al., 1996). The time between a perturbation and reflexive muscle activation is known as the latency period (Ebig et al., 1997; Lynch et al., 1996; Nawoczenski et al., 1985), which is essentially the duration of a muscles stretch reflex. In addition to the latency period is

the electromechanical delay (EMD), the delay between muscle activation and the production of tension at the muscles skeletal attachments. This lag occurs because time is required for the action potentials propagation along the sarcolemma, the excitation-contraction coupling process, and the removal of slack in the elastic elements (Alexander & Bennett-Clark, 1997; Cavanagh & Komi, 1979). If the combined muscle latency and the EMD are shorter than the time it takes for the ankle joint to reach its physiological motion limits, the muscles may help to decelerate ankle joint movement and reduce ligamentous strain.

Many laboratory studies have looked at muscle latency in response to an unexpected tilt platform perturbation (Beckman & Buchanan, 1995; Ebig et al., 1997; Isakov et al., 1986; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990; Mitchell et al., 2008a; Vaes et al., 1999). Muscle latencies in the literature are determined by measuring the interval between the platform release and the onset (or marked increase) of ankle muscle EMG activity (Mitchell et al. 2008a). In most tilt platform perturbation studies, the authors commonly examine the muscle latencies of the tilting limb muscles (Ebig et al., 1997; Isakov et al., 1986; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990; Mitchell et al., 2008a; Vaes et al., 1999), with only very few authors having studied the muscle latencies of the supporting limb muscles (Beckman & Buchanan, 1995; Mitchell et al., 2008a). The role of the support limb during a perturbation is not fully understood, and it is possible that the support limb may have other, as yet unknown responsibilities in FAI sufferers.

The aim of this study was therefore to research muscle latency time of the peroneus longus, tibialis anterior and gluteus medius muscles in the unilateral FAI subject's UA

and SA, compared to a healthy control group's DA and NDA, when acting as (i) a tilt limb, and (ii) a support limb. It was hypothesised that the FAI subjects will have significantly increased (delayed) muscle latencies in comparison to the healthy controls when acting as the tilt limb. It was also hypothesised that the FAI subjects will have significantly increased (delayed) muscle latencies in comparison to the healthy controls when acting as the support limb.

4.1.3 Method

4.1.3.1 Subjects

Forty male subjects were recruited for this study; twenty subjects suffered from functional ankle instability (age = 21.05 $\pm$ 4.95 years, height = 178.88 $\pm$ 5.89 cm, and mass = 78.64 $\pm$ 10.48 kg) and twenty subjects served as healthy controls (age = 20.4 $\pm$ 3.36 years, height = 180.11 $\pm$ 6.71 cm, and mass = 79.18 $\pm$ 12.25 kg). Institutional ethical approval was granted for this study. All subjects read the subject briefing document (Appendix One) and provided written informed consent (Appendix Two) before participation.

Refer to Pilot Study One (Section 3.2.2.1, paragraphs 2, 3 and 4) for inclusion and exclusion criteria.

4.1.3.2 Experimental Design

The same experimental design as Pilot Study One was used (Section 3.2.3.2); apart from the EMG signal and digitals sampled at 1000 Hz, and subjects were not required to return to the laboratory seven days later to repeat the procedure.

4.1.3.3 Data Analysis

The same data analysis procedure was used as in Pilot Study One (Section 3.2.3.3); apart from the EMG trace was processed using the RMS method and was smoothed by 2 ms.

4.1.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. A one-way between-groups ANOVA compared the mean muscle latencies of the UA, SA, DA and NDA, when the tilt and support limb was studied. Muscle latency was the dependent variable, whereas ankle (UA, SA, DA and NDA) was the independent variable. The Levene's Test of Homogeneity of Variances box was observed to verify the assumption of homogeneity of variances across groups
The ANOVA table was studied for significant differences; the alpha level was set at $P<0.05$. Tukey's post-hoc test was used to determine exactly where the significant findings occurred. Due to multiple comparisons being made between groups, a Bonferroni adjustment was carried out, and the alpha level was set at $P<0.0125$.

4.1.4 Results

Results from the between-groups ANOVA for the tilting limb showed a significant (P<0.0125) increase (delay) in the muscle latencies of the peroneus longus, tibialis anterior and gluteus medius muscles when comparing the UA of the FAI group to both the DA and NDA of the control group (Figure 4.1). The results also showed a significant (P<0.0125) increase (delay) in muscle latencies of the peroneus longus, tibialis anterior and gluteus medius muscles when comparing the SA of the FAI group to both the DA and NDA of the control group (Figure 4.1). No significant differences were found between the UA and SA of the FAI group. In addition to this, no significant differences were found between the DA and NDA of the control group.

Results from the between-groups ANOVA for the support limb showed no significant differences in the muscle latencies of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to both the DA and NDA of the control group (Figure 4.2). No significant differences were found between the UA and SA of the FAI group. In addition to this, no significant differences were found between the DA and NDA of the control group (Appendix Eleven).

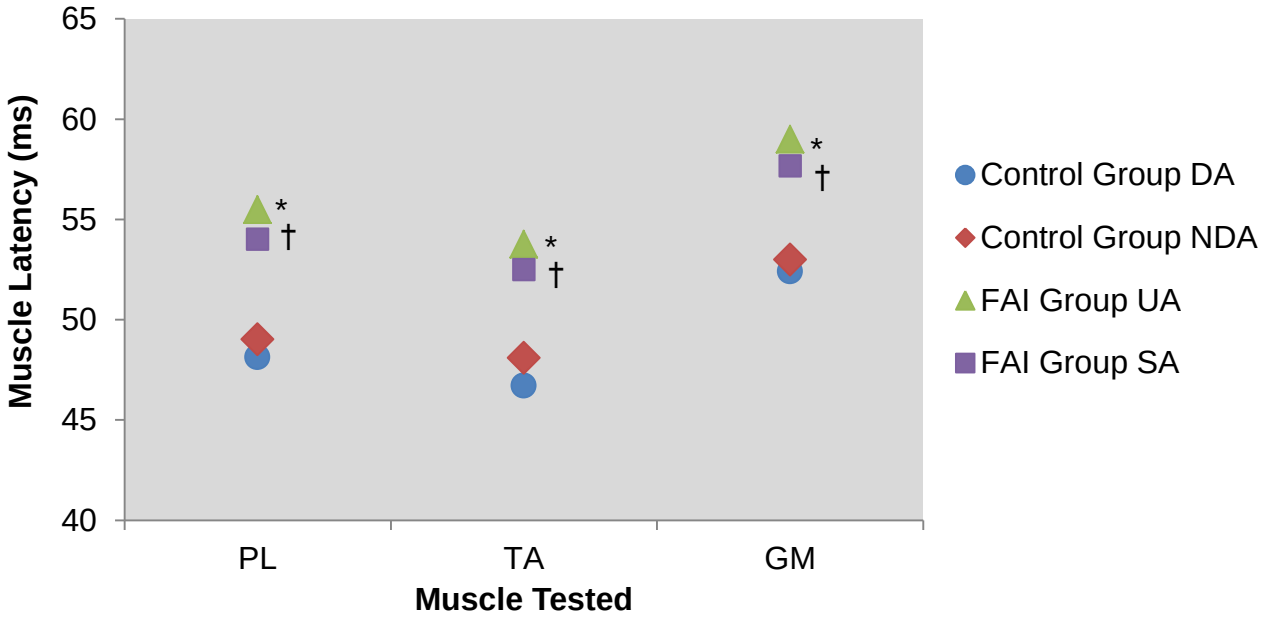


Figure 4.1. Muscle Latencies for the Control and FAI Group, when the Tilt Limb was Studied. * UA significantly (*P*<0.0125) slower than DA and NDA. † SA significantly (*P*<0.0125) slower than DA and NDA.

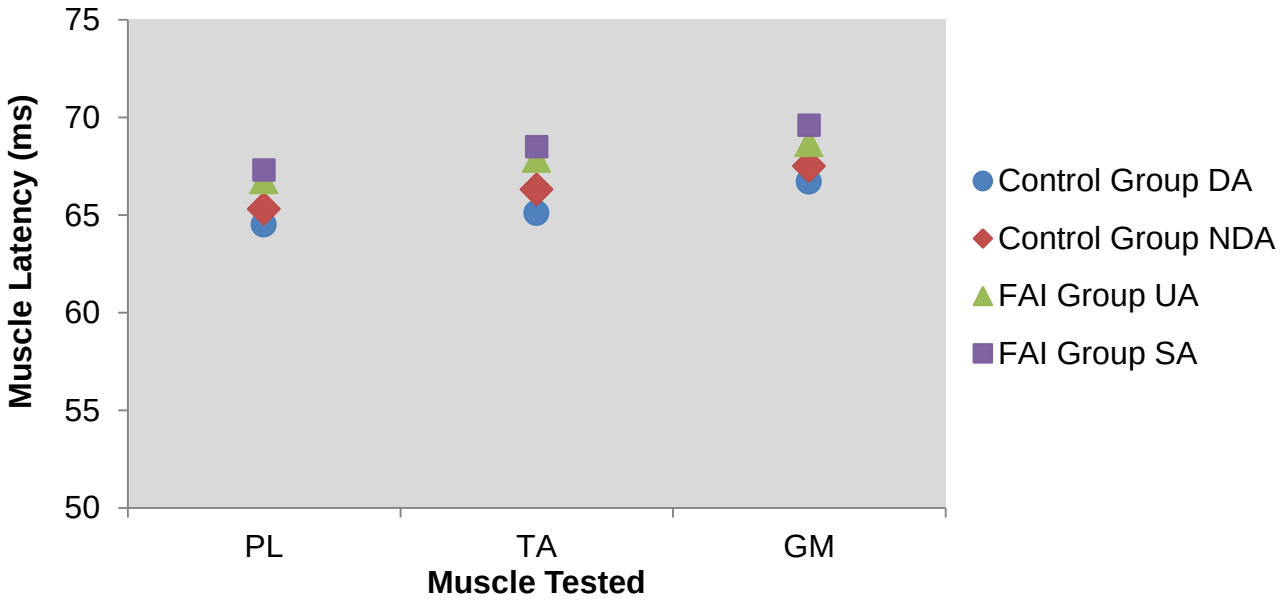


Figure 4.2. Muscle latencies for the Control and FAI Group, when the Support Limb was Studied.

4.1.5 Discussion

The aim of this study was to research muscle latency times of the peroneus longus, tibialis anterior and gluteus medius muscles in the unilateral FAI subject's UA and SA, compared to a healthy control group's DA and NDA, when acting as (i) a tilt limb, and (ii) a support limb. The results of the current study showed that when analysing the tilt limb there was a significant increase (delay) in muscle latency of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to the DA and NDA of the control group. Therefore, the first hypothesis of this study that the FAI subjects will have significantly increased (delayed) muscle latencies in comparison to the healthy controls when acting as the tilt limb can be formally accepted.

When analysing the support limb there were no significant differences in muscle latency of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to the DA and NDA of the control group. Therefore, the second hypothesis of this study that the FAI subjects will have significantly increased (delayed) muscle latencies in comparison to the healthy controls when acting as the support limb can be rejected.

4.1.5.1 Comparison of Results with Current Literature

Articular deafferentiation which was proposed by Freeman et al. (1965) stated that the basic mechanism of ankle instability following ankle injury develops due to the lesion of mechanoreceptors in the joint capsule and ligaments surrounding the ankle. According to this theory, dynamic stability of the ankle joint is dependent on the ability of the

evertors (peronei) to react quickly to sudden inversion perturbations, to develop sufficient tension to prevent injurious ranges of ankle motion, and thus prevent a lateral ligament sprain (Freeman et al., 1965). It has been suggested that an increase (delay) of the response time of the peronei to sudden inversion may have highly significant consequences in terms of risk of injury to the lateral ligaments of the ankle (Wilkerson & Nitz, 1994). The results of the current study agree with the above theory as when studying the tilt limb there was a significant increase (delay) of the muscle latencies in the FAI groups UA and SA when compared to the DA and NDA of the control group.

Konradsen and Ravn (1990) found that FAI subjects exhibited significantly slower peroneus longus and peroneus brevis muscle latencies compared to a healthy control group. Mitchell et al. (2008a) found significantly increased (delayed) muscle latency in the peroneus longus when comparing the FAI subjects injured limb to healthy controls. In addition, Beckman and Buchanan (1995) found significantly increased (delayed) muscle latency of the gluteus medius muscle in the FAI subjects compared to a group of healthy controls, when the tilt limb was studied. The results of the present study agree with the results of Konradsen and Ravn (1990), Mitchell et al. (2008a) and Beckman and Buchanan (1995) as there was a significant increase (delay) of the muscle latencies of the peroneus longus, tibialis anterior and gluteus medius muscles when comparing both the FAI group's UA and SA to a health control groups DA and NDA when the tilt limb was studied. These results immediately offer two interpretations: (1) the patients with unilateral FAI may have a predisposition to FAI, as evidenced by the increased (delayed) muscle latencies on the contra-lateral healthy limb; and (2) FAI affects muscle latencies at a central level that is high enough to influence stability during stance on either extremity.

Isakov et al. (1986) and Vaes et al. (1999) studied the peroneal muscle latencies in the tilting limb in healthy and unstable ankles. The authors did not find a significant difference when comparing the injured ankles of the FAI subjects, to healthy control ankles. These results disagree with the results from the present study as a significant difference was found between the FAI subjects UA and SA when compared to a healthy control groups DA and NDA when the tilt limb was studied.

Mitchell et al. (2008a) found significantly increased (slower) muscle latencies when comparing the injured ankle to the uninjured ankle in a unilateral FAI group when functioning as the tilt limb in the peroneus longus, peroneus brevis and tibialis anterior muscles. Karlsson and Andreasson (1992) also found that the involved limbs of individuals with unilateral FAI demonstrated significantly longer (delayed) peroneus longus and peroneus brevis muscle latencies, when compared to the healthy contra lateral limb. The results of Mitchell et al. (2008a) and Karlsson and Andreasson (1992) support the peripheral mechanism of control as only the injured limb in the unilateral FAI subjects was affected. However, the results from the present study disagree with the above studies as no differences were found between the UA and SA in the FAI group. A difference was only found when comparing the UA and SA of the FAI group to the DA and NDA of a healthy control group, which would suggest a more central mechanism of control, or possibly a genetic predisposition to FAI.

Ebig et al. (1997) found no significant differences between the injured and uninjured ankles in subjects with unilateral FAI for muscle latencies of the peroneal and tibialis anterior muscles. These results agree with the results of the present study, as there was no difference found between the UA and SA of the FAI subjects. However, when we

compared the results of the FAI subjects to a healthy control group there was a significant increase in muscle latency. A major limitation in the study by Ebig et al. (1997) is that they did not use a control group in their study. Therefore, it cannot be concluded if their results would have shown a difference in comparison to healthy controls, as found in the present study.

In the current study, there was no difference between the DA and NDA of the control group. There was also no significant difference between the UA and the SA of the FAI group. These findings agree with the results of Benesch et al. (2000) and Goldie, Evans and Bach (1992) who found no differences between the left and right limbs of the control subjects. This is an important result as there was not expected to be a difference between the DA and NDA of the control subjects, and therefore, these subjects were a good comparison for the FAI subjects.

Very few studies have examined the effect of the contralateral support limb to an ankle sprain mechanism. Beckman and Buchanan (1995) observed the muscle latencies of the peroneals of both the support limb and the tilting limb to an inversion perturbation. The authors found no significant differences in muscle latency in the injured compared to the uninjured limb of FAI subjects, when the tilt limb or support limb was studied. Mitchell et al. (2008a) also found that as a support limb there were no significant differences in muscle latencies between the injured and uninjured limbs of the FAI subjects in the peroneus longus, peroneus brevis, tibialis anterior or extensor digitorum longus muscles. The results of the present study agree with the results of Beckman and Buchanan (1995) and Mitchell et al. (2008a) as no significant differences were found in

muscle latencies for the peroneus longus, tibialis anterior and gluteus medius muscles between the UA, SA, DA and NDA when acting as a support limb.

Lofvenberg et al. (1995) examined support limb muscle latencies of the peroneus longus and tibialis anterior muscles and observed no significant differences between the injured ankle of the FAI subjects and a group of healthy controls. These results show a similarity to the results of the present study as no significant differences in muscle latencies were found between the UA and SA of the FAI group when compared to the DA and NDA of a healthy control group, when the support limb was studied. It is still possible that the support limb may have other as yet unknown influencing factors, which only further research may uncover.

4.1.5.2 Theorised Mechanisms Associated with Results

The results of the current study showed that when analysing the tilt limb there was a significant increase (delay) in muscle latencies of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to the DA and NDA of the control group, which would suggest a more central mechanism of control, or possibly a genetic predisposition to FAI in some individuals.

Studies which have found a delay in peroneal muscle latency (Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990) are in agreement with the finding that slower motor nerve conduction velocities of the superficial peroneal nerve were shown after inversion trauma (Kleinrensink et al. 1994). The principle evertor muscles (peroneus longus, peroneus brevis and extensor digitorum longus) are innervated by the superficial

peroneal nerve, whereas, the invertor muscles (tibialis anterior) are innervated by the deep peroneal nerve. The superficial peroneal nerve rather than the deep peroneal nerve is more likely to be affected by inversion trauma, because of its position in respect to the inversion-eversion axis (Cingel et al., 2006). Therefore, delay of neuromuscular response can be expected in the muscles innervated by the superficial peroneal nerve (Cingel et al., 2006).

The results from the present study found significantly delayed peroneal muscle latencies in the ankles of the FAI subjects, which would agree with the above statement. However, the present study also found a delay in the muscle latency of the tibialis anterior muscle which would disagree with the above statement. As the mechanism of an ankle sprain is usually a combination of forced talocrural joint plantarflexion and subtalar joint inversion (Mitchell et al., 2008a), it seems extraordinary for Cingel et al. (2006) to suggest that only the superficial peroneal nerve would be affected. Many authors seem to ignore the fact that an inversion sprain more often than not involves combined plantarflexion and inversion. This would mean that the deep peroneal nerve is also affected during the sprain mechanism, as shown by the delayed tibialis anterior muscle latency in the present study.

In the current study a deficit in muscle latency was found in the UA and SA of the FAI subjects, in comparison to a healthy control groups DA and NDA, when acting as the tilt limb. Another possible explanation for the injured subjects FAI may be that following an inversion stress to the ankle joint, the mechanoreceptors located within the ligaments and joint capsule may become stretched (Docherty, Arnold et al., 2006). This potentially means that if the mechanoreceptors become permanently lengthened, the protective

control mechanism from the muscles and nerves to prevent inversion occurring will also become delayed, and so there is a higher possibility of an inversion sprain occurring. This could be another reason for the persistent instability suffered by the FAI subjects in the present study.

4.1.5.3 Clinical Implications

The main clinical implications that have arisen from the findings of the present study are that any rehabilitation prescribed by sports injury professionals to subjects with unilateral FAI should ensure the exercises focus on both the UA and the SA, as deficits were present in both limbs of the FAI subjects. The present study also found that deficits did not only exist in the muscles surrounding the ankle joint, but were also present in the more proximal gluteus medius muscle. This finding indicates that sports injury professionals should also include rehabilitation exercises for the gluteus medius muscle.

4.1.5.4 Limitations and Recommendations for Future Research

Only male subjects were recruited for this study. A similar study should be repeated with both male and female subjects to see if further deficits are identified in females. In addition to this, different age groups should be studied to see if the same results occur, or if muscle latency deficits are further affected by age.

Injuries rarely occur with a person standing at rest. However, in the literature it has been stated that to make comparisons there has to be standardisation (Lynch et al., 1996).

The use of a tilt platform is a fairly static task. Future research should investigate muscle activity during activities such as walking, running or jumping to see if greater deficits occur in more dynamic situations.

It has often been stated in the literature that any deficits are exacerbated under the influence of fatigue (Gribble, Tucker & White, 2007). Some would suggest that fatigue, either central or peripheral, may play a role in contributing to the occurrence of lateral ankle sprains (Gutierrez, Jackson, Dorr, Margiotta & Kaminski, 2007). Research on elite soccer players has shown that injury risk is highest in the last 15 minutes of the contest (Rahnama et al., 2002), when fatigue has set in. Further research should investigate the effect of fatigue on muscle latencies in subjects with FAI to see if any further deficits are identified. This is investigated in Study Three of this thesis.

4.1.6 Conclusion

In summary, the results indicate that when analysing the tilt limb there was a significant increase (delay) in muscle latencies of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to the DA and NDA of the control group. However, when analysing the support limb there was no significant difference in muscle latencies of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to the DA and NDA of the control group. As muscle latencies were increased (delayed) in both the UA and SA of the FAI subjects, when compared to the DA and NDA of the healthy control group, this would suggest a more central mechanism of control, or possibly a genetic predisposition to FAI in some individuals.

Chapter Five

Postural Sway Reliability Study

5.1 Introduction to Chapter

This chapter includes Pilot Study Three. The pilot study was undertaken to determine the reliability of the postural sway data that was to be used in Study Two.

5.2 Pilot Study Three: Test-Retest Reliability of Postural Stability Using the AMTI Force Platform; Sampling Rate and Balance Duration

5.2.1 Abstract

Purpose: To evaluate the relative and absolute reliability of postural stability measures, when using different sampling rates (200, 500 and 1000 Hz) and balance timings (200 ms and 3 seconds), in healthy and FAI subjects following a single leg drop jump. **Aim:** To identify the most reliable combination of parameters to measure postural stability. **Method:** Ten males suffering from unilateral FAI and 10 male healthy controls performed 6 single leg drop jump landings, 3 on each leg, onto a force platform and remained balanced for 3 seconds. **Results:** The results highlighted two combinations that provided ICC's above 0.80 and low levels of SEM variance across all conditions, these were 200 Hz/200 ms and 200 Hz/3 seconds. **Conclusion:** These combinations should be considered in future research if the same equipment and protocol is used as in the present study. It is essential that the methods used to assess postural stability are determined as reliable in order to evaluate the extent of balance impairment and/or to determine the effectiveness of interventions.

5.2.2 Introduction

Injuries in the lower extremity, particularly the ankle, are common among young athletes (Roos, Brandsson & Karlsson, 2001; Tropp et al., 1984). Defective muscle function and impaired postural control are often seen following ankle ligament injuries, particularly lateral ankle sprains (Friden et al., 1989; Tropp et al., 1984). Postural control is a complex function involving somatosensory, vestibular and visual functions, as well as muscle activity to maintain the body's COP over the base of support when standing still and during movement (Ageberg, Zatterstrom & Moritz, 1998).

Stabilometry with force platforms is an objective method for the study of postural control in stance (Friden et al., 1989; Johansson & Magnusson, 1991; Tropp et al., 1984). Analysis usually includes a computation of the projection of the COP representing the resultant of gravitational forces and muscular stabilisation forces (Goldie, Bach & Evans, 1989; Goldie et al., 1992; Johansson & Magnusson, 1991). The subjects are examined with either open eyes or blindfolded standing on both legs (Ekdahl et al., 1989; Goldie et al., 1989) and/or on one leg (Friden et al., 1989; Goldie et al., 1989; Goldie et al., 1992; Tropp et al., 1984). The single limb stance is commonly used for evaluation of unilateral injuries (Friden et al., 1989; Goldie et al., 1994; Tropp et al., 1984).

Impaired postural control with increased amplitude and speed of COP movements has been reported in patients with ligament injuries in the lower extremity (Friden et al., 1989; Goldie et al., 1994; Tropp et al., 1984). Using an AMTI force platform, Ekdahl et al. (1989) found acceptable test-retest reliability, but Goldie et al. (1989) found low test-

retest reliability for COP measures when using a Kistler force platform. Alderton and Moritz (1996) found standing balance was unaffected with calf muscle fatigue, but there was a possible learning effect induced by repeat testing. Methods used in the literature often vary greatly which makes comparison of the results difficult.

When observing the sampling rate of the force platforms the rates vary in the literature. The sampling rates ranged from 20 Hz (Ageberg et al., 1998), to 50 Hz (Evans et al., 2004; Hale et al., 2007), to 180 Hz (Ross et al., 2005; Ross and Guskiewicz, 2004; Wikstrom et al., 2006) and to 200 Hz (Wikstrom et al., 2007). There is usually no justification for why this rate is chosen. In terms of reliability no study at present has investigated the sampling rates of the AMTI force platform on relative and absolute reliability. In the present study it can be hypothesised that when the force platform samples at 200 Hz, the least variation will be detected due to the decreased sampling rate, and therefore the most reliable results will be found.

In addition, when examining postural timings in the literature this varies between studies from 3 seconds up to 30 seconds (Baier & Hopf, 1998; Fu & Hui-Chan, 2005; Leanderson et al., 1999; McGuine et al., 2000; Simoneau, Degner, Kramper & Kittleson, 1997; Trojian & McKeag, 2006). No explanation is given by these authors for their balance time chosen, and often the long duration of balancing time is not specific to a 'real' sporting situation. No study to date has analysed postural stability in a subconscious time period (initial 200 ms). The 200 ms time period was identified by Wilkinson and Allison (1989) to be the average fastest reaction time in 20-29 year olds, therefore, anything prior to 200 ms would be beyond human conscious control. Analysis of the subconscious time period (200 ms) may identify postural sway deficits that are

sometimes not present in FAI subjects when analysing a conscious time period. Ross and Guskiewicz (2005) found that FAI subjects took 1.98 seconds to stabilize, whereas healthy controls were significantly quicker at 1.45 seconds. Therefore, the 200 ms may show variable results as the subjects will not have stabilized by this time, however, by 3 seconds stabilization may have occurred, producing less variable and more reliable results. It can therefore be hypothesized that the 3 second analysis will produce the most reliable results.

The aim of this study was therefore to evaluate the relative and absolute reliability of postural stability measures, when using different sampling rates (200, 500 and 1000 Hz) and balance timings (200 ms and 3 seconds), in healthy and FAI subjects following a single leg drop jump. The study aimed to identify the most reliable combination of parameters to measure postural stability.

5.2.3 Method

5.2.3.1 Subjects

The same subjects were used as in Pilot Study One (Section 3.2.3.1)

5.2.3.2 Experimental Design

An AMTI force platform (OR6-7 AMTI, Inc, Watertown, MA), with an AMTI amplifier (AMTI MSA-6 MiniAmp) and NetForce data collection software (Version 2.4.0) quantified postural sway during single limb balancing. The force platform sampled at a

rate of 200, 500 and 1000 Hz, and the amplifier was set at a gain of x1000. Centre of pressure excursions were calculated by the BioAnalysis software (Version 2.3.0).

Subject's age, mass and height were recorded. A warm-up was accomplished by a five minute cycle on a Monark cycle ergometer (Monark, Varberg, Sweden) at 50 rpm with a resistance of 50 Watts. The subject performed a single leg drop jump onto the force platform. The force platform was positioned in front of a 30 cm high wooden bench. Subjects were barefoot and stood on the wooden bench with the test leg relaxed and non-weight bearing. The subject used the contra-lateral leg to propel them from the bench and 'balance' their landing on the test leg on the centre of the force platform. Following landing, subjects remained balanced in a single leg stance for 3 seconds. The force platform triggered as the subjects touched the plate, and data collection stopped after 3 seconds. During the single leg stance, we did not control for arm position, trunk flexion, or lower extremity flexion during foot contact or stance, in order to reflect a more functional balance strategy. Subjects had a 2 minute rest between tests and a total of three trials were performed randomly on each leg (i.e. the UA and SA in the FAI group, and the DA and NDA in the healthy control group). Subjects were retested if they hopped on the weight bearing leg or touched down with their non-weight bearing leg during the trial. This procedure was then repeated with the force platform sampling at 200, 500 and 1000 Hz. Following this the subject performed a five minute cool down on the cycle ergometer, at 50 rpm with a resistance of 50 Watts. To assess test-retest reliability the subjects were required to return to the laboratory 7 days later to repeat the above procedure.

5.2.3.3 Data Analysis

The centre of pressure in the mediolateral direction (*x*) and anteroposterior direction (*y*) was calculated by the BioAnalysis software using the following formulas:

$$COP(x) = [(My + (Zoff * Fx))/Fz] * (-1)$$

$$COP(y) = (Mx – (Zoff * Fy))/Fz$$

where *Zoff* is the vertical offset from the top plate to the origin of the force platform (a negative #), *My* is the moment about the *y* axis, *Mx* is the moment about the *x* axis, *Fy* is the force along the *y* axis, *Fx* is the force along the *x* axis and *Fz* is the force along the *z* axis. The software then calculated the COP excursion distances in the +*x*, -*x*, +*y* and -*y* directions. Mediolateral and anteroposterior COP excursions were calculated by summing the two components on a particular axis. The absolute +*x* and –*x* components were added to provide the mediolateral total, while the absolute +*y* and –*y* components formed the anteroposterior total.

All COP excursion data was analysed over 3 seconds and 200 ms.The mean sway distance (cm) for each direction was calculated, and this mean value was used for statistical analysis.

5.2.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. Systematic bias, which refers to a difference in measurements in a

particular direction between repeated tests, was assessed with 6 (number of combinations) 4 x 2 (ankle [UA, SA, DA or NDA] x time [first week testing or second week testing]) mixed factorial analysis of variance (ANOVA) for each of the sway directions tested (anterior, posterior, anteroposterior, medial, lateral and mediolateral). The within-subject factor was time of test, and the between-subject factor was ankle tested. Sphericity was verified for all data being compared by the Mauchly test. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of homogeneity of intercorrelations. The Multivariate Test box (Wilk's Lambada value) was studied for two-way interactions and then main effects, to identify differences for the within-subject factor (time) (P<0.05). The Test of Between-Subject Effects box was observed to identify differences for the between-subject factor (ankle) (P<0.05). Tukey's post-hoc test was used to determine exactly where the significant findings occurred for the between-subject factor. Due to multiple comparisons being made between groups, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.0125.

Relative reliability was assessed by calculating the $ICC_{(2,1)}$. From the ICC value the SEM was calculated, which represented absolute reliability (Section 3.2.3.4, paragraph 2).

The postural sway data from this pilot study was also used to calculate power, and therefore predict a suitable sample size to be used in Study Two and Study Four of this thesis (Appendix Twelve).

5.2.4 Results

5.2.4.1 Relative Reliability

The relative reliability results showed all 6 combinations produced moderate to excellent reliability (ICC range: 0.63 to 0.93) (Appendix Thirteen). The results from the control group show that when the sampling rate of the force platform was set at 200 Hz the highest relative reliability results were produced (ICC range: 0.80 to 0.91) across all sway directions. When the sampling rate was increase to 500 and 1000 Hz the relative reliability decreased (500 Hz ICC range: 0.77 to 0.86, 1000 Hz ICC range: 0.72 to 0.79) across all sway directions. When the balance timings were observed in the control group, the 3 second analysis produced slightly higher relative reliability results. With the 200 ms analysis the ICC results ranged from 0.72 to 0.87, and with the 3 second analysis the results ranged from 0.72 to 0.91.

The results from the FAI group showed that when the sampling rate of the force platform was set at 200 Hz the highest relative reliability results were produced (ICC range: 0.80 to 0.86) across all sway directions. When the sampling rate was increased to 500 and 1000 Hz the relative reliability decreased (500 Hz ICC range: 0.73 to 0.83, 1000 Hz ICC range: 0.67 to 0.80) across all sway directions. When the balance timings were observed in the FAI group, both timings produced similar relative reliability results. With the 200 ms analysis the ICC results ranged from 0.67 to 0.86, and with the 3 second analysis the results ranged from 0.67 to 0.87.

As previously mentioned, Currier (1990) suggested that an ICC above 0.80 was acceptable for clinical work. There were only two combinations that produced ICC's above this value throughout each table; these were 200 Hz/200 ms and 200 Hz/3 seconds (Table 5.1 and 5.2). Therefore, in future studies it would seem that either of these combinations would be appropriate when analysing postural stability using the same methods and equipment as the present study.

5.2.4.2 Absolute Reliability

The absolute reliability results showed similar trends to the relative reliability results (Appendix Thirteen). The results from the control group show that when the sampling rate of the force platform was set to 200 Hz this produced the lowest SEM variance (range: 2.03 to 9.83%) across all sway directions. When the sampling rate was increased to 500 and 1000 Hz the SEM variance increased (500 Hz range: 2.12 to 10.23%, 1000 Hz range: 2.32 to 10.83%) across all sway directions. When the balance timings were observed in the control group the SEM variance was the lowest with the 200 ms analysis (2.03 to 4.65%) across all sway directions. With the 3 second analysis the SEM variance increased (range: 5.03 to 10.83%) across all sway directions.

The results from the FAI group show that when the sampling rate of the force platform was set to 200 Hz this produced the lowest SEM variance (range: 2.14 to 9.93%) across all sway directions. When the sampling rate was increased to 500 and 1000 Hz the SEM variance increased (500 Hz range: 2.32 to 10.43%, 1000 Hz range: 2.42 to 10.93%) across all sway directions. When the balance timings were observed in the FAI group the SEM variance was the lowest with the 200 ms analysis (2.14 to 4.75%)

across all sway directions. With the 3 second analysis the SEM variance increased (range: 5.13 to 10.93%) across all sway directions. It has been stated that SEM variances below 10% are deemed acceptable. There were only two combinations that produced SEM variances below this value throughout each table; these were 200 Hz/200 ms and 200 Hz/3 seconds (Table 5.1 and 5.2). Therefore, in future studies it would seem that either of these combinations would be suitable.

In addition to the reliability results, the mixed factorial ANOVA found no significant difference in sway distance between the first week of testing and the second week of testing in the FAI groups UA and SA and the controls groups DA and NDA, across all sway directions, during the 3 second analysis.

Table 5.1. Test-Retest Results for the 200 Hz Sampling Rate and 200 ms Balance Time.

Group and Sway Direction	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)
CONTROL GROUP		**Dominant Ankle**				**Non-Dominant Ankle**		
Anterior	3.38 (0.18)	3.45 (0.23)	0.87	2.03	3.22 (0.23)	3.37 (0.22)	0.83	2.11
Posterior	5.12 (0.36)	5.20 (0.32)	0.85	2.13	4.98 (0.32)	5.03 (0.37)	0.80	2.22
Medial	1.60 (0.09)	1.63 (0.11)	0.85	2.25	1.67 (0.12)	1.63 (0.10)	0.82	2.21
Lateral	2.58 (0.16)	2.62 (0.19)	0.84	2.45	2.61 (0.18)	2.67 (0.30)	0.80	2.31
Anteroposterior	8.50 (1.21)	8.65 (1.30)	0.82	4.01	8.20 (1.22)	8.40 (1.25)	0.80	4.11
Mediolateral	4.18 (0.32)	4.25 (0.35)	0.81	4.11	4.28 (0.30)	4.30 (0.41)	0.80	4.21
FAI GROUP		**Unstable Ankle**				**Stable Ankle**		
Anterior	3.45 (0.21)	3.42 (0.23)	0.86	2.14	3.39 (0.18)	3.41 (0.20)	0.82	2.22
Posterior	5.34 (0.43)	5.30 (0.40)	0.83	2.24	5.21 (0.41)	5.16 (0.44)	0.80	2.32
Medial	1.74 (0.11)	1.80 (0.17)	0.83	2.31	1.70 (0.09)	1.72 (0.10)	0.81	2.35
Lateral	4.50 (0.23)	4.54 (0.26)	0.83	2.41	4.42 (0.35)	4.47 (0.28)	0.81	2.36
Anteroposterior	8.79 (0.92)	8.72 (0.90)	0.80	4.11	8.60 (0.90)	8.57 (0.89)	0.80	4.24
Mediolateral	6.24 (0.63)	6.34 (0.62)	0.80	4.22	6.12 (0.64)	6.19 (0.62)	0.81	4.32

Results are presented as mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Table 5.2. Test-Retest Results for the 200 Hz Sampling Rate and 3 Second Balance Time.

Group and Sway Direction	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)
CONTROL GROUP		**Dominant Ankle**				**Non-Dominant Ankle**		
Anterior	7.74 (0.83)	7.65 (0.75)	0.91	5.03	8.56 (0.91)	8.67 (1.01)	0.89	5.56
Posterior	12.53 (1.12)	12.43 (1.03)	0.89	5.14	12.81 (1.21)	12.93 (1.19)	0.84	5.67
Medial	5.23 (0.46)	5.37 (0.45)	0.88	5.23	5.57 (0.52)	5.49 (0.49)	0.89	5.32
Lateral	6.21 (0.71)	6.27 (0.73)	0.86	5.43	6.32 (0.73)	6.29 (0.73)	0.84	5.42
Anteroposterior	20.27 (2.53)	20.08 (2.65)	0.84	9.83	21.37 (2.56)	21.60 (2.69)	0.85	9.72
Mediolateral	11.44 (0.98)	11.64 (0.95)	0.83	9.96	11.89 (0.90)	11.78 (0.94)	0.84	9.82
FAI GROUP		**Unstable Ankle**				**Stable Ankle**		
Anterior	8.46 (1.02)	8.42 (0.98)	0.87	5.13	7.87 (0.84)	7.94 (0.72)	0.85	5.66
Posterior	13.42 (1.04)	13.40 (1.10)	0.87	5.24	13.12 (0.97)	13.12 (0.94)	0.80	5.77
Medial	5.62 (0.57)	5.60 (0.51)	0.85	5.33	5.45 (0.50)	5.49 (0.53)	0.85	5.42
Lateral	6.62 (0.67)	6.69 (0.70)	0.85	5.49	6.75 (0.70)	6.69 (0.73)	0.81	5.48
Anteroposterior	21.88 (2.51)	21.82 (2.59)	0.82	9.93	20.99 (2.48)	21.06 (2.49)	0.83	9.82
Mediolateral	12.24 (1.12)	12.29 (1.03)	0.81	9.98	12.20 (1.16)	12.18 (1.09)	0.81	9.90

Results are presented as mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

5.2.5 Discussion

The aim of this study was to evaluate the relative and absolute reliability of postural stability measures, when using different sampling rates (200, 500 and 1000 Hz) and balance timings (200 ms and 3 seconds), in healthy and FAI subjects following a single leg drop jump. The study aimed to identify the most reliable combination of parameters to measure postural stability. The results highlighted two combinations that provided ICC's above 0.80 and low levels of SEM variance across all conditions, these were 200 Hz/200 ms and 200 Hz/3 seconds. The hypothesis that stated the 3 second analysis will produce the most reliable results can therefore be rejected, as both balance durations produced reliable combinations. The mixed factorial ANOVA also found no significant difference in sway distance between the first week of testing and the second week of testing in the FAI groups UA and SA and the controls groups DA and NDA, across all sway directions, during the 200 ms and 3 second analysis.

The sampling rates of the force platform in the present study were set at 200, 500 and 1000 Hz. The results found that the 200 Hz condition produced the most reliable results in the FAI group and the control group, across all sway directions. It was hypothesised that when the force platform sampled at 200 Hz the least variation would be detected due to the decreased sampling rate, and therefore the most reliable results would be found. This hypothesis can therefore be formally accepted. The studies in the literature employ a variety of sampling rates, often without any explanation of why that level of sampling was chosen.

Using an AMTI force platform, Ekdahl et al. (1989) found acceptable test-retest reliability for COP measures, which would agree with the results of the present study. However, even though the same make of force platform was used, Ekdahl et al. (1989) only sampled at 50 Hz and studied COP measures during standing balance, whereas the present study sampled at 200 Hz and observed COP measures following a single leg drop jump, which makes comparison of results difficult. The present study found that the 200 Hz was the most reliable sampling rate, and therefore in future studies it would seem that this sampling rate would be appropriate when analysing postural stability, but only when using the same methods and equipment as the present study.

The balance analysis times in the present study were set at 200 ms and 3 seconds. The results of the present study found that in the control group the 3 second analysis found slightly higher relative reliability results. However, in the FAI group the reliability results were similar. It was hypothesised that the 3 second analysis would produce the most reliable results; due to the time it takes for the body to stabilize balance. Ross and Guskiewicz (2005) found that FAI subjects took 1.98 seconds to stabilize, whereas healthy controls were significantly quicker at 1.45 seconds. Therefore, the 200 ms may show very variable results as the subjects will not have stabilized by that time, however, by 3 seconds stabilization may have occurred, producing less variable and more reliable results. Even though the time to stabilization was not a variable in the present study, it may be a reason for why there was slightly higher reliability results found in the control group when looking at the 3 second results. This hypothesis can therefore be accepted for the control group. In contrast, the FAI subjects showed similar reliability results between the 200 ms and 3 second analysis, so for these subjects the hypothesis would have to be rejected.

Although high reliability coefficients (such as ICC's) have been previously reported for postural stability studies, SEM values have received little attention in the literature. The SEM is a critical factor which shows the accuracy to which a protocol is reproducible. The SEM value in the present study was expressed as a percentage in order to allow clinical usage of these measures. As demonstrated by the results of the current study, re-test values for the control subjects across all conditions ranged from 2.03 to 10.84% to the initial test, and for the FAI subjects ranged from 2.14 to 10.94% to the initial test. It should therefore, seem appropriate in future studies to attribute differences in postural stability results to injury or intervention, should they exceed the SEM values outlined in Appendix Thirteen.

5.2.5.1 Clinical Implications

As previously mentioned in Pilot Study One (Section 3.2.5.1) the relevance of the reliability findings of this study lies predominantly in the research field. It may be argued that the increase in reliability is marginal between some of the analysis combinations. For example, when observing the 200 Hz/200 ms combination across all subjects and conditions the ICC's range from 0.80 to 0.93, and the SEM's range from 2.08 to 9.98%, when the 500 Hz/200 ms combination is observed the ICC's range from 0.73 to 0.86, and the SEM's range from 2.17 to 10.44%. These changes may seem trivial, but in the field of research where reliable protocols are a necessity, the two reliable combinations found in the present study (200 Hz/200 ms and 200 Hz/3 seconds) improved the reliability of the protocol, and should be considered over the other combinations in future research.

5.2.5.2 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. The results are also only applicable if the same equipment and protocol as the present study are used. Future studies may wish to repeat this study but using different stabliometric devices, and varied protocols.

5.2.6 Conclusion

It is essential that the methods used to assess postural stability are determined as reliable in order to evaluate the extent of balance impairment and/or to determine the effectiveness of interventions. Prior to the results of the present study, there was no available research that investigated force platform sampling rate and subject balance duration on the reliability of COP measures. Most studies do not provided explanations for why they have used a certain sampling rate, or balance time. The present study found that the most reliable combinations that provided ICC's above 0.80 and low levels of SEM variance across all conditions were 200 Hz/200 ms and 200 Hz/3 seconds. These combinations should be considered in future research if the same equipment and protocol is used as in the present study.

5.3 Development of Research

It is essential that the methods used to assess postural stability are determined as reliable in order to evaluate the extent of balance impairment. Pilot Study Three

addressed the issue of sampling rates and balance duration on the force platform following a single leg drop jump. The results from Pilot Study Three highlighted two combinations that provided ICC's above 0.80 and low levels of SEM variance across all conditions, these were 200 Hz sampling rate with 200 ms balance duration, and 200 Hz sampling rate with 3 seconds balance duration. Therefore, these combinations were deemed suitable for use in Study Two and Study Four of the thesis. Pilot Study Three also provided data to calculate power, and therefore predict a suitable sample size to be used in Study Two and Study Four of this thesis.

Chapter Six

Postural Sway in Healthy versus Functionally Unstable Subjects

6.1 Study Two: Postural Sway in Healthy and Functionally Unstable Subjects Following a Single Leg Drop Jump Landing

6.1.1 Abstract

Aim: To investigate single limb postural sway following a drop jump landing over (i) 3 seconds, and (ii) 200 ms, in the unilateral FAI subjects UA and SA, compared to a healthy control groups DA and NDA. **Method:** Twenty males suffering from unilateral FAI and 20 male healthy controls performed 6 single leg drop jump landings, 3 on each leg, from a 30 cm high bench onto a force platform and remained balanced for 3 seconds. **Results:** The results indicated that when analysing the 3 second data there were no significant differences in postural sway for any of the sway directions between the UA, SA, DA and NDA. When analysing the 200 ms data there was a significant (P<0.0125) increase in postural sway in the lateral and mediolateral directions in both the UA and SA of the FAI group when compared to the DA and NDA of the control group. **Conclusion:** The results indicate that the FAI subject's postural control may be decreased, but only on a subconscious level as seen by an increase in lateral and mediolateral sway under the 200 ms analysis. It may be possible that after this initial 200 ms the FAI subject is able to regain control of their stability with conscious postural modifications. Postural sway was increased within the time frame that an ankle sprain would usually occur, and therefore, this increase in sway in the 200 ms time period may be a risk factor for repeated sprains in FAI sufferers. Bilateral deficits were also identified in the FAI subjects, which may indicate that FAI affects the postural control system at a level that is high enough to influence stability on either extremity, or possibly a genetic predisposition to ankle sprains.

6.1.2 Introduction

Impaired postural control is often seen following ankle ligament injuries, particularly lateral ankle sprains (Friden et al., 1989; Tropp et al., 1984). Many studies have identified postural sway deficits in the injured limb compared to the uninjured limb in unilateral FAI subjects (Hale et al., 2007; Harkins et al., 2005; Olmsted et al., 2002). Several authors have found no difference between the injured and uninjured limbs of unilateral FAI subjects, however, when comparing these results with a healthy control group they revealed significantly higher centre of pressure excursions (Evans et al., 2004; Hiller et al., 2007; Tropp et al., 1984). In contrast to this, other research has failed to show that postural sway deficits exist with FAI subjects (Gribble et al., 2006).

Methods of postural sway analysis vary greatly in the literature. One of the most popular methods due to its precision measurement is the force platform, which measures several variables including total sway, peak sway, sway velocity and ground reaction forces (Evans et al., 2004; Fu & Hui-Chan, 2005; Hertel et al., 2001; Hiller et al., 2007; Konradsen & Ravn, 1991; Lofvenberg et al., 1995; Tropp et al., 1984). Other methods of postural sway analysis include the balance error scoring system (Docherty, Arnold et al., 2006) and the SEBT (Gribble, Hertel et al., 2007; Olmstead et al.,2002). However, these methods have been reported to be crude and subjective, and may lack sensitivity when evaluating small changes in postural sway (Hertel et al., 2001).

When examining the methodology of the literature closer, the postural sway analysis times differ between studies from 5 seconds, up to 30 seconds (Baier & Hopf, 1998; Fu & Hui-Chan, 2005; Leanderson et al., 1999; McGuine et al., 2000; Simoneau et al.,

1997; Trojian & McKeag, 2006). No explanation is given by these authors for their balance time chosen, and often the long duration of balancing time is not specific to a 'real' sporting situation. No study to date has analysed postural control in a subconscious time period (initial 200 milliseconds (ms)). The 200 ms time period was identified by Wilkinson and Allison (1989) to be the average fastest reaction time in 20-29 year olds, therefore, anything prior to 200 ms would be beyond human conscious control. Analysis of the subconscious time period (200 ms) may identify postural sway deficits that are sometimes not observed in FAI subjects when analysing a conscious time period.

The aim of this study was therefore to evaluate single limb postural sway following a drop jump landing over (i) 3 seconds, and (ii) 200 ms, in the unilateral FAI subjects UA and SA, compared to a healthy control groups DA and NDA. It was hypothesised that the FAI subjects will have significantly increased postural sway in comparison to the healthy controls during the 3 second analysis. It was also hypothesised that the FAI subjects will have significantly increased postural sway in comparison to the healthy controls during the 200 ms analysis.

6.1.3 Method

6.1.3.1 Subjects

The same subjects were used as in Study One (Section 4.1.3.1).

6.1.3.2 Experimental Design

The same experimental design as Pilot Study Three was used; apart from the force plate sampled at a rate of 200 Hz, and subjects were not required to return to the laboratory seven days later to repeat the procedure (Section 5.2.3.2).

6.1.3.3 Data Analysis

The same data analysis as Pilot Study Three was used (Section 5.2.3.3).

6.1.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. A one-way between-groups ANOVA compared the mean postural sway of the UA, SA, DA and NDA, when the tilt and support limb was studied. Postural sway was the dependent variable, whereas ankle (UA, SA, DA and NDA) was the independent variable. The Levene's Test of Homogeneity of Variances box was observed to verify the assumption of homogeneity of variances across groups
The ANOVA table was studied for significant differences; the alpha level was set at $P<0.05$. Tukey's post-hoc test was used to determine exactly where the significant findings occurred. Due to multiple comparisons being made between groups, a Bonferroni adjustment was carried out, and the alpha level was set at $P<0.0125$.

6.1.4 Results

6.1.4.1 3 Second Postural Sway Data

Results from the between-groups ANOVA for the 3 second analysis showed no significant differences when comparing sway distance for any of the directions (anterior, posterior, anteroposterior, medial, lateral or mediolateral) between the UA, SA, DA and NDA (Table 6.1 and Figure 6.1).

Table 6.1. Sway Distance (cm) for the 3 Second Analysis for the Control and FAI Group.

Sway Direction	Control Group		FAI Group	
	DA	NDA	UA	SA
Anterior	7.74 (0.83)	8.56 (0.91)	8.78 (1.02)	7.52 (0.89)
Posterior	12.56 (1.21)	12.89 (1.11)	13.43 (1.19)	12.96 (0.90)
Medial	5.27 (0.46)	5.68 (0.52)	5.65 (0.59)	5.42 (0.51)
Lateral	6.15 (0.59)	6.21 (0.73)	6.65 (0.70)	6.70 (0.62)

Results presented as Mean (SD). DA: Dominant Ankle, NDA: Non-Dominant Ankle, UA: Unstable Ankle, SA: Stable Ankle

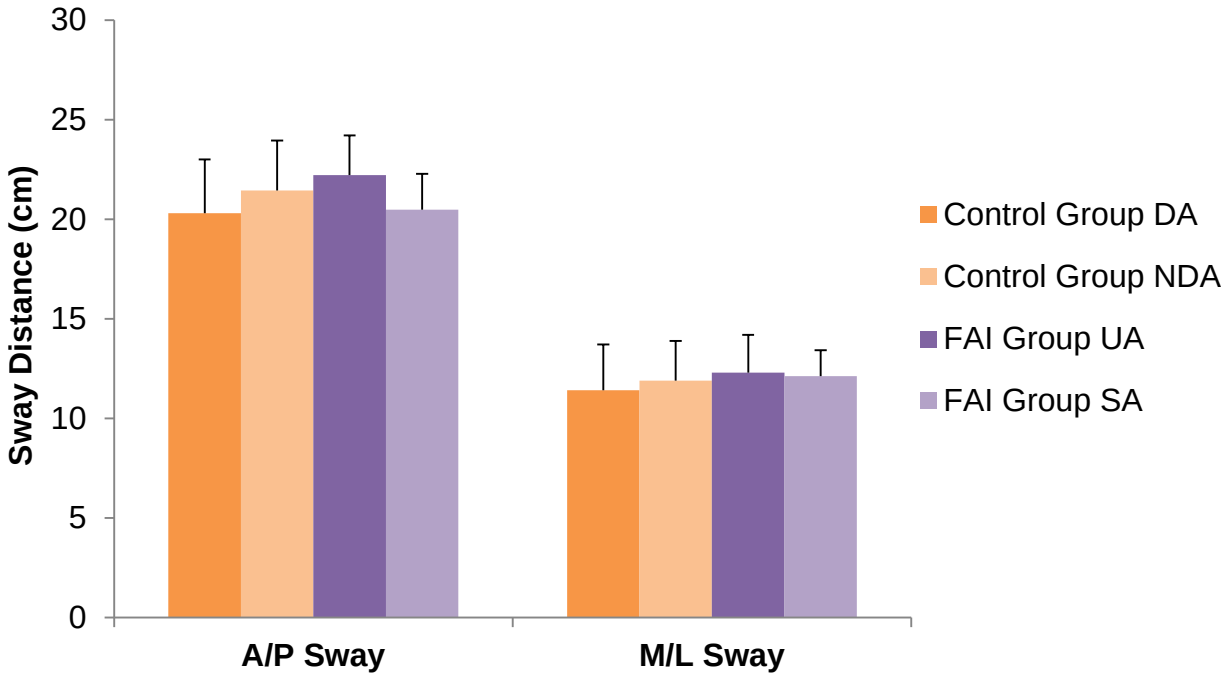


Figure 6.1. Anteroposterior (A/P) Sway and Mediolateral (M/L) Sway in the Control and FAI Group During the 3 Second Analysis (Mean ± SD).

6.1.4.2 200 ms Postural Sway Data

Results from the between-groups ANOVA for the 200 ms analysis showed no significant differences when comparing sway distance for the anterior, posterior, anteroposterior or medial sway directions between the UA, SA, DA and NDA (Table 6.2 and Figure 6.2). However, when comparing sway distances in the lateral and mediolateral directions a significant (P<0.0125) increase was found between the UA of the FAI group to both the DA and NDA of the control group. A significant (P<0.0125) increase in sway distance was also found when comparing the SA of the FAI group to both the DA and NDA of the control group (Table 6.2 and Figure 6.2).

Table 6.2. Sway Distance (cm) for the 200 ms Analysis for the Control and FAI Group.

Sway Direction	Control Group		FAI Group	
	DA	NDA	UA	SA
Anterior	3.38 (0.23)	3.22 (0.18)	3.45 (0.24)	3.42 (0.20)
Posterior	5.12 (0.38)	4.98 (0.32)	5.31 (0.41)	5.23 (0.43)
Medial	1.61 (0.09)	1.67 (0.06)	1.76 (0.11)	1.75 (0.14)
Lateral	2.51 (0.15)	2.61 (0.18)	4.53 (0.24)*	4.40 (0.20)†

Results presented as Mean (SD). DA: Dominant Ankle, NDA: Non-Dominant Ankle, UA: Unstable Ankle, SA: Stable Ankle. * UA significantly (*P*<0.0125) higher than DA and NDA. † SA significantly (*P*<0.0125) higher than DA and NDA.

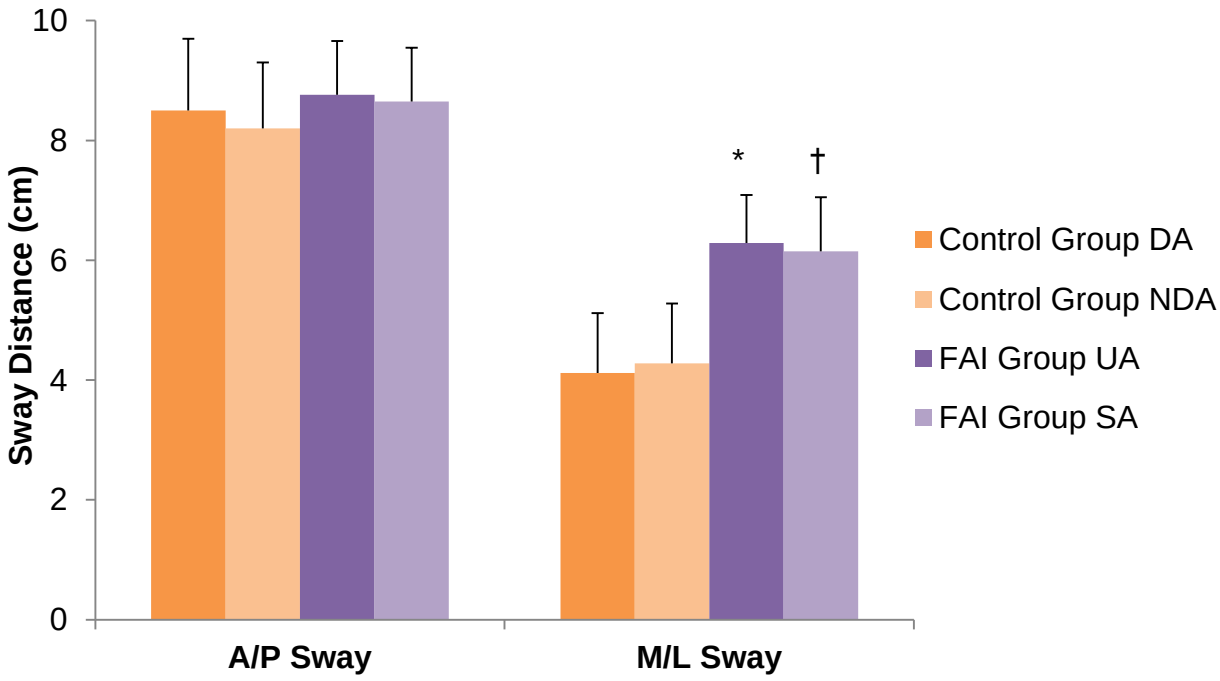


Figure 6.2. Anteroposterior (A/P) Sway and Mediolateral (M/L) Sway in the Control and FAI Group During the 200 ms Analysis (Mean ± SD). * FAI group's UA significantly (*P*<0.0125) different to the control groups DA and NDA. † FAI group's SA significantly (*P*<0.0125) different to the control groups DA and NDA.

6.1.5 Discussion

The aim of this study was to evaluate single limb postural sway following a drop jump landing over (i) 3 seconds, and (ii) 200 ms, in the unilateral FAI subjects UA and SA, compared to a healthy control groups DA and NDA. The results of the current study show that when analysing the 3 second data there were no significant differences when comparing sway distance for any of the sway directions (anterior, posterior, anteroposterior, medial, lateral or mediolateral) between the UA, SA, DA and NDA. Therefore, the first hypothesis of this study that the FAI subjects will have significantly increased postural sway in comparison to the healthy controls during the 3 second analysis can be formally rejected.

The results of the present study also show that when analysing the 200 ms data there were no significant differences when comparing the sway distance for the anterior, posterior, anteroposterior or medial sway directions between the UA, SA, DA and NDA. However, when comparing the sway distances in the lateral and mediolateral directions a significant ($P<0.0125$) increase was found between the UA of the FAI group to both the DA and NDA of the control group. A significant ($P<0.0125$) increase in sway distance was also found when comparing the SA of the FAI group to both the DA and NDA of the control group. Therefore, the second hypothesis of this study that the FAI subjects will have significantly increased postural sway in comparison to the healthy controls during the 200 ms analysis can be partially accepted, as increased sway was only found in the lateral and mediolateral directions.

6.1.5.1 Comparison of Results with Current Literature

Generally, postural equilibrium is established by the integration of 3 components: afferent information from the vestibular, visual and somato-sensory systems, interpretation of the afferent information into a motor command, and finally efferent information that produces the actual movement (Fukuoka et al., 2001; Maurer et al., 2006). In the literature it has been repeatedly argued that if any of these components are disrupted postural control will be affected (Docherty, Valovich-McLeod et al., 2006).

Freeman et al. (1965) hypothesised that because the tensile strength of the mechanoreceptors is less than the connective tissue within which they are embedded, these mechanoreceptors must be disrupted when ankle ligaments and capsules are torn or stretched. Subsequently, Freeman et al. (1965) theorised that disruption of these mechanoreceptors results in decreased sensory input into the central nervous system which may in turn lead to faulty positioning and diminished reflex responses, thus leading to an increased incidence of recurrent ankle sprains. When looking at the results from the 3 second data the current study opposes this theory, however, it must be observed that Freeman et al. (1965) used subjects who had a history of only one ankle sprain rather than multiple ankle sprains, and subjects were tested within 48 hours of incurring the injury. Therefore, the sample was not representative of subjects with FAI. However, when looking at the results from the 200ms data the present study conforms to the ideas of Freeman et al. (1965). The disrupted mechanoreceptors may decrease the FAI subject's postural control, but only on a subconscious level as seen by an increase in lateral and mediolateral sway under the 200 ms analysis. It should be noted that postural sway was increased within the time frame that an ankle sprain would

usually occur, and therefore, this increase in sway in the 200 ms time period may be a risk factor for repeated sprains in FAI sufferers. It may be possible that after this initial 200 ms the FAI subject is able to regain control of their postural stability on a conscious level, and this is supported by there being no differences in postural sway when observing the 3 second data.

Evans et al. (2004), Hiller et al. (2007) and Tropp et al. (1984) found that subjects with FAI did not differ in unilateral stance abilities on the injured versus the uninjured ankles. However, a comparison of both limbs in the subjects with FAI with a healthy control group revealed significantly higher centre of pressure excursions in the lateral direction. The results of Evans et al. (2004), Hiller et al. (2007) and Tropp et al. (1984) are consistent with the 200 ms results of the present study as there was a significant difference in lateral and mediolateral sway when comparing the UA and SA of the FAI subjects to the DA and NDA of the healthy controls. Again in agreement with Evans et al. (2004), Hiller et al. (2007) and Tropp et al. (1984) there were no differences between the UA and SA of the FAI subjects. Therefore, the results of the 200 ms data may show that on a subconscious level the subjects with functionally unstable ankles may have a predisposition to FAI, as evidenced by the decreased performance on the contra-lateral healthy limb, or that FAI affects the postural control system at a central level which may influence stability during stance on either extremity.

The finding of bilateral changes in subjects with unilateral ankle instability has been shown in many studies that used external controls as a comparison (Konradsen & Ravn, 1991; Lofvenberg et al., 1995; Tropp & Odenrick, 1988). These studies found no difference between legs of the subjects with unilateral FAI, but a difference was shown

between these subjects and healthy control subjects. These findings indicate that a central processing problem may be present in people with FAI (Hiller et al., 2007). This supports the theory of motor programme control where receptors from the paired lower limb joints provide afferent information, and damage to one joint and its receptors results in insufficient information reaching higher centres, and therefore precision movement is jeopardised.

In contrast to this, several studies have identified unilateral differences of increased postural sway on the injured limb when studying unilateral FAI subjects (Gribble, Hertel et al., 2007; Harkins et al., 2005; Olmsted et al., 2002). Hale et al. (2007) found that subjects with FAI demonstrated deficits in postural control during the star excursion balance test on the injured limb but not the contra-lateral healthy limb. Mulloy-Forkin et al. (1996) also found that 63% of subjects displayed a balance deficit whilst standing on the injured ankle with eyes closed, compared to the contralateral healthy limb. These findings would support the peripheral control mechanism for postural control, as only the injured side was affected (Olmsted et al., 2002).These results however, contrast the results of the present study as no difference in postural sway was found between the UA and SA of the FAI subjects in the present study.

Several authors have also failed to show differences in balance performances between FAI subjects and healthy controls (Baier & Hopf, 1998; Delahunt, 2007b; Hertel et al., 2001; Leanderson et al., 1999). Hertel et al. (2001) and Leanderson et al. (1999) found no significant differences in postural sway between the injured and uninjured limb in unilateral FAI subjects, when compared to healthy controls, four weeks and ten weeks, respectively, following injury. The 3 second results of the present study are consistent

with the results of Hertel et al. (2001) and Leanderson et al. (1999) as there were no differences found between the UA and SA of the FAI subjects when compared to a healthy control group. However, the results of Hertel et al. (2001) and Leanderson et al. (1999) contrast with the 200 ms data in the present study as the FAI subjects had an increase in lateral and mediolateral sway in the UA and SA when compared to healthy controls.

Researchers should always be cautious when comparing their findings to the results of other studies as subject characteristics such as gender, age group, body weight, training history and injury history must be taken into consideration (Kurdak, Ozgunen, Adas, Zeren, Aslangiray, Yazici et al., 2005). As well as subject characteristics, other issues such as method of postural sway assessment, duration of postural sway test, number of testers, experience of testers and environmental conditions should also be considered.

6.1.5.2 Theorised Mechanisms Associated with Results

It is possible that the FAI subject's mechanoreceptors were disrupted by the injury, and the postural response (when looking at the 200 ms data) was delayed. However, the results from the 3 second data suggest that factors other than damaged mechanoreceptors (due to sprained ligaments) may be the cause of FAI, or perhaps that other afferents are compensating for the injured mechanoreceptors as no significant differences were identified. Muscle and skin afferents may be providing adequate feedback whilst the foot is in contact with the ground, and skin and muscles are being compressed. The results of the current study indicate that if decreased

proprioception is a cause of functional instability, it is not apparent when the FAI subject has conscious control over their postural sway, and that there may be a decrease in proprioception during the first 200 ms of sway which is beyond conscious human control. The FAI subjects in the current study still complained of recurring episodes of the ankle 'giving way' in everyday activities (that would require conscious control), which indicates that some other entity may be the cause of this problem.

There are several other biomechanical reasons to try and explain the increased postural sway measures found in the present study. Wilkerson et al. (1997) proposed that the invertor muscles may play a significant role in preventing loss of postural stability over a fixed foot. When the centre of mass is displaced over a fixed foot with both its medial and lateral borders anchored, the shank moves laterally resulting in closed chain eversion (Wlikerson et al., 1997). When the centre of mass is displaced beyond the lateral border of the foot and the limits of closed chain eversion is reached, the medial border of the foot will begin to rise, subsequently resulting in rapid inversion of the foot. Hence, eccentric activity of the invertor muscles, which control lateral postural stability, may play a significant role in the maintenance of dynamic ankle stability. Thus, if the invertors are weak there may be a bilateral predisposition to inversion sprains. Therefore, in the present study it is possible that the above theory is a reason for the increased lateral and mediolateral sway in the FAI subjects when the 200 ms data was analysed.

When balancing in a single limb stance the foot pronates and supinates in an effort to keep the body's centre of gravity above the base of support, which is referred to as the ankle strategy (Hertel, 2002). Individuals with FAI have been shown to use more of a

hip strategy to maintain unilateral stance. This alteration in postural control strategy is possibly due to changes in central neural control that occur in the presence of ankle joint dysfunction (Hertel, 2002). It is possible that the subjects in the current study did display an increased reliance on the hip strategy. However, video analysis was not used in the current study so frontal plane hip movement could not be assessed. When analysing the 3 second data, even though the unilateral FAI subjects did not display an increase in postural sway, it is possible that they may have still used more of a hip strategy; it just might be that they were as efficient at using the hip strategy as they were the ankle strategy (due to repetitive long term injury) and so no postural sway deficits were apparent. When the 200 ms data was analysed there was a significant increase in lateral and mediolateral postural sway between the UA and SA of the FAI subjects and the DA and NDA of the healthy controls. It may be possible that during the initial 200 ms the FAI subjects are more reliant on the hip strategy, but because they are unable to consciously control this, there is an increase in postural sway, which may mean an increased risk of suffering an inversion ankle sprain.

In agreement with this proposed theory, Van Deun et al. (2007) found that during the transition from a double leg stance position to a single leg stance position there was a later onset of the gluteal and hamstring muscles in subjects with FAI compared to healthy control subjects. Van Deun et al. (2007) concluded that impairments in muscle activation are not only present in structures around the injured ankle but also exist around other joint complexes, such as the hip. The authors concluded that one possible explanation is that the central nervous system decreases the reliance on proprioceptive information from one location where this source of information is confounded, and increases the reliance on input from other locations that provide reliable information for

maintaining postural balance. This has been defined as sensory re-weighting (Van Deun et al., 2007), and it is possible that the FAI subjects in the current study used this to remain balanced and therefore no postural sway deficits were identified when analysing the 3 second data. However, when the 200 ms data was analysed and significant increases in lateral and mediolateral sway were apparent in the FAI subjects, it may be possible that the impairments from the injured structure are not picked up by the central nervous system immediately, therefore, sensory re-weighting cannot occur instantly, and this may be a reason for the increases in postural sway that were detected in the FAI subjects during their subconscious time frame.

6.1.5.3 Clinical Implications

The main clinical implications that have arisen from the present study are that rehabilitation exercises prescribed by sports injury professionals to subjects with unilateral FAI should ensure that the exercises focus on both the UA and SA, as deficits in postural sway were present in both limbs of the FAI subjects. In addition, if clinicians have access to the use of force platforms they should consider analysing a subconscious time period, as well as the more common conscious time scales, as the present study only found deficits in FAI subjects under the 200 ms analysis.

6.1.5.4 Limitations and Recommendations for Future Research

Only male subjects were recruited for this study. A similar study should be repeated with both male and female subjects to see if the same subconscious lateral and mediolateral postural sway deficits occur in females. In addition to this, different age groups should

be studied to see if the same deficits occur. Wilkinson and Allison (1989) stated that the average fastest reaction time in the 20-29 year olds was approximately 200 ms (Wilkinson & Allison, 1989). It was found that reaction time was fastest in the 20's, declining rapidly below that age and more gradually above it, such that the 20's were significantly faster than the teens and under 10's, but when compared to the older age groups they were only significantly faster than the decades 50 and above (Wilkinson & Allison, 1989). This would mean that if different age groups were studied, the subconscious time period would have to be adjusted accordingly.

It has previously been reported that postural equilibrium is controlled by the afferent information from the vestibular, visual and somato-sensory systems (Fukuoka et al., 2001; Maurer et al., 2006). The present study did not control for visual or vestibular cues. Further research should look at the effect of blindfolding a subject, minimising vestibular signs by wearing headphones, and the effect of a combination of both. It would be interesting to see if the subconscious postural sway deficits were increased when visual and vestibular cues were removed, but it would also be intriguing to see if the conscious postural sway scores showed any significant differences between FAI subjects and healthy controls.

It has often been stated in the literature that any deficits are exacerbated under the influence of fatigue (Gribble, Hertel et al., 2007). Some would suggest that fatigue, either central or peripheral, may play a role in contributing to the occurrence of lateral ankle sprains (Gutierrez et al., 2007). Research on elite soccer players has shown that injury risk is highest in the last 15 minutes of the contest (Rahnama et al., 2002), when fatigue has set in. Further research should investigate the effect of fatigue on postural

sway in subjects with FAI to see if any further deficits are identified. This is investigated in Study Four of this thesis.

6.1.6 Conclusion

In summary, the results indicate that the FAI subject's postural control may be decreased, but only on a subconscious level as seen by an increase in lateral and mediolateral sway under the 200 ms analysis. It may be possible that after this initial 200 ms the FAI subject is able to regain control of their stability with conscious postural modifications. Postural sway was increased within the time frame that an ankle sprain would usually occur, and therefore, this increase in sway in the 200 ms time period may be a risk factor for repeated sprains in FAI sufferers. We recommend that future researchers investigate this subconscious time period, as longer time frames may not be representative of the time period that an ankle sprain occurs within. Bilateral deficits were also present in the FAI subjects, which may indicate FAI affects the postural control system at a level that is high enough to influence stability on either extremity, or possibly a genetic predisposition to FAI in some individuals.

Chapter Seven

Isokinetic Fatigue Reliability Studies

7.1 Introduction to Chapter

This chapter includes Pilot Studies Four to Nine. These pilot studies were undertaken to establish the reliability of the localised ankle and hip isokinetic fatigue protocols to be used in Studies Three and Four.

7.2 Pilot Study Four: The Effect of Isokinetic Testing Speed on the Reliability of Muscle Fatigue Indicators During an Ankle Inversion-Eversion Fatigue Protocol

7.2.1 Abstract

Aim: To investigate the reliability of fatigue indicators calculated from peak torque and total work during isokinetic speeds of 60, 90, 120 and 180° · s^{-1} during an ankle fatigue protocol. **Method:** Ten males suffering from unilateral FAI and ten male healthy controls performed five maximal inversion-eversion concentric contractions on an isokinetic dynamometer. Following a four minute rest period, subjects were instructed to perform repeated maximal inversion-eversion concentric contractions to fatigue, which was defined as three consecutive repetitions below 50% of the maximum peak torque value. Each testing speed was randomised with 24 hours between speeds. Subject's returned to the laboratory 7 days later for repeat testing, identical to the first week. Muscle fatigue was determined for each testing speed by the fatigue index, the percent decrease in performance and the slope of the regression equation. **Results:** The most reliable fatigue determination method was the slope of the regression equation, when testing at a speed of 120° · s^{-1}. **Conclusion:** Clinicians can now perform an isokinetic fatigue

protocol on the ankle evertors with the reassurance that the procedure is reliable in both healthy individuals, and individuals with a history of FAI.

7.2.2 Introduction

Ankle injuries, specifically lateral ligament sprains, are a common sport related problem (Garrick, 1977; Garrick, 1987; Jackson, Ashley & Powell, 1974; Moseley & Chimenti, 1995; Ruth, 1961). These injuries result in more time loss than any other single injury in athletics (Garrick, 1987). Residual ankle deficits following an acute lateral ligament sprain has been well documented (Bosien, Staples & Russell, 1955; De Carlo & Talbot, 1986; Freeman, 1965a, Freeman, 1965b, Freeman et al., 1965, Rijke, Jones & Vierhout, 1988; Tropp et al., 1985). Symptoms include loss of strength (Bosien et al., 1955), decreased joint position sense (Glencross & Thornton, 1981), delayed peroneal muscle reaction time (Hertel, 2000), altered common peroneal nerve function (Hertel, 2000), decreased postural stability as compared with the uninjured limb (Freeman, 1965b) and as compared with a noninjured group of subjects (Tropp, 1986; Tropp et al., 1985) and FAI (Evans, Hardcastle & Frenyo, 1984; Freeman, 1965a). Freeman (1965a) described FAI as a “feeling of giving way.” It is a symptom often found in individuals who suffer repeated ankle sprains.

The strength of the ankle evertors has been a popular area of research in relation to FAI patients. The evertor muscles are often described as playing a major role in the prevention of ligamentous injuries (Willems et al., 2002).The strength of the evertors, specifically peroneus longus and peroneus brevis, have been suggested to provide support to the lateral ligaments (Glick, Gordon & Nishimoto, 1976) and resist sudden

inversion during a lateral ankle sprain (Willems et al., 2002). While some studies have reported a decrease in the strength of the ankle evertors after inversion sprain when tested manually (Bosien et al., 1955; Staples, 1975; Staples, 1972) or isokinetically (Tropp, 1986), others have reported no decrease in strength as compared with the uninjured ankle when tested isokinetically (Lentell et al., 1990).

In recent years isokinetic dynamometry has become a popular method to objectively measure muscle fatigue (Gleeson & Mercer, 1992; Larsson, Karlsson, Eriksson & Gerdle, 2003; Pincevero, Gear & Sterner, 2001). Fatigue has been defined as “any reduction in the force generating capacity of the total neuromuscular system regardless of the force required in any given situation” (Bigland-Richie & Woods, 1984). The majority of studies using isokinetic methods have focused on peak torque, rather than total work (Gleeson & Mercer, 1996). Peak torque represents the highest point of the moment-angular position curve (Bosquet et al., 2010), however, it may not accurately describe the overall modification of the curve. This is why total work, which specifically represents the area under the curve, should also be considered (Hislop & Perrine, 1967).

There also seems to be a lack of agreement in the literature regarding the most appropriate technique to determine fatigue. Thorstensson and Karlsson (1976) originally proposed that muscle fatigue should be determined via the fatigue index (FI), calculated as the ratio of the mean peak torque of the last three contractions to the mean peak torque of the first three contractions (Bosquet et al., 2010). Another method that has been commonly used in the literature is the slope of the regression equation (Pincevero et al., 2001). This method considers the linear relationship between the total work of

each contraction and the number of maximal contractions, providing the rate of decrease of total work and thus an estimation of muscle fatigue (Pincevero et al., 2001). The final method that has been frequently used in the literature is the percent decrement score (Glaister, Stone, Stewart, Hughes & Moir, 2004). Although not specific to isokinetic dynamometry, the suitability of this method has been argued since it considers data from each effort in its calculation (Glaister et al., 2004).

In relation to isokinetic testing of the ankle musculature, it has been consistently demonstrated that peak torque and total work are reliable measures (Amaral De Noronha & Borges Junior, 2004; Aydog, Aydog, Cakci, & Doral, 2004; Kaminski & Dover, 2001; Leslie, Zachazewski & Browne, 1990). The validity of isokinetic dynamometry has also repeatedly been demonstrated (Drouin, Valovich-McLeod, Shultz, Gansneder & Perrin, 2004; Houweling, Head & Hamzeh, 2009; Janssen & Le-Ngoc, 2009; Orri & Darden, 2008; Zawadzki, Bober & Siemienski, 2010). Taylor, Sanders, Howick & Stanley (1991) demonstrated the mechanical validity of the Biodex isokinetic dynamometer in relation to human torque, joint position and limb velocity.

Reliability studies are frequently performed on healthy populations (Bosquet et al., 2010, Brown, Whitehurst, Bryant & Buchalter, 1993; Sole, Hamren, Milosavljevic, Nicholson & Sullivan, 2007; Taylor et al., 1991), however, isokinetic dynamometry is commonly used to test subjects that are recovering from injury. Amaral De Noranha and Borges Junior (2004) stated “it can be a mistake to assume that reliable tests for healthy subjects will be just as reliable when testing subjects with pathologic conditions.” Many sufferers of FAI go long periods of time without suffering an ankle sprain, and are therefore termed healthy patients but with a history of FAI. Clinicians and health

professionals in sport will often use isokinetic dynamometry to test this population, as well as healthy individuals, throughout the sporting season. It is therefore important that the equipment used is reliable in both healthy subjects, and patients with a history of FAI (Gautrey, Watson and Mitchell, 2013a).

Therefore, the purpose of this study was to determine the effect of isokinetic testing speed on the relative and absolute reliability of the fatigue index, percent decrease in performance and slope of the regression equation during an ankle inversion-eversion fatigue protocol, in subjects with FAI and healthy controls.

7.2.3 Method

7.2.3.1 Subjects

The same subjects were used as in Pilot Study One (Section 3.2.3.1)

7.2.3.2 Experimental Design

Subject's age, mass and height were recorded. All testing was carried out on the Biodex System 2 Isokinetic Dynamometer (Biodex Medical Systems, Shirley, New York). The reliability of the Biodex dynamometer has been shown to be high, with ICC's ranging from 0.92-0.98 for peak torque and 0.88-0.97 for total work (Brown et al., 1993). The Biodex isokinetic dynamometer was set up according to the Biodex System 2 Manual, and was calibrated according to manufacturer's specifications prior to testing. The cushion control was set to zero, to allow the subject the greatest availability of velocity

attainment prior to deceleration (Brown, Whitehurst, Gilbert & Buchalter, 1995). All subjects completed a practice session on the isokinetic dynamometer a week prior to the main testing procedure.

The seat of the isokinetic dynamometer was set at 0° orientation and reclined to 15°. The powerhead was set at 0° orientation and tilted to an angle of 90°. The footplate was set up for inversion/eversion movement by aligning the green and red dot. The footplate was affixed to the powerhead shaft so the red dots aligned, and was secured in place with the locking knob. The footplate was positioned perpendicular to the floor (heelcup at bottom) with 45° of footplate tilt, so that the footrest faced the positioning chair. The subject was seated on the chair, and instructed to extend their leg so that their barefoot rested on the footplate in 10° of plantarflexion. The footplate was adjusted so that the powerhead shaft aligned with the ankle inversion/eversion axis of rotation, while the tibia was horizontal to the floor. The ankle axis of rotation was located through the fibular malleolus and the body of the talus. Subject's knee flexion was 30°, and hip flexion was 60°. The multi support pad was then installed. The pad was positioned under the calf, distal to the knee, to support the limb with the desired degree of hip and knee flexion. The heel support of the footplate was then adjusted to maintain proper vertical position of the foot. The foot straps were then tightened to secure the foot in place. The subject was stabilized with the shoulder straps, pelvic strap and multi-support strap, and the opposite leg was secured using the thigh strap attached to the chair (Figure 7.1). The remote comfort stop was placed in the subject's hand. All dynamometer setup positions were recorded to ensure identical patient set-up on the return visits.

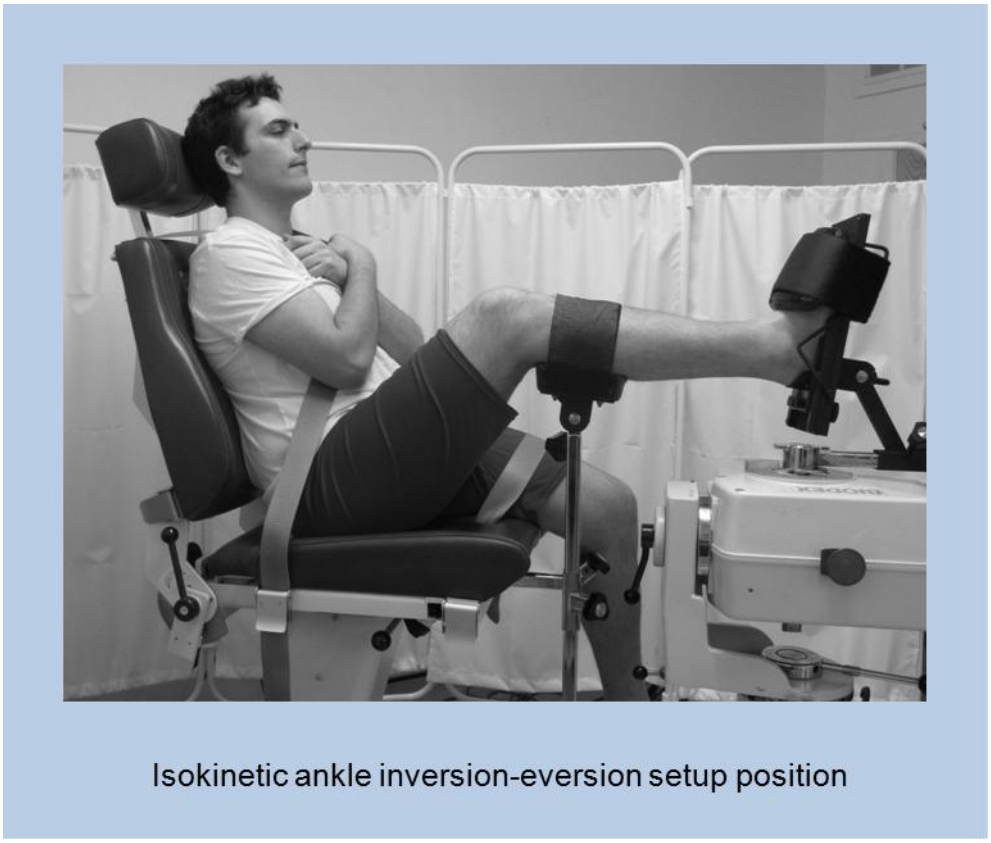


Figure 7.1. Isokinetic Ankle Inversion-Eversion Setup Position. Subject's knee flexion was set at 30° and hip flexion was set at 60°.

The subject's range of motion was set to 20° inversion and 15° eversion (Porter & Kaminski, 2004). The subject was then instructed to perform five concentric maximal repetitions, to determine their maximum peak torque. Each subject began in full inversion and was instructed to push their foot outwards (eversion) and pull their foot inwards (inversion) as hard and as fast as possible. The maximum peak torque value was established and subjects were given a four minute rest period (Salavati et al., 2007). Following this, subjects were instructed to evert and invert their ankle repeatedly as hard and as fast as possible until they reached fatigue. Fatigue was defined as three consecutive repetitions below 50% of the maximum peak torque value (Emery, Maitland & Meeuwisse, 1999; Gautrey et al., 2013a; Gear, 2011; Salavati et al., 2007). The same

strong verbal encouragements were given to each subject throughout the test to motivate them to develop maximal torque during each repetition (McNair, Depledge, Brettkelly & Stanley, 1996). Each testing speed (60, 90, 120 and 180° · s^{-1}) was randomised with a minimum of 24 hours between speeds. The subjects were then given a seven day rest period, and were asked to return to the laboratory to repeat the four speeds, with a minimum of 24 hours between speeds (Gautrey et al., 2013a). Therefore, each subject visited the laboratory on eight separate occasions to complete all testing sessions. Subjects were asked to refrain from any vigorous exercise during the week, and were tested at the same time of day to reduce the effect of diurnal variation.

7.2.3.3 Data Analysis

Many authors have identified that an inverse relationship exists between load range and velocity during concentric contractions (Brown, Whitehurst, Gilbert et al., 1995; Gautrey, Watson & Mitchell, 2013b). It has been stated by Brown, Whitehurst, Gilbert et al. (1995) that if the pre-set velocity is not reached the result is an absence of machine offered resistance. In the present study, all velocities (60, 90, 120 and 180° · s^{-1}) were reached by all subjects, and load ranged showed an inverse relationship to velocity (Figure 7.2). Therefore, all peak torque and total work data was reduced for load range prior to analysis.

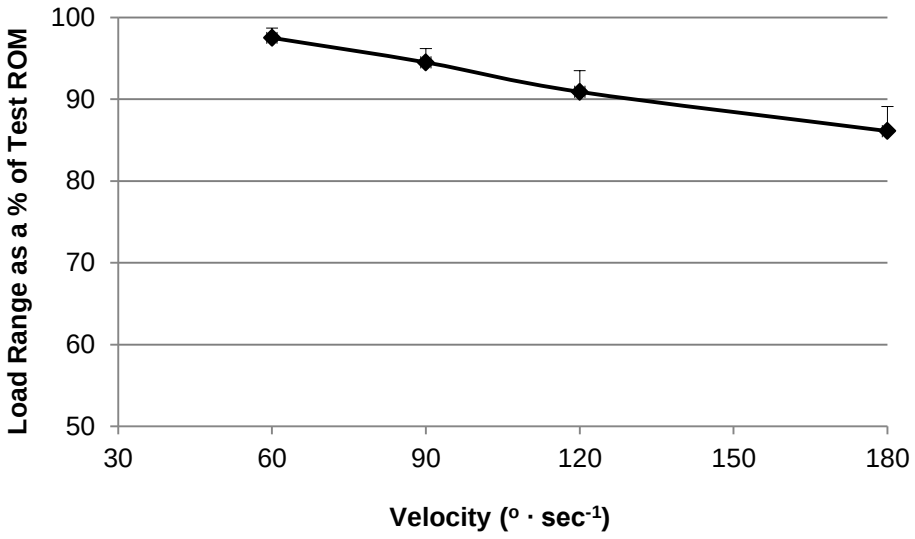


Figure 7.2**.** Load Range as a Percentage of the Total Test Range of Motion (ROM) During Ankle Eversion (Mean ± SD).

Peak torque (N·m) and total work (J) were determined for each repetition, following load range reduction, and summed to compute cumulated performance. Muscle fatigue was only determined for the ankle evertors, as the peroneus longus and peroneus brevis were the focused muscles for the fatigue protocol, due to their protective function of resisting inversion during an ankle sprain. Ankle evertor muscle fatigue was determined when the procedure was performed at 60, 90, 120 and 180° · s^{-1}. Fatigue was calculated using three methods: the fatigue index (Kannus, 1994), the slope of the regression equation (Pincevero et al., 2001) and the percent decrease in performance (Glaister et al., 2004). The fatigue index was calculated by the following equation:

FI = 100 – ((Mean performance of last 3 reps/Mean performance of first 3 reps) x100)

Where performance represented peak torque or total work. The slope was determined via linear regression by plotting performance (i.e. peak torque or total work) against

each repetition, for each subject. The slope of the regression line was added to the graph, and using the mathematical formula y = *m*x + c, the *m* value was taken as the slope, which represented the rate of decrease in performance during the test. The percent decrease in performance was calculated by the following equation:

DP = 100 – ([Cumulated performance/(Maximal performance x n)] x 100)

Where performance represented peak torque or total work, maximal performance represented peak torque max or total work max and n was the number of repetitions.

7.2.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. Systematic bias, which refers to a difference in measurements in a particular direction between repeated tests, was assessed with seven (peak torque [3 fatigue indicators], total work [3 fatigue indicators] and number of repetitions to fatigue) 2 x 4 x 2 (subjects type [healthy or FAI] × speed [60, 90, 120 and 180° · s^{-1}] × time [first week testing or second week testing]) mixed factorial analysis of variance (ANOVA). The two within-subject factors were speed and time of test, and the between-subject factor was subject type. Sphericity was verified for all data being compared by the Mauchly test. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of homogeneity of intercorrelations. The Multivariate Test box (Wilk's Lambada value) was studied for three-way interactions, then two-way interactions and then main effects, to

identify differences for the within-subject factors (speed and time) (P<0.05). The Test of Between-Subject Effects box was observed to identify differences for the between-subject factor (subject type) (P<0.05). The Pairwise Comparisons post-hoc test was used to determine exactly where the significant findings occurred for the within-subject factor when there were more than two conditions (speed). Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.0125.

Relative reliability was assessed by calculating the $ICC_{(2,1)}$. From the ICC value the SEM was calculated, which represented absolute reliability (Section 3.2.3.4, paragraph 2).

7.2.4 Results

7.2.4.1 Peak Torque

The mixed factorial ANOVA showed no significant differences for the fatigue indicators (fatigue index, percent decrease in performance and the slope of the regression equation) between the first week and second week of testing. There was also no significant difference between the two groups tested (healthy and FAI). However, there was a significant decrease (P<0.0125) in peak torque with each increase in velocity. When studying the relative reliability results for the healthy subject's peak torque (Table 7.1) the values ranged from moderate to very high depending on which fatigue determination method was used. The fatigue index showed moderate to high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.70, 0.79, 0.59

and 0.75, respectively. Absolute reliability for the fatigue index showed SEM values of 0.60%, 1.21%, 1.40% and 1.51%, for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method showed high to very high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.86, 0.90, 0.85 and 0.77, respectively. Absolute reliability for the percent decrease in performance showed SEM values of 1.73%, 0.77%, 1.59% and 0.88%, for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation showed very high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.91, 0.90, 0.94 and 0.91, respectively. Absolute reliability for the slope of the regression equation showed SEM values of 0.05%, 0.03%, 0.03% and 0.04% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The results showed the slope of the regression equation to be the most reliable method of fatigue determination, when testing at a speed of 120° · s^{-1}.

When studying the relative reliability results for the FAI subjects the peak torque values (Table 7.1) ranged from moderate to very high depending on which fatigue indicator was used. The fatigue index showed moderate to high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.65, 0.81, 0.58 and 0.68, respectively. Absolute reliability for the fatigue index showed SEM values of 1.30%, 2.04%, 0.43% and 0.53% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method showed high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.86, 0.77, 0.78 and 0.82, respectively. Absolute reliability for the percent decrease in performance showed SEM values of 1.53%, 1.86%, 0.60% and 0.62% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation showed high to very high reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.88, 0.90, 0.93 and 0.85,

respectively. Absolute reliability for the slope of the regression equation showed SEM values of 0.04%, 0.02%, 0.02% and 0.10% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The results showed the slope of the regression equation to be the most reliable method of fatigue determination, when testing at a speed of 120° · s^{-1}.

7.2.4.2 Total Work

The mixed factorial ANOVA showed no significant differences for the fatigue indicators (fatigue index, percent decrease in performance and the slope of the regression equation) between the first week and second week of testing. There was also no significant difference between the two groups tested (healthy and FAI). However, there was a significant decrease (P<0.0125) in total work with each increase in velocity. When studying the relative reliability results for the healthy subject's total work (Table 7.2) the values ranged from moderate to very high depending on which fatigue determination method was used. The fatigue index showed moderate reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.52, 0.60, 0.65 and 0.64, respectively. Absolute reliability for the fatigue index showed SEM values of 1.59%, 0.62%, 1.78% and 2.46% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method showed high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.76, 0.84, 0.89 and 0.85, respectively. Absolute reliability for the percent decrease in performance showed SEM values of 1.41%, 0.76%, 0.86% and 1.06% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation showed high to very high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.90, 0.91, 0.95 and 0.89, respectively. Absolute reliability for the slope of the regression equation

showed SEM values of 0.04%, 0.03%, 0.02% and 0.09% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The results showed the slope of the regression equation to be the most reliable method of fatigue determination, when testing at a speed of 120°.

When studying the reliability results for the FAI subjects the total work values (Table 7.2) ranged from moderate to very high depending on which fatigue determination method was used. The fatigue index showed moderate to high reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.89, 0.66, 0.55 and 0.66, respectively. Absolute reliability for the fatigue index showed SEM values of 2.32%, 0.62%, 2.42% and 1.26% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method showed high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.75, 0.71, 0.81 and 0.72, respectively. Absolute reliability for the percent decrease in performance showed SEM values of 1.59%, 0.74%, 0.96% and 1.97% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation showed high to very high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.87, 0.85, 0.93 and 0.90, respectively. Absolute reliability for the slope of the regression equation showed SEM values of 0.04%, 0.02%, 0.02% and 0.04% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The results showed the slope of the regression equation to be the most reliable method of fatigue determination, when testing at a speed of 120° · s^{-1}.

Table 7.1. Muscle Fatigue Indicators from Peak Torque Data During Ankle Eversion. Data are presented as Mean (SD).

PARAMETER	HEALTHY SUBJECTS				FAI SUBJECTS			
	TEST 1	TEST 2	ICC	SEM (%)	TEST 1	TEST 2	ICC	SEM (%)
60° · s^{-1}								
FI (%)	75.59 (6.73)	71.38 (6.93)	0.70	0.60	76.57 (6.73)	74.72 (6.75)	0.65	1.30
DP (%)	45.78 (4.05)	47.98 (4.12)	0.86	1.73	47.89 (4.05)	45.67 (4.11)	0.86	1.53
Slope (Nm·rep^{-1})	-0.51 (0.16)	-0.57 (0.15)	0.91	0.05	-0.58 (0.16)	-0.55 (0.15)	0.88	0.04
N^{o} Reps to Fatigue	48.00 (3.71)	50.00 (3.76)	0.60	1.21	47.00 (3.55)	49.00 (3.48)	0.69	1.10
90° · s^{-1}								
FI (%)	73.00 (6.46)	73.75 (6.37)	0.79	1.21	74.35 (6.73)	71.45 (6.86)	0.81	2.04
DP (%)	44.29 (3.92)	43.22 (3.75)	0.90	0.77	49.10 (4.05)	46.31 (3.90)	0.77	1.86
Slope (Nm·rep^{-1})	-0.54 (0.15)	-0.49 (0.13)	0.90	0.03	-0.56 (0.16)	-0.54 (0.17)	0.90	0.02
N^{o} Reps to Fatigue	48.00 (3.65)	47.00 (4.11)	0.55	1.03	47.00 (3.65)	49.00 (3.45)	0.60	1.21
120° · s^{-1}								
FI (%)	70.44 (6.47)	70.49 (6.29)	0.59	1.40	72.19 (6.76)	72.78 (6.73)	0.58	0.43
DP (%)	42.87 (3.82)	40.74 (3.90)	0.85	1.59	46.71 (3.99)	47.54 (4.05)	0.78	0.60
Slope (Nm·rep^{-1})	-0.57 (0.16)	-0.51 (0.13)	0.94	0.03	-0.58 (0.14)	-0.55 (0.16)	0.93	0.02
N^{o} Reps to Fatigue	46.00 (3.55)	49.00 (3.76)	0.63	1.09	44.00 (3.44)	48.00 (3.23)	0.59	1.01
180° · s^{-1}								
FI (%)	67.75 (6.46)	65.77 (6.57)	0.75	1.51	70.18 (6.67)	69.50 (6.80)	0.68	0.53
DP (%)	40.80 (3.92)	39.72 (3.99)	0.77	0.88	45.61 (3.86)	44.90 (4.06)	0.82	0.62
Slope (Nm·rep^{-1})	-0.53 (0.15)	-0.51 (0.17)	0.91	0.04	-0.56 (0.16)	-0.54 (0.16)	0.85	0.10
N^{o} Reps to Fatigue	42.00 (3.65)	45.00 (3.45)	0.64	1.23	42.00 (3.87)	44.00 (3.76)	0.63	1.09

Table 7.2. Muscle Fatigue Indicators from Total Work Data During Ankle Eversion. Data are presented as Mean (SD).

PARAMETER	HEALTHY SUBJECTS				FAI SUBJECTS			
	TEST 1	TEST 2	ICC	SEM (%)	TEST 1	TEST 2	ICC	SEM (%)
60° · s^{-1}								
FI (%)	79.08 (6.95)	75.76 (7.13)	0.52	1.59	74.50 (7.10)	74.17 (6.70)	0.89	2.32
DP (%)	53.15 (2.67)	52.31 (2.59)	0.76	1.41	51.84 (2.62)	49.89 (2.91)	0.75	1.59
Slope (Nm·rep^{-1})	-0.32 (0.09)	-0.36 (0.11)	0.90	0.04	-0.37 (0.11)	-0.36 (0.11)	0.87	0.04
N^{o} Reps to Fatigue	48.00 (3.43)	50.00 (3.76)	0.60	1.21	47.00 (3.45)	49.00 (3.44)	0.69	1.10
90° · s^{-1}								
FI (%)	75.16 (6.83)	75.16 (6.83)	0.60	0.62	72.07 (6.77)	71.94 (6.91)	0.66	0.62
DP (%)	49.54 (2.68)	50.26 (2.79)	0.84	0.76	48.00 (2.85)	47.06 (2.77)	0.71	0.74
Slope (Nm·rep^{-1})	-0.29 (0.07)	-0.33 (0.06)	0.91	0.03	-0.35 (0.08)	-0.33 (0.08)	0.85	0.02
N^{o} Reps to Fatigue	48.00 (3.66)	47.00 (3.76)	0.55	1.03	47.00 (3.56)	48.00 (3.66)	0.60	1.21
120° · s^{-1}								
FI (%)	75.93 (6.56)	73.94 (6.63)	0.65	1.78	70.88 (6.85)	68.90 (6.74)	0.55	2.42
DP (%)	47.36 (8.26)	46.17 (8.34)	0.89	0.86	45.89 (2.79)	44.64 (2.68)	0.81	0.96
Slope (Nm·rep^{-1})	-0.34 (0.07)	-0.36 (0.08)	0.95	0.02	-0.36 (0.07)	-0.34 (0.08)	0.93	0.02
N^{o} Reps to Fatigue	46.00 (3.54)	49.00 (3.54)	0.63	1.09	44.00 (3.44)	48.00 (3.76)	0.59	1.01
180° · s^{-1}								
FI (%)	75.25 (6.56)	74.04 (6.61)	0.64	2.46	70.06 (6.93)	68.01 (7.03)	0.66	1.26
DP (%)	46.30 (8.26)	43.74 (8.11)	0.85	1.06	44.82 (2.86)	43.26 (2.77)	0.72	1.97
Slope (Nm·rep^{-1})	-0.36 (0.06)	-0.33 (0.07)	0.89	0.09	-0.33 (0.08)	-0.32 (0.07)	0.90	0.04
N^{o} Reps to Fatigue	42.00 (3.65)	45.00 (3.60)	0.64	1.23	42.00 (3.76)	44.00 (3.87)	0.63	1.09

7.2.4.3 Number of Repetitions to Fatigue

The mixed factorial ANOVA showed no significant differences for the number of repetitions to fatigue between the first week and second week of testing. There were also no significant differences between the two groups tested (healthy subjects and FAI subjects), or the four speeds tested (60, 90, 120 and 180° · s^{-1}). The number of repetitions to fatigue was correlated to the ICC reliability values to see if a relationship was present. The healthy subjects produced *r* values of 0.57 and 0.13, for testing session 1 and testing session 2, respectively. The FAI subjects produced *r* values of 0.30 and 0.18, for testing session 1 and testing session 2, respectively. The results showed there was no relationship present when correlating the number of repetitions to fatigue with the ICC reliability value.

7.2.5 Discussion

The aim of this study was to examine the test-retest reliability of muscle fatigue indicators calculated from peak torque and total work during isokinetic speeds of 60, 90, 120 and 180° · s^{-1} during an isokinetic ankle inversion/eversion fatigue protocol. The main findings that emerged from the study were firstly, the slope of the regression equation was the most reliable method of fatigue determination in healthy subjects and FAI subjects , when using peak torque or total work values, and secondly, the most reliable fatigue measures occurred at the speed of 120° · s^{-1}.

The choice of either peak torque or total work to assess average performance during a fatigue test did not demonstrably influence relative or absolute reliability. The same

conclusion applied to the speed of the isokinetic dynamometer as relative and absolute reliability values were not influenced by a change in speed. When observing the peak torque values at 60, 90, 120 and 180° · s^{-1} the relative reliability values (ICC) for the slope of the line measure were consistently between 0.90-0.94 for the healthy subjects and 0.85-0.93 for the subjects with FAI. Absolute reliability (SEM) also produced consistently low values between 0.03%-0.05% for the healthy subjects, and 0.02%-0.10% for the subjects with FAI. The same can be observed with the total work values for the different isokinetic dynamometry speeds, as relative reliability for the slope of the line measure were between 0.89-0.95 for healthy subjects, and 0.85-0.93 for the subjects with FAI. Absolute reliability values were again consistent with the total work measure producing values between 0.02%-0.09% for the healthy subjects and 0.02%-0.04% for the subjects with FAI. It is also apparent from the above results that the type of subjects tested (healthy subjects or FAI subjects) did not influence relative of absolute reliability results.

However, the different fatigue determination methods did produce large variations in relative and absolute reliability values. The slope of the line measurement consistently produced high relative and absolute reliability values. Whereas, the fatigue index and the percentage decrease in performance produced lower and more variable relative and absolute reliability values (Tables 7.1 and 7.2).

A limited number of studies have looked at the reliability of different fatigue measures. Bosquet et al. (2010) found high relative reliability (peak torque ICC's = 0.82-0.88, total work ICC's = 0.81-0.87) for the slope of the line method. Pincivero et al. (2001) studied the reliability of the fatigue index and the slope of the line during isokinetic quadriceps

femoris muscle fatigue. They found moderate to high ICC's for the non-dominant leg (0.78-0.92) and high ICC's for the dominant leg (0.82-0.89) when analysed by the slope of the line method (Pincevero et al., 2001). These results are similar to those from the present study as we found the slope of the line to be the most reliable method when observing both relative and absolute reliability.

The appropriateness of a method to objectively quantify muscle strength or endurance is dependent upon its reliability and the inherent error associated with that method. Kaminski, Perrin, Mattacola, Szczerba and Bernier (1995) illustrated moderate to high test-retest reliability (ICC's: 0.69-0.91) during concentric ankle inversion-eversion on the isokinetic dynamometer. Leslie et al. (1990) showed that peak torque measurements during isokinetic ankle inversion-eversion at 30 and 120° · s^{-1}, displayed high test-retest reliability with ICC's ranging from 0.72-0.89. In addition to this Karnofel, Wilkinson and Lentell (1989) tested concentric ankle inversion-eversion at 60 and 120° · s^{-1}, and found that all test-retest coefficients were above 0.78. Aydog et al. (2004) reported that ankle inversion in healthy young adults were highly reliable (ICC's: 0.92-0.96), and for eversion values ranged from 0.87-0.94. Amaral De Noronha and Borges Junior (2004) were the only authors found who studied the reliability of ankle inversion-eversion in individuals with FAI. They found that the results were reliable with ICC's ranging from 0.71-0.95. It should be recognised that the ability of reproducing the testing protocol with respect to adequate calibration, gravity correction, and standard patient set up in the current study was likely to have improved accuracy, and should be deemed important components for improving the reliability of a test (Gross, Huffman, Phillips & Wray, 1991; Munro, 1997; Pincevero, Lephart & Karunakara, 1997; Winter, Wells & Orr, 1981).

The accuracy to which these protocols are reproducible is also a critical factor as determined by the SEM. Although high reliability coefficients, such as ICC's, have been previously reported for isokinetic strength, SEM values have received little attention in the literature. The SEM value in the present study was expressed as a percentage in order to allow clinical usage of these measures. As demonstrated by the results of the current study, re-test values for peak torque and total work varied by 0.02 – 2.46% to the initial test. It should therefore, seem appropriate in future studies to attribute differences in isokinetic results to intervention, training improvements or injury, should they exceed the SEM values outlined in tables 7.1 and 7.2.

There seems to be a lack of consensus in the literature on the most appropriate or reliable speed to be used for ankle isokinetic dynamometry. The ankle has been well documented with authors opting for a range of speeds from $30° \cdot s^{-1}$ (Amaral De Noronha & Borges Junior, 2004; Kaminski & Dover, 2001; Kaminski, Perrin & Gansneder, 1999; Leslie et al., 1990; Willems et al., 2002), to $60° \cdot s^{-1}$ (Aydog et al., 2004; Kaminski et al., 1999), to $90° \cdot s^{-1}$ (Bernier et al., 1997; Kaminski et al., 1999; Kaminski et al., 1995), to $120° \cdot s^{-1}$ (Amaral De Noronha & Borges Junior, 2004; Kaminski & Dover, 2001; Kaminski et al., 1999; Leslie et al., 1990; Willems et al., 2002), to $150° \cdot s^{-1}$ (Kaminski et al., 1999), to $180° \cdot s^{-1}$ (Aydog et al., 2004; Kaminski et al., 1999) and $240° \cdot s^{-1}$ (Hartsell & Spaulding, 1999). The majority of studies that have investigated the reliability of concentric ankle inversion-eversion have selected a slower speed of $30° \cdot s^{-1}$, and a faster speed of $120° \cdot s^{-1}$ (Amaral De Noronha & Borges Junior, 2004; Kaminski & Dover, 2001; Leslie et al., 1990) and have found that these speeds were reliable (Amaral De Noronha & Borges Junior, 2004; Kaminski & Dover, 2001;

Leslie et al., 1990). The present study is in agreement with this finding as we found 120° · s^{-1} to be the most reliable testing speed.

Granata, Abel and Damiano (2000) found that during walking at the subjects' freely selected pace the maximum dorsi-flexion angular velocity was 135° · s^{-1}, and maximum plantar-flexion angular velocity was 200° · s^{-1} in healthy subjects. The present study found 120° · s^{-1} to be the most reliable testing speed, so even though this speed may be far from 'explosive sporting movement' velocities, it may replicate speeds from more endurance based activities as shown by Granata et al. (2000).

Both peak torque and total work decreased during the fatigue protocol. Three methods were used to quantify this force reduction: the fatigue index (Kannus, 1994), percentage decrease in performance (Glaister et al., 2004) and the slope of the regression equation (Pincevero et al., 2001). The fatigue index and the percent decrease in performance measure the percentage of force reduction throughout the trial. The slope represents the rate of decrease in performance. The main assumption, stated by Bosquet et al. (2010), for using this measure is the linearity of the relationship between peak torque or total work and the number of repetitions.

Previous studies have reported a linear relationship between peak torque or total work and the number of repetitions during 20 (Maffiuletti, Bizzini, Desbrosses, Babault & Munzinger, 2007) and 30 (Thorstensson & Karlsson, 1976) maximal concentric contractions. Hence, the slope could be used to quantify muscle fatigue (Gerdle & Elert, 1994; Larsson et al., 2003). Bosquet et al. (2010) stated there was a tendency of the line to plateau after 40 repetitions, and suggested that an exponential model would be

more appropriate than a linear one to fit performance data measured for longer protocols. However, the present study did not use a fixed number of repetitions, and one subject reached 56 repetitions before 3 contractions were below 50% of their maximum peak torque. This subject still presented with a linear model, rather than an exponential decrease which would contrast with the above literature. The above studies were all performed on the knee, whereas the present study was fatiguing the ankle musculature. We would recommend that future investigators examine and plot their data before choosing the slope of the line as their fatigue determination method, as a linear model is required. As a point of interest the number of repetitions to fatigue was correlated to the ICC reliability values to see if a relationship was present. However, the results showed no correlation between these two variables.

There is limited research that has focused on a fatiguing protocol of the ankle musculature; most research investigates peak torque with between 3 (Bernier et al., 1997; Kaminski et al., 1999; Kaminski et al., 1995; Willems et al., 2002) and 5 (Amaral De Noronha & Borges Junior, 2004; Aydog et al., 2004; Kaminski & Dover, 2001, Willems et al., 2002) maximum repetitions. Current theory suggests that the ankle evertors play a crucial role in the prevention of ligamentous injuries (Willems et al., 2002).The strength of the evertors, specifically peroneus longus and peroneus brevis, have been suggested to provide support to the lateral ligaments (Glick et al., 1976) and resist sudden inversion during a lateral ankle sprain (Willems et al., 2002). Decreased strength of the ankle evertors, potentially brought on by fatigue, has been proposed as one of the possible causes of FAI (Bosien et al., 1955). Therefore, it was crucial to develop a reliable ankle fatigue protocol for the ankle evertors, so that research can continue to investigate this phenomenon.

7.2.5.1 Clinical Implications

The results from the current study showed that the isokinetic dynamometer was a reliable device for testing the fatigability of the ankle evertors in healthy individuals but also individuals with FAI. Many individuals in the sporting population suffer from a history of FAI, and the results from this study conclude that clinicians and other health professionals can perform isokinetic testing protocols on the ankle evertors with confidence that the protocol is reliable, in not only healthy individuals, but also the large population of individuals with a history of FAI.

7.2.5.2 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. It must be remembered that the results are only applicable if the same equipment and protocol is used as in the current study. Future studies may wish to repeat this study but using different makes of isokinetic dynamometers and varied protocols.

7.2.6 Conclusion

In summary, the most reliable fatigue determination method for the ankle evertors was the slope of the regression equation, when testing at a speed of $120° \cdot s^{-1}$. However, future investigators should examine and plot their data before choosing this as their fatigue indicator, as a linear model is required. The choice of either peak torque or total work to assess performance during a fatigue protocol did not demonstrably influence

relative or absolute reliability. The test-retest reliability that was performed in the current study has valuable research and clinical relevance. Many athletic or rehabilitation activities typically involve numerous bouts of testing. The protocols and methods used for testing should always be established as reliable before testing commences, so that differences found can be reported as true. Clinicians can now perform an isokinetic fatigue protocol on the ankle evertors with the reassurance that the procedure is reliable in both healthy individuals, and individuals with a history of FAI.

7.3 Development of Research

Pilot Study Four addressed the issue of identifying the most reliable isokinetic speed to be used for Study Three and Four. The results identified $120° \cdot s^{-1}$ as the most reliable testing speed during isokinetic ankle inversion-eversion. Therefore, this speed was deemed suitable for use in Study Three and Four.

7.4 Pilot Study Five: Test-Retest Reliability of Three Setup Positions During Isokinetic Ankle Inversion-Eversion Exercise

7.4.1 Abstract

Aim: To compare the test-retest reliability of three setup positions during isokinetic ankle inversion-eversion exercise, and to investigate the effect of setup position on peak torque and total work. **Method:** Sixteen male healthy subjects performed three maximal concentric ankle inversion-eversion repetitions at 60, 120, 180, 240, 300 and 360° · s^{-1}, during 10° dorsiflexion, neutral dorsiflexion/plantarflexion, and 10° plantarflexion. Setup position was randomised with 24 hours between testing sessions. Subjects returned to the laboratory 7 days later for repeat testing. **Results:** The results indicated that the 10° plantarflexion position was the most reliable setup, with ICC results ranging from 0.84-0.95 for peak torque and total work, at speeds 60 through 240° · s^{-1}, during ankle inversion and eversion. The SEM results for the 10° plantarflexion position were also the least variable, ranging from 2.56-9.90% for peak torque at speeds 60 through 360° · s^{-1}, and 2.00-9.90% for total work at speeds 60 through 300° · s^{-1}, during ankle inversion and eversion. The results also showed significantly greater (P<0.0167) peak torque and total work values for the 10° plantarflexion position. **Conclusion:** Clinicians should consider adopting this new ankle setup, as it most accurately represented the peak performance of the muscles tested.

7.4.2 Introduction

In recent years, the role of the ankle invertors and ankle evertors have become of great interest (Cawthorn, Cummings, Walker & Donatelli, 1991; Grey & Basmajian, 1968). Many clinicians and sports injury professionals have recognised that the ankle musculature plays an essential role, especially in relation to stabilisation of the ankle joint (Osternig, 1986; Staples, 1975; Staples, 1972; Willems et al., 2002). Strength testing of the ankle musculature using the isokinetic dynamometer is often undertaken by sports injury professionals and is of great importance for screening, rehabilitation and injury prevention purposes (Amaral De Noronha & Borges Junior, 2004; Aydog et al., 2004; Bernier et al., 1997; Bosien et al., 1955; Kaminski & Dover, 2001; Kaminski et al., 1995; Kaminski et al., 1999; Lentell et al., 1990; Tropp, 1986; Wilkerson et al., 1997). Injuries such as ankle sprains may be preventable if the risk factors can be addressed (Kovaleski, Heitman, Trundle & Gilley, 1995), however, the success of such screening procedures depends on the accuracy and reproducibility of the methods used.

Throughout the heel-off to toe-off phase of gait, the ankle moves from approximately 10° dorsiflexion at heel-off to 25° plantarflexion at toe-off (Cawthorn et al., 1991; Murray, 1967). Electromyographic studies have found that the ankle musculature, specifically gastrocnemius, peroneus longus, peroneus brevis, soleus and tibialis posterior, have increased muscle activity during this phase of gait (Cawthorn et al., 1991; Grey & Basmajian, 1968). The muscles that control foot inversion and eversion are most active between 10° dorsiflexion and 25° plantarflexion (Cawthorn et al., 1991). Therefore it is reasonable to suggest that strength testing of these muscles should be performed at a position within this range of ankle movement.

Several investigators have demonstrated that peak torque values are directly affected by the position in which the limb is tested (Aydog et al., 2004; Cawthorn et al., 1991; Leslie et al., 1990). Cawthorn et al. (1991) tested ankle inversion and eversion in three positions (10° dorsiflexion, neutral dorsiflexion/plantarflexion, and 10° plantarflexion) at $160° \cdot s^{-1}$. Cawthorn et al. (1991) concluded that 10° plantarflexion was better than the other positions because reliability was highest and torque output was greatest at this position. These findings indicated that the neutral ankle position that is commonly used in isokinetic inversion-eversion strength testing (Sepic, Murray, Mollinger, Spurr & Gardner, 1986; Wong, Glasheen-Wray & Andrews, 1984), may not be optimal, and the test position should be carefully chosen in order to most accurately represent the peak performance of the muscles tested. From these findings it can be hypothesised that in the present study the 10° plantarflexion position will produce the highest reliability and the greatest peak torque and total work values.

The primary aim of this study was therefore to compare the relative and absolute reliability of three setup positions (10° dorsiflexion, neutral dorsiflexion/plantarflexion, and 10° plantarflexion) during isokinetic ankle inversion-eversion exercise, across a velocity spectrum of 60 to $360° \cdot s^{-1}$. The secondary aim of this study was to investigate the effect of setup position on peak torque and total work.

7.4.3 Method

7.4.3.1 Subjects

Sixteen male subjects (age = 22.2 ± 2.1 years, height = 178.8 ± 4.2 cm, and mass = 78.4 ± 4.9 kg) volunteered to participate in the study. Institutional ethical approval was granted for this study. All subjects read the subject briefing document (Appendix One) and provided written informed consent (Appendix Two) before participation. Inclusion criteria consisted of males, aged 18-25 years, who participated in semi-professional football (two training sessions and one match per week) and who were right leg dominant. The dominant leg was defined as the preferred kicking leg and in the unilateral FAI group the right ankle was the unstable ankle.

Subjects were excluded from the study if they were under the influence of alcohol or any other psycho-active substance, if they had a cold, flu, inner ear or sinus infection in the last two weeks, if they suffered from any musculo-skeletal injuries, knee or hip injuries, fractures to the lower limbs, visual impairments, vestibular deficits, or signs of injury such as pain and/or swelling in their ankles. Subjects were also excluded if they had ever been told by a doctor that they should not exercise, if they did not participate in regular (≥2 x week) aerobic exercise, and if they did not feel fully fit and eager to act as a subject (Appendix Three).

7.4.3.2 Experimental Design

Subject's age, mass and height were recorded. A warm-up was accomplished by a five minute cycle on a Monark cycle ergometer (Monark, Varberg, Sweden) at 50 rpm with a resistance of 50 Watts. Testing was performed on the Biodex System 2 Isokinetic Dynamometer (Biodex Medical Systems, Shirley, New York). The system reliability of the Biodex dynamometer has been shown to be high, with ICC's ranging from 0.92-0.98 for peak torque and 0.88-0.97 for total work (Brown et al., 1993). Taylor et al. (1991) also demonstrated the mechanical validity of the Biodex isokinetic dynamometer in relation to human torque, joint position and limb velocity.

Apparatus Setup

The Biodex was set up according to the Biodex System 2 Manual, and was calibrated according to manufacturer's specifications prior to testing. The cushion control was set to zero, to allow the subject the greatest availability of velocity attainment prior to deceleration (Brown, Whitehurst, Gilbert et al., 1995; Taylor et al., 1991). All subjects completed a practice session on the isokinetic dynamometer a week prior to the main testing procedure.

The seat of the isokinetic dynamometer was set at 0° orientation and reclined to 15°. The powerhead was set at 0° orientation and tilted to an angle of 90°. The footplate was set up for inversion/eversion movement by aligning the green dot to the red dot. The footplate was affixed to the powerhead shaft so the red dots aligned, and was secured in place with the locking knob. The footplate was positioned perpendicular to the floor (heelcup at bottom), so that the footrest faced the Biodex chair. The subject was seated

on the chair, and instructed to extend their leg so that their barefoot rested on the footplate.

The footplate was then positioned into one of the three testing positions: 10° dorsiflexion, neutral dorsiflexion/plantarflexion, or 10° plantarflexion (Figure 7.3). The footplate was adjusted so that the powerhead shaft aligned with the ankle inversion/eversion axis of rotation, while the tibia was horizontal to the floor. The ankle axis of rotation was located through the fibular malleolus and the body of the talus. Subject's knee flexion was 30°, and hip flexion was 60°. The multi support pad was then installed. The pad was positioned under the calf, distal to the knee, to support the limb with the desired degree of hip and knee flexion. The heel support of the footplate was then adjusted to maintain proper vertical position of the foot. The foot straps were then tightened to secure the foot in place. The subject was stabilized with the shoulder straps, pelvic strap and multi-support strap, and the opposite leg was secured using the thigh strap attached to the chair (Figure 7.1).

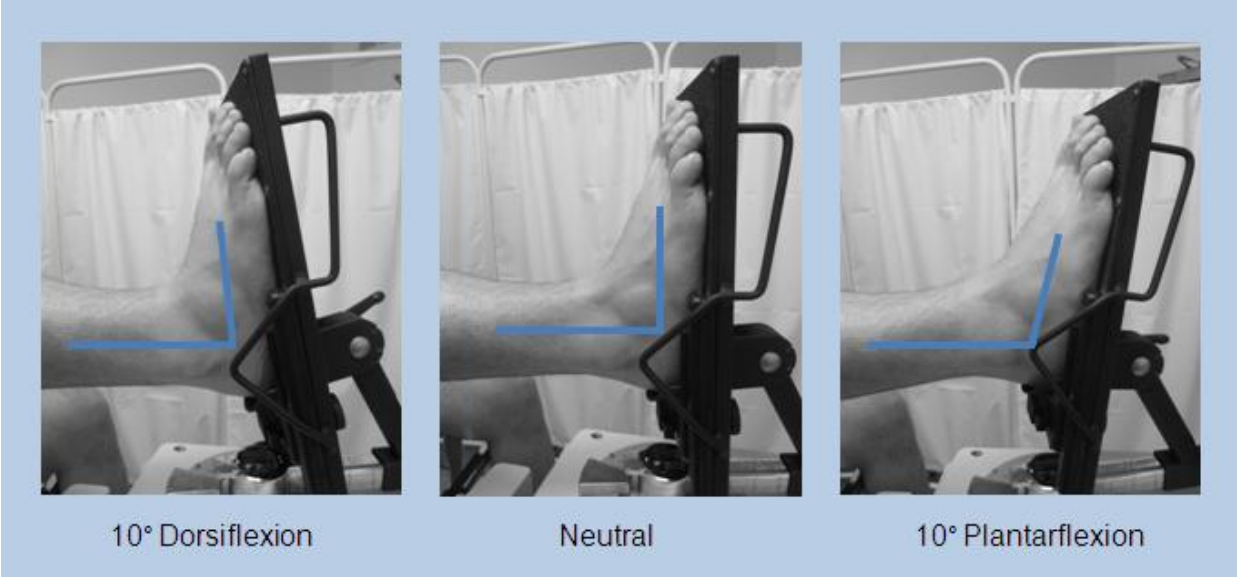


Figure 7.3. Three Different Isokinetic Ankle Inversion-Eversion Setup Positions

Testing Protocol

The subject's range of motion was set to 20° inversion and 15° eversion (Porter & Kaminski, 2004). A warm-up on the isokinetic device consisted of three submaximal reciprocal concentric inversion and eversion repetitions with increasing intensity (i.e. first repetition at 25% perceived effort, second repetition at 50% perceived effort, and third repetition at 75% perceived effort) (Brown, Whitehurst, Gilbert et al., 1995), at 60° · s^{-1} through 360° · s^{-1} (Brown et al., 1993; Gautrey et al., 2013b; Timm & Fyke, 1993). In addition the subject completed two maximal intensity repetitions at each speed (Brown, Whitehurst, Gilbert et al., 1995; Findley, Brown, Whitehurst, Keating, Murray & Gardner, 2006).

Testing began from a dead stop (Brown, Whitehurst, Gilbert et al., 1995) with the subject's ankle in 20° of inversion and consisted of three maximal concentric reciprocal ankle eversion and inversion gravity corrected repetitions in a fixed order at 60, 120, 180, 240, 300 and 360° · s^{-1}, with a 30 second rest between velocities (Timm & Fyke, 1993). Each subject was encouraged to contact the mechanical end stops during both inversion and eversion movements. The same verbal encouragement was given to each subject throughout the test to motivate them to develop maximal torque during each repetition (McNair et al., 1996) but no visual feedback of torque generation was provided.

Setup position (10° dorsiflexion, neutral dorsiflexion/plantarflexion, and 10° plantarflexion) was randomised with 24 hours between positions. Subjects returned to the laboratory 7 days later for repeat testing, identical to the first week.

7.4.3.3 Data Analysis

Data was collected using the Biodex Advantage Software (version 4.5, Biodex Medical Systems, Shirley, New York). It has previously been shown that there is an inverse relationship between load range and velocity during concentric contractions (Brown, Whitehurst, Gilbert et al., 1995; Gautrey et al., 2013b). Brown, Whitehurst, Gilbert et al. (1995) stated if the pre-set velocity is not reached the result is an absence of machine offered resistance. All velocities were reached by all subjects in the present study, and load range demonstrated an inverse relationship to velocity, as previously found by Gautrey et al. (2013b). Therefore, prior to the analysis of peak torque and total work, all data was reduced for load range.

Peak torque was determined for each condition by locating the highest point of the curve within the load range ROM. Total work was determined by calculating the area under the curve within the load range ROM. All torque data was then normalised with respect to the subject's body weight (Kurdak et al., 2005).

7.4.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. Systematic bias, which refers to a difference in measurements in a particular direction between repeated tests, was assessed with four (peak torque [ankle inversion and ankle eversion] and total work [ankle inversion and ankle eversion]) 3 x 6 x 2 (setup position [10° dorsiflexion, neutral dorsiflexion/plantarflexion, and 10° plantarflexion] x speed [60, 120, 180, 240, 300 and 360° · s^{-1}] × time [first week testing

or second week testing]) repeated measures analysis of variance (ANOVA). Sphericity was verified for all data being compared by the Mauchly test. The Multivariate Test box (Wilk's Lambada value) was studied for three-way interactions, then two-way interactions and then main effects (P<0.05). The Pairwise Comparisons post-hoc test was used to determine exactly where the significant findings occurred when there were more than two conditions (setup position and speed). Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.0167 for setup position and P<0.008 for speed.

Relative reliability was assessed by calculating the $ICC_{(2,1)}$. From the ICC value the SEM was calculated, which represented absolute reliability (Section 3.2.3.4, paragraph 2).

7.4.4 Results

7.4.4.1 Peak Torque

The peak torque relative reliability results for ankle inversion (Table 7.3) and ankle eversion (Table 7.4) during the three setup positions show that the 10° plantarflexed position was the most reliable setup. It can be seen that at speeds 60 through 360° · s^{-1} excellent to high relative reliability (ICC's > 0.70) results were found. However, Currier (1990) suggested that an ICC value > 0.80 was acceptable for clinical work, and therefore only speeds 60 through 240° · s^{-1} would be adequate. When observing the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position only speeds

60 through 180° · s^{-1} produced ICC's above the acceptable level (>0.80) for both ankle inversion and ankle eversion.

When observing the SEM results for ankle inversion (Table 7.3) and ankle eversion (Table 7.4) the results show that the 10° plantarflexion position has the lowest, and therefore the least variable results. It has been stated that SEM values below 10% are an acceptable level of variance. The 10° plantarflexion position for speeds 60 through 360° · s^{-1} all had SEM values below 10% (range: 2.56-9.90%) and therefore all fall within the recommended level of variance. When observing the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position only speeds 60 through 240° · s^{-1} produced SEM values below the recommended 10% threshold, for both ankle inversion and ankle eversion. The results also highlight that during the three setup positions, peak torque relative and absolute reliability decreased with each increase in velocity (Table 7.3 and 7.4). It can therefore be seen from the ICC and SEM results, that the 10° plantarflexion was the most reliable setup position.

Results from the repeated measures ANOVA showed significantly greater (P<0.0167) peak torque values for the 10° plantarflexion position when compared to the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position at speeds 60 through 360° · s^{-1} for both ankle inversion (Figure 7.4) and ankle eversion (Figure 7.5). The results therefore show that the 10° plantarflexion position enabled the greatest peak torque values to be produced. The repeated measures ANOVA also showed a significant decrease (P<0.008) in normalised peak torque values with each increase in velocity (Table 7.3 and 7.4).

Table 7.3. Normalised Peak Torque Values for Ankle Inversion During the Three Setup Positions.

VELOCITY (° · s^{-1})	10° DORSIFLEXION				NEUTRAL DORSIFLEXION/PLANTARFLEXION				10° PLANTARFLEXION			
	TEST 1 (Nm·Kg^{-1})	TEST 2 (Nm·Kg^{-1})	ICC	SEM (%)	TEST 1 (Nm·Kg^{-1})	TEST 2 (Nm·Kg^{-1})	ICC	SEM (%)	TEST 1 (Nm·Kg^{-1})	TEST 2 (Nm·Kg^{-1})	ICC	SEM (%)
60	0.32 (0.03)	0.33 (0.04)	0.90	3.03	0.33 (0.04)	0.33 (0.04)	0.91	3.03	0.39 (0.04)	0.40 (0.03)	0.94	2.56
120	0.28 (0.04)*	0.28 (0.03)*	0.86	3.57	0.30 (0.03)*	0.29 (0.03)*	0.89	3.45	0.33 (0.03)*	0.32 (0.04)*	0.92	3.13
180	0.24 (0.03)*	0.25 (0.03)*	0.80	4.00	0.25 (0.03)*	0.26 (0.03)*	0.84	4.00	0.28 (0.03)*	0.27 (0.03)*	0.90	3.70
240	0.18 (0.03)*	0.19 (0.02)*	0.71	5.26	0.19 (0.02)*	0.18 (0.03)*	0.75	5.55	0.20 (0.04)*	0.21 (0.03)*	0.84	5.00
300	0.13 (0.02)*	0.11 (0.02)*	0.68	10.33	0.12 (0.03)*	0.14 (0.02)*	0.70	10.69	0.17 (0.03)*	0.15 (0.02)*	0.73	6.67
360	0.07 (0.02)*	0.07 (0.01)*	0.65	14.29	0.08 (0.01)*	0.09 (0.02)*	0.66	11.11	0.12 (0.02)*	0.10 (0.02)*	0.70	8.33

Data are presented as Mean (SD). * Significantly different (P<0.008) from previous velocity

Table 7.4. Normalised Peak Torque Values for Ankle Eversion During the Three Setup Positions.

VELOCITY ($° \cdot s^{-1}$)	10° DORSIFLEXION				NEUTRAL DORSIFLEXION/PLANTARFLEXION				10° PLANTARFLEXION			
	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)
60	0.31 (0.03)	0.29 (0.03)	0.88	3.23	0.30 (0.03)	0.29 (0.03)	0.89	3.33	0.35 (0.03)	0.36 (0.03)	0.95	2.78
120	0.23 (0.03)*	0.25 (0.03)*	0.83	4.00	0.25 (0.03)*	0.25 (0.04)*	0.83	4.00	0.30 (0.04)*	0.29 (0.03)*	0.92	3.33
180	0.20 (0.03)*	0.19 (0.03)*	0.80	5.00	0.21 (0.03)*	0.20 (0.03)*	0.81	4.76	0.25 (0.03)*	0.26 (0.03)*	0.89	3.85
240	0.11 (0.02)*	0.10 (0.02)*	0.70	9.09	0.11 (0.02)*	0.12 (0.03)*	0.73	8.33	0.17 (0.03)*	0.17 (0.03)*	0.88	5.88
300	0.08 (0.01)*	0.09 (0.02)*	0.66	11.11	0.08 (0.02)*	0.08 (0.02)*	0.67	12.50	0.12 (0.03)*	0.11 (0.03)*	0.72	8.33
360	0.06 (0.01)*	0.05 (0.01)*	0.61	16.67	0.05 (0.01)*	0.06 (0.02)*	0.65	16.70	0.08 (0.02)*	0.10 (0.02)*	0.71	9.90

Data are presented as Mean (SD). * Significantly different (P<0.008) from previous velocity

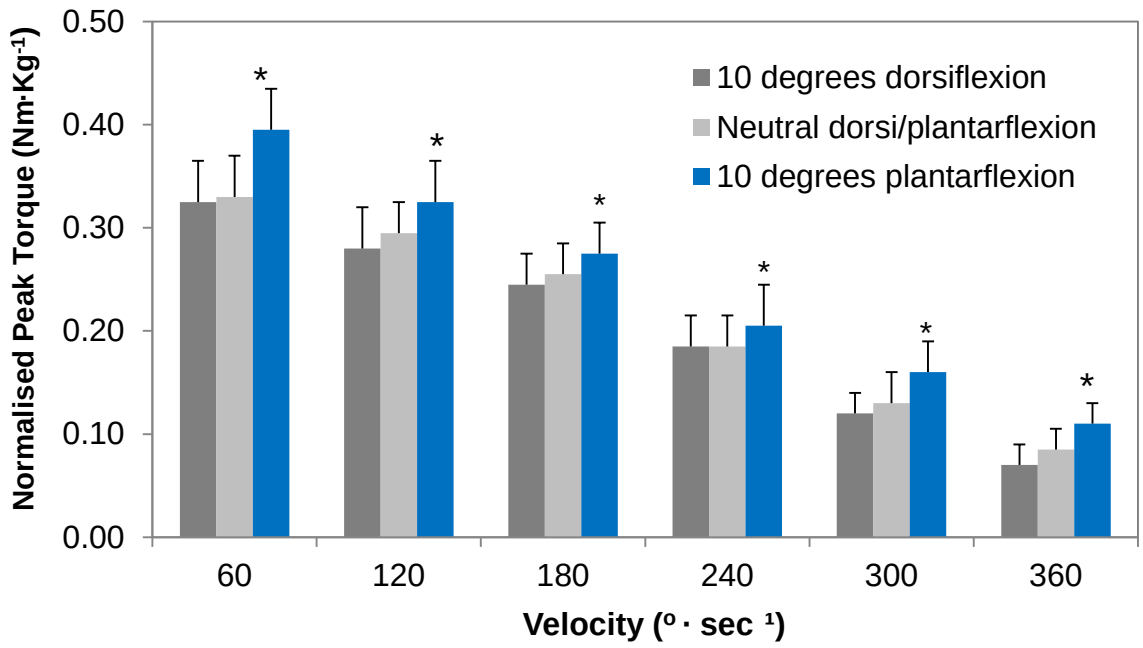


Figure 7.4. Normalised Peak Torque for Ankle Inversion During the Three Setup Positions Across a Velocity Spectrum (Mean ± SD). *10 degrees plantarflexion position significantly (*P*<0.0167) different to other positions.

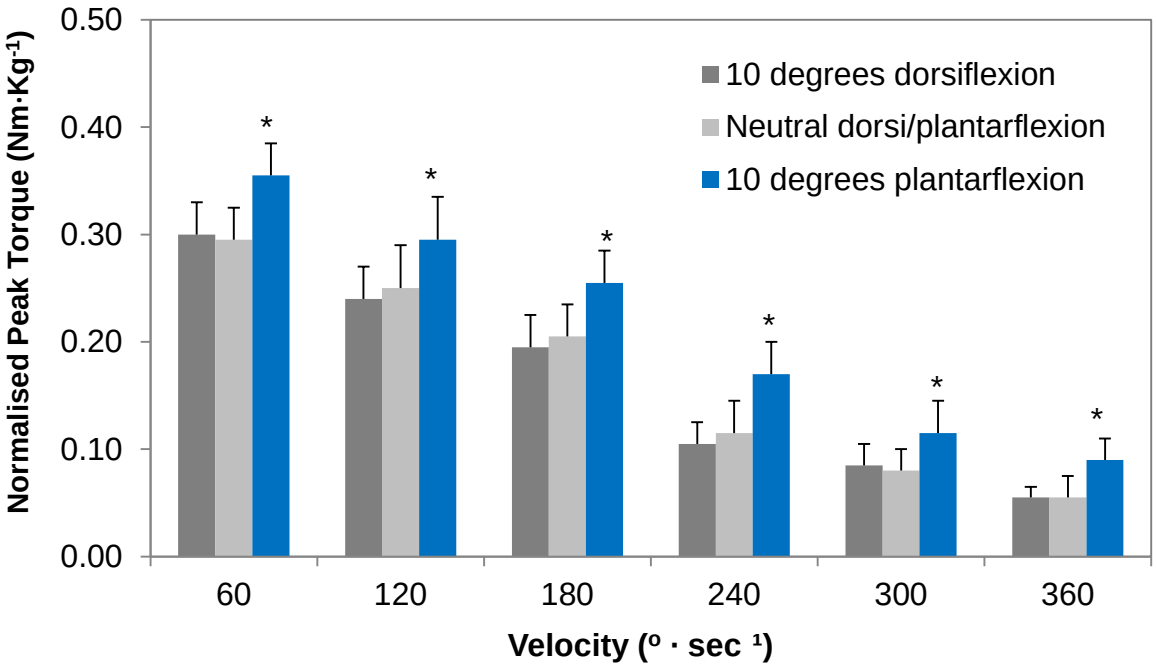


Figure 7.5. Normalised Peak Torque for Ankle Eversion During the Three Setup Positions Across a Velocity Spectrum (Mean ± SD). *10 degrees plantarflexion position significantly (*P*<0.0167) different to other positions.

7.4.4.2 Total Work

The total work relative reliability results for ankle inversion (Table 7.5) and ankle eversion (Table 7.6) during the three setup positions show that the 10° plantarflexed position was the most reliable setup. It can be seen that at speeds 60 through 360° · s^{-1} excellent to high reliability (ICC's > 0.70) results were found. However, ICC values > 0.80 have been suggested as an acceptable level for clinical work (Currier, 1990), and therefore only speeds 60 through 240° · s^{-1} would be satisfactory. When observing the ankle inversion results (Table 7.5) for the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position only speeds 60 and 120° · s^{-1} produced ICC's above the acceptable level (>0.80). When examining ankle eversion results (Table 7.6) for the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position only speeds 60 through 180° · s^{-1} produced ICC's above the acceptable level (>0.80).

When observing the SEM results for ankle inversion (Table 7.5) and ankle eversion (Table 7.6) the results show that the 10° plantarflexion position had the least variable results. As previously mentioned, 10% has been stated as an acceptable SEM value. The 10° plantarflexion position for speeds 60 through 300° · s^{-1} all had SEM values below 10% (range: 2.00-9.90%) and therefore all fall within the recommended level of variance. When observing the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position only speeds 60 through 240° · s^{-1} produced SEM values below the recommended 10% threshold, for both ankle inversion and ankle eversion. The results also highlight that during the three setup positions, total work relative and absolute reliability decreased with each increase in velocity (Table 7.5 and 7.6). It can therefore

be seen from the ICC and SEM results, that the 10° plantarflexion was the most reliable setup position.

Results from the repeated measures ANOVA showed significantly greater (P<0.0167) total work values for the 10° plantarflexion position when compared to the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position at speeds 60 through 360° · s^{-1} for both ankle inversion (Figure 7.6) and ankle eversion (Figure 7.7). The results therefore show that the 10° plantarflexion position enabled the greatest total work values to be produced. The repeated measures ANOVA also showed a significant decrease (P<0.008) in normalised total work values with each increase in velocity (Table 7.5 and 7.6).

Table 7.5. Normalised Total Work Values for Ankle Inversion During the Three Setup Positions.

VELOCITY ($° \cdot s^{-1}$)	10° DORSIFLEXION				NEUTRAL DORSIFLEXION/PLANTARFLEXION				10° PLANTARFLEXION			
	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)
60	0.17 (0.02)	0.17 (0.02)	0.89	3.53	0.16 (0.01)	0.17 (0.02)	0.88	4.08	0.20 (0.02)	0.21 (0.02)	0.95	2.13
120	0.12 (0.02)*	0.13 (0.02)*	0.85	5.38	0.12 (0.01)*	0.12 (0.01)*	0.84	5.21	0.16 (0.01)*	0.16 (0.02)*	0.92	3.54
180	0.09 (0.02)*	0.09 (0.01)*	0.79	9.15	0.09 (0.01)*	0.08 (0.01)*	0.78	4.97	0.12 (0.01)*	0.11 (0.01)*	0.91	2.50
240	0.06 (0.01)*	0.05 (0.01)*	0.72	8.82	0.05 (0.01)*	0.06 (0.01)*	0.73	8.65	0.08 (0.01)*	0.09 (0.01)*	0.86	4.16
300	0.02 (0.01)*	0.03 (0.01)*	0.66	19.44	0.02 (0.01)*	0.02 (0.01)*	0.69	18.4	0.05 (0.01)*	0.06 (0.01)*	0.74	8.49
360	0.01 (0.01)*	0.01 (0.01)*	0.64	23.6	0.01 (0.01)*	0.01 (0.01)*	0.65	22.8	0.02 (0.01)*	0.02 (0.01)*	0.71	18.45

Data are presented as mean (SD). * Significantly different (P<0.008) from previous velocity

Table 7.6. Normalised Total Work Values for Ankle Eversion During the Three Setup Positions.

VELOCITY ($° \cdot s^{-1}$)	10° DORSIFLEXION				NEUTRAL DORSIFLEXION/PLANTARFLEXION				10° PLANTARFLEXION			
	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)
60	0.15 (0.01)	0.16 (0.02)	0.91	3.56	0.16 (0.02)	0.16 (0.01)	0.90	3.20	0.18 (0.02)	0.19 (0.02)	0.93	2.78
120	0.10 (0.01)*	0.11 (0.02)*	0.86	4.98	0.11 (0.01)*	0.11 (0.01)*	0.86	4.90	0.15 (0.01)*	0.14 (0.01)*	0.91	2.00
180	0.08 (0.01)*	0.09 (0.02)*	0.81	8.76	0.08 (0.01)*	0.08 (0.01)*	0.82	8.25	0.11 (0.01)*	0.11 (0.01)*	0.90	.4.34
240	0.04 (0.01)*	0.05 (0.01)*	0.73	9.75	0.06 (0.01)*	0.04 (0.01)*	0.74	9.69	0.07 (0.01)*	0.08 (0.01)*	0.84	6.54
300	0.02 (0.01)*	0.02 (0.01)*	0.67	18.35	0.02 (0.01)*	0.02 (0.01)*	0.69	17.98	0.04 (0.01)*	0.04 (0.01)*	0.73	9.90
360	0.01 (0.01)*	0.01 (0.01)*	0.66	22.25	0.01 (0.01)*	0.01 (0.01)*	0.65	23.05	0.02 (0.01)*	0.01 (0.01)*	0.70	17.53

Data are presented as mean (SD). * Significantly different (P<0.008) from previous velocity

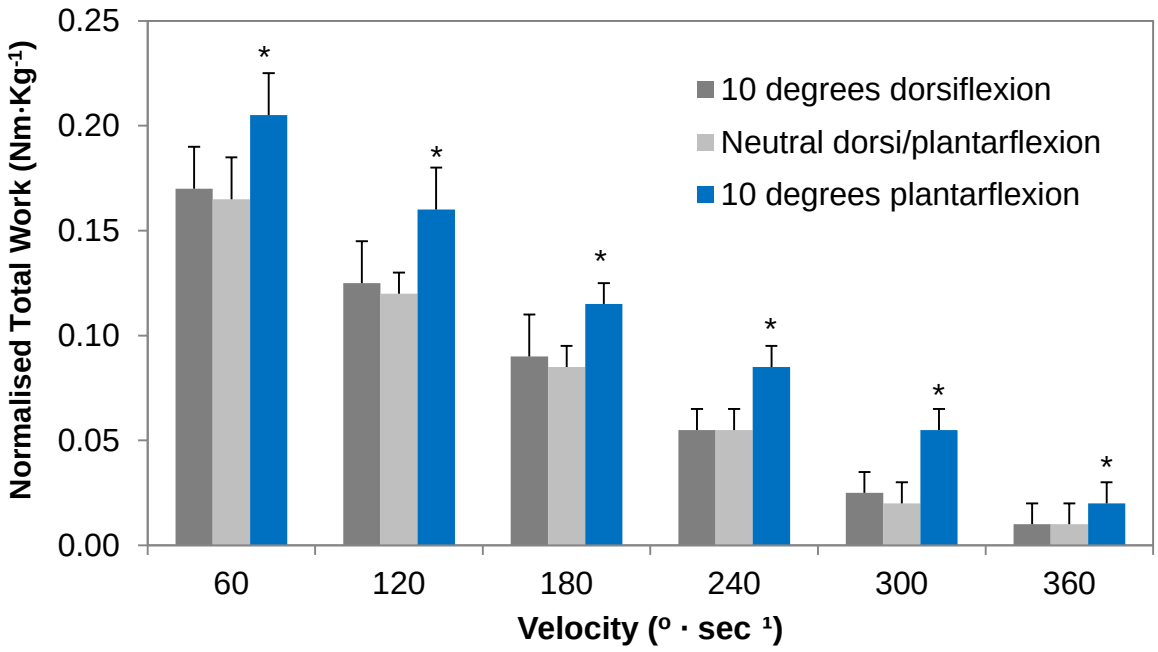


Figure 7.6. Normalised Total Work for Ankle Inversion During the Three Setup Positions Across a Velocity Spectrum (Mean ± SD). *10 degrees plantarflexion position significantly (*P*<0.0167) different to other positions

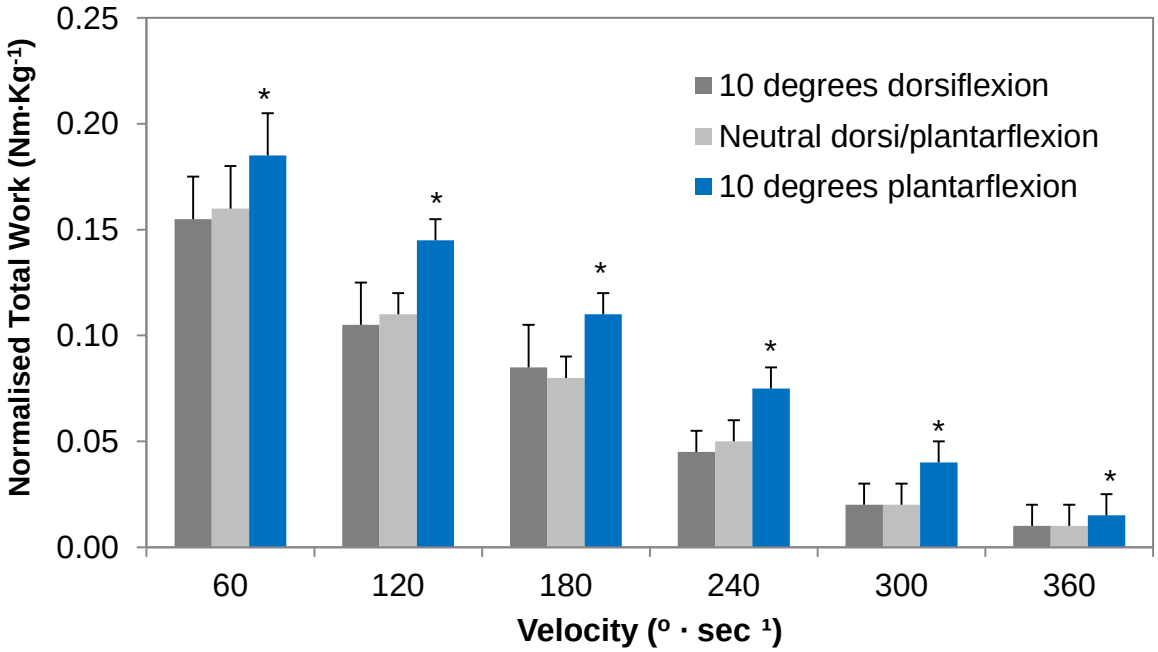


Figure 7.7. Normalised Total Work for Ankle Eversion During the Three Setup Positions Across a Velocity Spectrum (Mean ± SD). *10 degrees plantarflexion position significantly (*P*<0.0167) different to other positions

7.4.5 Discussion

The primary aim of this study was to compare the relative and absolute reliability of three setup positions (10° dorsiflexion, neutral dorsiflexion/plantarflexion, and 10° plantarflexion) during isokinetic ankle inversion-eversion exercise. The results highlighted that the 10° plantarflexion position was the most reliable setup position, with the highest ICC results and the lowest SEM variance during ankle inversion and ankle eversion. The secondary aim of this study was to investigate the effect of setup position on the magnitude of peak torque and total work. The results showed significantly greater peak torque and total work values for the 10° plantarflexion position when compared to the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position at speeds 60 through 360° $\cdot$ s^{-1} for both ankle inversion and ankle eversion. Therefore, the hypothesis that the 10° plantarflexion position will produce the highest reliability and the greatest peak torque and total work values can be formally accepted.

7.4.5.1 Peak Torque and Total Work Reliability

The peak torque and total work relative reliability results for ankle inversion and ankle eversion during the three setup positions show that the 10° plantarflexed position was the most reliable setup. Speeds 60 through 240° $\cdot$ s^{-1} demonstrated ICC values > 0.80, which has been suggested as acceptable for clinical work (Currier, 1990). Several other authors have found reliable results for inversion-eversion testing on the isokinetic dynamometer (Aydog et al., 2004; Cawthorn et al., 1991). Cawthorn et al. (1991) found reliable results (ICC's ranging: 0.87-0.94) for 10° plantarflexion, neutral dorsiflexion/plantarflexion and 10° dorsiflexion, however, they only tested at the speed

of 160° · s^{-1}. Aydog et al. (2004) also found high reliability results (ICC's ranging: 0.87-0.96) for ankle inversion-eversion, when testing at the speeds of 60 and 180° · s^{-1}. These results are in agreement with the results from the present study, however, the current study is the only one obtainable that has tested through a velocity spectrum ranging from 60 to 360° · s^{-1}.

The present study found that with each increase in velocity, there was a decrease in reliability (Tables 7.3 to 7.6). This is possibly due to the subjects finding it more difficult to obtain the higher velocities, as shown by the inverse relationship between load range and increased velocity (Gautrey et al., 2013b). As the velocity of the dynamometer increases, the subject finds it more difficult to achieve this velocity, as a result of this the peak torque and total work values become more variable, and therefore lower reliability values are produced. The slower speeds of 60 through 240° · s^{-1} all demonstrate ICC's above 0.80 which is clinically acceptable (Currier, 1990). Speeds 300 and 360° · s^{-1} still show high reliability, but the values are below the clinically acceptable level of 0.80. These results may suggest that researchers, clinicians and sports injury professionals should opt for a speed between 60 and 240° · s^{-1} if they are conducting repeated tests, and require a reliable protocol.

The accuracy to which these protocols are reproducible is also a critical factor as determined by the SEM. Although high reliability coefficients (such as ICC's) have been previously reported for isokinetic strength, SEM values have received little attention in the literature. The SEM value in this study was expressed as a percentage in order to allow clinical usage of these measures. As demonstrated by the results of the current study, re-test values for peak torque for the 10° plantarflexion position ranged from

2.56-9.90% to the initial test for speeds 60 through 360° · s^{-1}, and for total work ranged from 2.00-9.90% to the initial test for speeds 60 through 300° · s^{-1}. It should therefore, seem appropriate in future studies to attribute differences in isokinetic results to intervention, training improvements or injury, should they exceed the SEM values outlined in tables 7.3 to 7.6.

The relevance of the reliability findings in the present study lies predominantly in the research domain. It may be argued that the increase in reliability is marginal between the three setup positions. For example, when observing the ankle eversion peak torque results at 60° · s^{-1} (Table 7.4) the relative reliability improved from 0.88 and 0.89 with the 10° dorsiflexion position and the neutral dorsiflexion/plantar flexion position, respectively, to 0.95 with the 10° plantarflexion position. The SEM variance was also improved from 3.23% and 3.33% with the 10° dorsiflexion position and the neutral dorsiflexion/plantar flexion position, respectively, to 2.78% with the 10° plantarflexion position. These changes may seem small, but in the field of research where reliable protocols are a necessity, the 10° plantarflexion position improved the reliability of the protocol.

Another important difference between the setup positions, is that the 10° plantarflexion positions peak torque and total work results show clinically reliable measures (>0.80) for speeds 60 through 240° · s^{-1} for both ankle inversion and eversion. However, the 10° dorsiflexion position and neutral dorsiflexion/plantarflexion position only show clinically reliable peak torque results for speeds 60 through 180° · s^{-1} for ankle inversion and eversion. For the total work results the 10° dorsiflexion position and neutral dorsiflexion/plantarflexion position only show reliable total work results for speeds 60

and 120° · s^{-1} for inversion, and 60 through 180° · s^{-1} for eversion. If researchers, clinicians or sports injury professionals wish to test patients at velocities between 180 and 240° · s^{-1}, they should opt for the 10° plantarflexion position as this notably increased the reliability in comparison to the 10° dorsiflexion position and neutral dorsiflexion/plantarflexion position.

7.4.5.2 Peak Torque and Total Work Magnitude

The results of the present study showed significantly greater peak torque and total work values for the 10° plantarflexion position when compared to the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position at speeds 60 through 360° · s^{-1} for both ankle inversion and ankle eversion. These results are consistent with the findings of Cawthorn et al. (1991) who measured peak torque on a MERAC testing table in 10° dorsiflexion, neutral dorsiflexion/plantarflexion, and 10° plantarflexion. Cawthorn et al. (1991) demonstrated a 17.8% difference in peak torque of the foot invertors from the strongest to the weakest position, and a 16.5% difference in peak torque of the foot evertors from the strongest to weakest position. Cawthorn et al. (1991) concluded that the 10° plantarflexion position was better than the other positions tested as torque output was greatest at this position. The results of the present study are consistent with those of Cawthorn et al. (1991) as the 10° plantarflexion position also resulted in the highest peak torque and total work outputs.

Even though not specific to the ankle joint, Walmsley and Szybbo (1987) demonstrated significant differences in mean peak torque measurements with varied testing positions for the shoulder. They found differences of 12% from the strongest to the weakest

position for shoulder internal rotators, and 12.7% difference for the shoulder external rotators when testing at 180° · s^{-1}. Walmsley and Szybbo (1987) reported that the differences were greater when testing at 120° · s^{-1} and greater still when testing at 60° · s^{-1}. Even though not specific to the ankle joint, these results are consistent with the results of the present study which found a significant increase in peak torque and total work results with the 10° plantarflexion position, in comparison to the neutral dorsiflexion/plantarflexion position and the 10° dorsiflexion position.

The present study found that the 10° plantarflexion position produced the highest mean peak torque and total work values; this was followed by the neutral dorsiflexion/plantar flexion position, and then the 10° dorsiflexion position. These findings are consistent with Cawthorn et al. (1991) who also found that the plantarflexed position gave the greatest mean peak torque values and the dorsiflexed position gave the lowest. One explanation for these results involves physiological considerations.

The posterior tibialis, a primary foot invertor, and the peroneus longus and brevis, primary foot evertors, all have tendons that pass around the malleoli to their respective distal attachments. As the foot is increasingly dorsiflexed, these muscles are lengthened due to the arrangement of the tendons around the malleoli. A muscle contracts as a result of overlapping actin and myosin filaments. It contracts most strongly when the muscle length allows the filaments to overlap most efficiently. This length is termed the optimal length and is the length at which the greatest tension of that muscle is developed. This optimum length is normally found near the mid range of the muscle. Lengthening the muscle past the optimal length causes a drop in the amount of tension that the muscle can develop (Cawthorn et al., 1991).

The muscles that were tested in the present study are required to function between approximately 10° dorsiflexion and 25° plantarflexion during gait (Cawthorn et al., 1991; Grey & Basmajian, 1968). The position of 10° plantarflexion is closest to the midrange of these muscles than the other two testing positions. If 10° plantarflexion allows more efficient overlapping of the actin and myosin filaments, then the force generated by the muscles will be greatest at this position, as shown by the results of the present study.

7.4.5.3 Clinical Implications

The 10° plantarflexion position appears to be preferable to positions of 10° dorsiflexion or neutral, as this position produced the highest reliability and greatest peak torque and total work values. Therefore clinicians should opt for the 10 ° plantarflexion setup over other setup positions. Results indicating that the plantarflexed position is the strongest position for ankle invertors and evertors may have implications for therapeutic and testing protocols. For example, facilitation of very weak invertors or evertors might be better accomplished with the ankle in the plantarflexed position.

7.4.5.4 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. It must be remembered that the results are only applicable if the same equipment and protocol is used as in the current study. Future studies may wish to repeat this study but using different makes of isokinetic dynamometers and varied protocols.

7.4.6 Conclusion

In summary, the results indicate that the 10° plantarflexion position led to an increase in reliability and magnitude of peak torque and total work measurements. Marginally higher ICC results and lower SEM variance were found at speeds 60 though 240° · s^{-1}. The changes may appear small, but in the field of research where reliable protocols are a necessity, the 10° plantarflexion position improved the reliability of the protocol. The results suggest that researchers, clinicians and sports injury professionals should opt for the 10° plantarflexion position at speeds between 60 and 240° · s^{-1} if they are conducting repeated tests, and require a reliable protocol.

7.5 Development of Research

Pilot Study Five addressed the issue of identifying the most reliable ankle setup position to be used in Study Three and Four. The results indicated that the 10° plantarflexion position was the most reliable setup, and also enabled the subject to produce significantly higher peak torque values. Therefore, this setup position was deemed appropriate for Study Three and Four.

7.6 Pilot Study Six: The Effect of Velocity on Load Range During Isokinetic Ankle Inversion-Eversion Exercise

7.6.1 Abstract

Aim: To quantify the components of acceleration, load range and deceleration through a velocity spectrum during concentric ankle inversion and eversion isokinetic exercise, and to investigate the effect of load range on peak torque and work done. **Method:** Sixteen male healthy subjects performed three maximal concentric reciprocal ankle inversion and eversion gravity corrected repetitions in a fixed order at 60, 120, 180, 240, 300 and 360° · s^{-1}, with a 30 second rest between velocities. **Results:** Inversion and eversion results revealed that load range significantly decreased while acceleration and deceleration ROM significantly increased (P<0.05) with each increase in velocity. When the total peak torque data was corrected for load range there was a significant decrease (P<0.05) in peak torque at velocities of 240° · s^{-1} and above, for inversion and eversion. Load range correction also resulted in a significant (P<0.05) decrease in work done at velocities of 180° · s^{-1} and above for inversion, and 240° · s^{-1} and above for eversion. **Conclusion:** The results demonstrate an inverse relationship between isokinetic velocity and load range during concentric ankle inversion and eversion, and suggest a need for the clinician to carefully consider velocity selection when performing isokinetic tests.

7.6.2 Introduction

Isokinetic dynamometry has been commonly used to test the strength of the invertors and evertors of the ankle, in both healthy and injured populations (Amaral De Noronha & Borges Junior, 2004; Aydog et al., 2004; Bernier et al., 1997; Kaminski & Dover, 2001; Kaminski et al., 1999; Kaminski et al., 1995; Wilkerson et al., 1997; Willems et al., 2002). The strength of the evertors has been a popular area of research in relation to individuals with a history of FAI. The evertor muscles are often described as playing a major role in the prevention of ligamentous injuries (Willems et al., 2002). The strength of the evertors, specifically peroneus longus and peroneus brevis, have been suggested to provide support to the lateral ligaments (Osternig, 1986) and resist sudden inversion during a lateral ankle sprain (Willems et al., 2002). While some studies have reported a decrease in the strength of the ankle evertors after inversion sprain when tested manually (Bosien et al., 1955; Staples, 1975; Staples, 1972) or isokinetically (Tropp, 1986), others have reported no decrease in strength as compared with the uninjured ankle when tested isokinetically (Lentell et al., 1990). It has been stated that the evertors of the ankle should be evaluated in healthy participants, to try and identify individuals with a possible pre-disposition to ankle sprains (Beckman & Buchanan, 1995).

The isokinetic dynamometer has been frequently used for rehabilitation or training purposes (Brown & Whitehurst, 2003; Hamdoun-Kahlaoui, Lebib, Miri, Ghorbel, Koubaa & Rahali-Khachlouf, 2010; Hammami, Coroian, Julia, Amri, Mottet, Herisson & Laffont, 2012; Murray, Brown, Zinder, Noffal, Bera, & Garrett, 2007; Nickols-Richardson, Miller, Wooten, Ramp & Herbert, 2007; Osternig, 2000). As the isokinetic dynamometer only

offers resistance once the pre-set velocity is attained, any strength gains achieved from isokinetic exercise may be proportional to the total amount of range of motion (ROM) actually sustained at the pre-set isokinetic velocity (Brown, Whitehurst, Gilbert et al., 1995). It is therefore of great interest to investigate what percentage of the ROM of a concentric action is actually spent at the pre-selected velocity, over a velocity spectrum.

A concentric action performed on an isokinetic device involves three main components: acceleration, sustained velocity, and deceleration (Brown, Whitehurst, Gilbert et al., 1995; Osternig, 1975; Taylor et al., 1991). The acceleration component has been defined as the individual's ability to "catch" the dynamometer (Davies, 1992; Glick et al., 1976). The "catch" phase is completed once the individual attains the pre-set velocity, and the resistance is met, which then prevents any further acceleration (Davies, 1992; Glick et al., 1976). The sustained velocity component of the repetition has also been termed load range (Brown, Whitehurst, Gilbert et al., 1995; Findley et al., 2006; Kurdak et al., 2005). To be more precise the concept of load range has been described as external machine resistance encountered through a pre-set sustained velocity within a defined ROM (Brown, Whitehurst, Gilbert et al., 1995). The final component, mechanical deceleration (Brown, Whitehurst, Gilbert et al., 1995), offers resistance while the isokinetic dynamometer decreases speed at the end of the defined ROM. However, Brown, Whitehurst, Gilbert et al. (1995) has argued that this phase is neither directly governed by the tester nor quantifiable as torque produced under controlled isokinetic conditions, and therefore ceases to be isokinetic (Brown, Whitehurst, Gilbert et al., 1995).

Previous research has shown that torque patterns are significantly affected when the load range phase of the motion is taken into consideration (Brown & Whitehurst, 2000; Brown, Whitehurst, Gilbert et al., 1995; Kovaleski et al., 1995). In short, this means that actual torque may differ by a large magnitude if evaluated outside of the load range (Findley et al., 2006). Kurdak et al. (2005) found a significant decrease when comparing load range peak torque to total peak torque at speeds above $270° \cdot s^{-1}$ for knee extension and above $300° \cdot s^{-1}$ for knee flexion. Kurdak et al. (2005) also found a significant decrease when comparing load range work and total work at speeds above $90° \cdot s^{-1}$ for both knee extension and knee flexion. Gautrey et al. (2013b) also found that when the total peak torque data was corrected for load range there was a significant decrease in peak torque at velocities of $300° \cdot s^{-1}$ and above, for both hip abduction and hip adduction. Load range correction also resulted in a significant decrease in work done at velocities of $120° \cdot s^{-1}$ and above, for both hip abduction and hip adduction. These results highlight the importance of correcting the data for load range as it is apparent that large errors can occur if this process is not undertaken.

Increased angular velocity results in a reduction in load range, thus data from the measurements that were performed at higher angular velocities may not actually reflect load range values (Gautrey et al., 2013b; Kurdak et al., 2005). This is in agreement with the classic force – velocity curve, which explains the relationship between skeletal muscle contraction velocity and torque production (Widrick, Trappe, Costill & Fitts, 1996): as velocity increases, torque decreases (Brown & Whitehurst, 2000). Therefore extra caution is required to make correct interpretation of isokinetic results (Brown & Whitehurst, 2000).

Load range has been investigated previously, however, only during unilateral knee flexion/extension (Brown, Whitehurst, Gilbert et al., 1995; Osternig, 1986; Taylor et al., 1991; Wilk, Romaniello, Soscia, Arrigo & Andrews, 1994), bilateral knee flexion/extension (Scibelli, Brown, Whitehurst, Bryant & Buchalter, 1993), shoulder external/internal rotation (Brown, Whitehurst, Findley, Gilbert & Buchalter, 1995) and hip abduction/adduction (Gautrey et al., 2013b). Each study found an inverse relationship between load range and velocity, yet the primary focus of these studies was load range, apart from Brown, Whitehurst, Gilbert et al. (1995) and Gautrey et al. (2013b) who also considered the impact of the acceleration and deceleration components. The quantification of each component may lead to a more complete understanding of load range magnitude and position within the exercised ROM. This information may better equip the clinician in more accurate velocity prescription during isokinetic exercise. From the findings of previous literature it can be hypothesised that with each increase in velocity there will be a decrease in the load range component, and an increase in the acceleration and deceleration components. It was also hypothesised that load range corrected peak torque and total work data will be significantly different to the uncorrected data at higher velocities.

The primary aim of this study was therefore to quantify the components of load range, acceleration, and deceleration through a velocity spectrum during concentric ankle inversion and eversion isokinetic exercise. The secondary aim of this study was to investigate the effect of load range on peak torque and work done.

7.6.3 Method

7.6.3.1 Subjects

The same subjects were used as in Pilot Study Five (Section 7.4.3.1).

7.6.3.2 Experimental Design

The same experimental design was used as in Pilot Study Five (Section 7.4.3.2); apart from the footplate was positioned in the 10° plantarflexion position, and subjects were not required to return to the laboratory seven days later to repeat the procedure.

7.6.3.3 Data Analysis

Data was collected via the Biodex Advantage Software (version 4.5, Biodex Medical Systems, Shirley, New York), which allowed the separation of each contraction into its component parts for individual analysis. The same definitions as stated by Brown, Whitehurst, Gilbert et al. (1995) were used; the range of motion prior to velocity attainment was termed acceleration, while ROM after load range was termed deceleration (Figures 7.8 and 7.9). Load range was determined for ankle inversion and eversion by subtracting the sum of acceleration ROM and deceleration ROM from the total test ROM using the available cursors on the screen (Brown, Sjostrom, Comeau, Whitehurst, Greenwood & Findley, 2005; Brown & Whitehurst, 2003; Brown, Whitehurst & Findley, 2005; Kovaleski et al., 1995; Wilk et al., 1994). Taylor et al. (1991) stated that velocity overshoot was measured at 3.5% on the Biodex dynamometer, this is not

reflected in the velocity tracings but was included in the load range component. Brown, Whitehurst, Gilbert et al. (1995) recommended using 100% of the pre-selected velocity because there is no machine-offered resistance below full velocity attainment (Brown, Whitehurst, Findley et al., 1995; Osternig, 1975; Osternig, 1986; Osternig, Sawhill, Bates & Hamill, 1983; Scibelli et al., 1993; Taylor et al., 1991; Wilk et al., 1994).

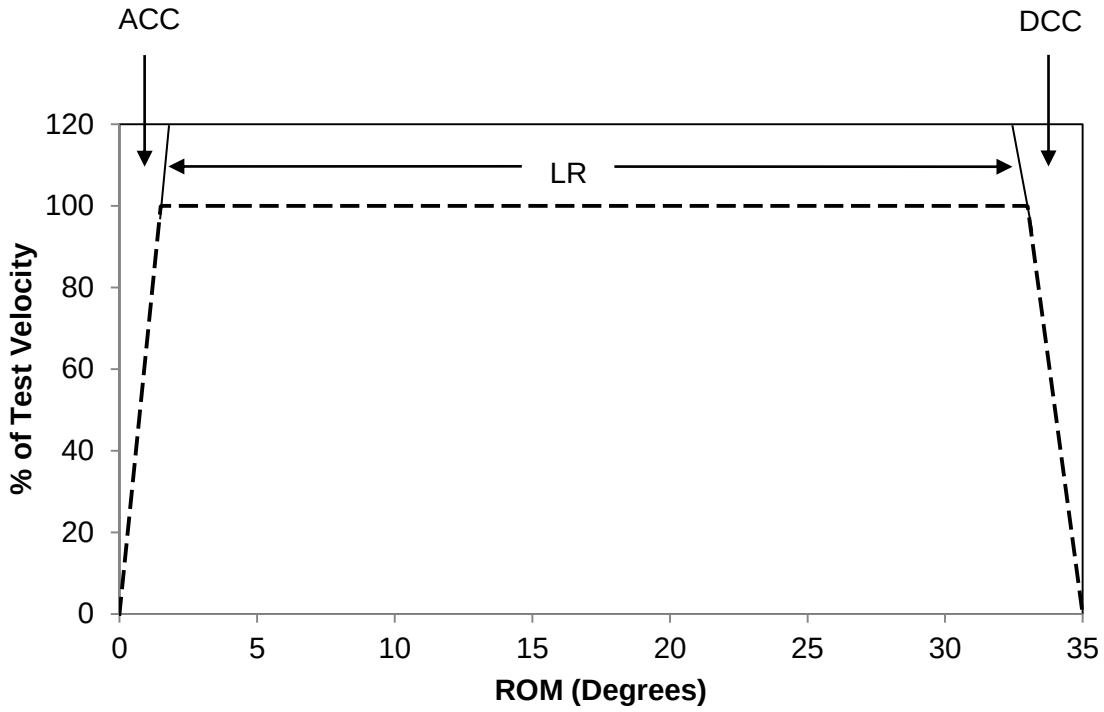


Figure 7.8. Example of a Mean Velocity Tracing at 60° · s^{-1} Showing Acceleration (ACC), Load Range (LR), and Deceleration (DCC) Range of Motion (ROM) During Ankle Eversion.

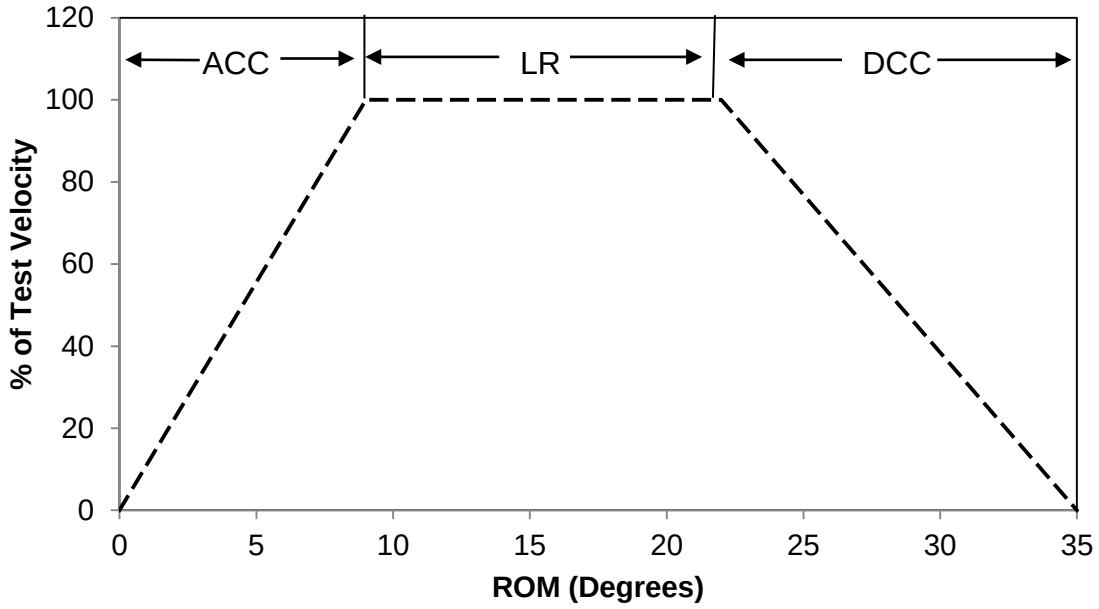


Figure 7.9. Example of a Mean Velocity Tracing at 360° · s^{-1} Showing Acceleration (ACC), Load Range (LR), and Deceleration (DCC) Range of Motion (ROM) During Ankle Eversion.

Following the determination of load range ROM, the total peak torque, load range peak torque, total work and load range work were calculated for both ankle inversion and ankle eversion across all velocities. Total peak torque was determined by locating the highest point of the curve. The load range peak torque was determined by locating the highest point of the curve within the load range ROM. Total work done was determined by calculating the area under the curve. The load range work done was determined by calculating the area under the curve within the load range ROM. All torque data was normalised with respect to the subject's body weight (Kurdak at al., 2005).

7.6.3.4 Statistical Analysis

Using SPSS (version 19) a 6 x 2 (speed [60, 120, 180, 240, 300 and 360° · s^{-1}] x movement [ankle inversion and ankle eversion]) repeated measures analysis of variance (ANOVA) was performed for the acceleration, load range and deceleration data. Sphericity was verified for all data being compared by the Mauchly test. The Multivariate Test box (Wilk's Lambada value) was studied for two-way interactions and then main effects (P<0.05). The Pairwise Comparisons post-hoc test was used to determine exactly where the significant findings occurred when there were more than two conditions (speed). Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.008.

A 6 x 2 x 2 (speed [60, 120, 180, 240, 300 and 360° · s^{-1}] x analysis type [total values and load range values] x movement [ankle inversion and ankle eversion]) repeated measures ANOVA was performed for peak torque and work data. Sphericity was verified for all data being compared by the Mauchly test. The Multivariate Test box (Wilk's Lambada value) was studied for three-way interactions, then two-way interactions and then main effects (P<0.05). The Pairwise Comparisons post-hoc test was used to determine exactly where the significant findings occurred when there were more than two conditions (speed). Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.008.

7.6.4 Results

The 6 x 2 (speed x movement) repeated measures ANOVA results revealed that load range significantly decreased while acceleration and deceleration ROM significantly increased with each increase in velocity, for both ankle inversion and ankle eversion (Table 7.7). There was no significant difference found between inversion and eversion results. The amount of ROM spent in load range significantly decreased from 31.9˚ to 16.0˚ for inversion, and from 31.6˚ to 15.4˚ for eversion, at 60 through 360° · s^{-1}. The amount of ROM spent in acceleration significantly increased from 1.1˚ to 7.9˚ for inversion, and from 1.4˚ to 8.9˚ for eversion, at 60 through 360° · s^{-1}. The amount of ROM spent in deceleration significantly increased from 2.0˚ to 11.1˚ for inversion, and from 2.0˚ to 10.7˚ for eversion, at 60 through 360° · s^{-1}. Observing the results as a percentage of the total test ROM the inversion load range (Figure 7.10) significantly decreased from 91.4% to 45.8%, and eversion load range (Figure 7.11) significantly decreased from 90.3% to 44.0%, at 60 through 360° · s^{-1} respectively.

The 6 x 2 x 2 (speed x analysis type x movement) repeated measures ANOVA results revealed that normalised total peak torque (Figures 7.12 and 7.13) values significantly decreased with each increase in velocity for both ankle inversion and ankle eversion. There was no significant difference found between ankle inversion and eversion results. The normalised total peak torque values significantly decreased from 0.32 $Nm \cdot Kg^{-1}$ to 0.14 $Nm \cdot Kg^{-1}$ for inversion, and from 0.30 $Nm \cdot Kg^{-1}$ to 0.10 $Nm \cdot Kg^{-1}$ for eversion, at 60 through 360° · s^{-1}.

Normalised load range peak torque (Figures 7.12 and 7.13) values significantly decreased with each increase in velocity for both ankle inversion and ankle eversion. There was no significant difference found between ankle inversion and eversion results. The normalised load range peak torque values significantly decreased from 0.32 $Nm \cdot Kg^{-1}$ to 0.08 $Nm \cdot Kg^{-1}$ for inversion, and from 0.30 $Nm \cdot Kg^{-1}$ to 0.06 $Nm \cdot Kg^{-1}$ for eversion, at 60 through 360° $\cdot s^{-1}$. The 6 x 2 x 2 repeated measures ANOVA results also showed a significant difference (P<0.05) between normalised total peak torque and load range peak torque from speeds of 240° $\cdot s^{-1}$ and above for both ankle inversion and ankle eversion.

Table 7.7. Ankle Inversion and Ankle Eversion Acceleration, Load Range and Deceleration Range of Motion Across Velocities.

Velocity (° $\cdot s^{-1}$)	Acceleration (Degrees)	Load Range (Degrees)	Deceleration (Degrees)
Inversion			
60	1.1 (0.1)	31.9 (0.2)	2.0 (0.1)
120	2.0 (0.2) *	30.1 (0.2) *	2.9 (0.3) *
180	3.3 (0.3) *	27.2 (0.5) *	4.5 (0.3) *
240	4.8 (0.6) *	25.0 (0.8) *	5.2 (0.5) *
300	6.6 (0.7) *	20.1 (1.0) *	8.3 (0.6) *
360	7.9 (1.1) *	16.0 (1.6) *	11.1 (0.9) *
Eversion			
60	1.4 (0.2)	31.6 (0.2)	2.0 (0.2)
120	2.3 (0.3) *	29.7 (0.3) *	3.0 (0.2) *
180	3.4 (0.4) *	27.1 (0.5) *	4.5 (0.3) *
240	5.0 (0.6) *	24.6 (0.8) *	5.4 (0.4) *
300	7.3 (0.9) *	19.5 (0.9) *	8.2 (0.6) *
360	8.9 (1.0) *	15.4 (1.5) *	10.7 (0.8) *

Data are presented as Mean (SD). * Significantly (P<0.008) different from previous velocity

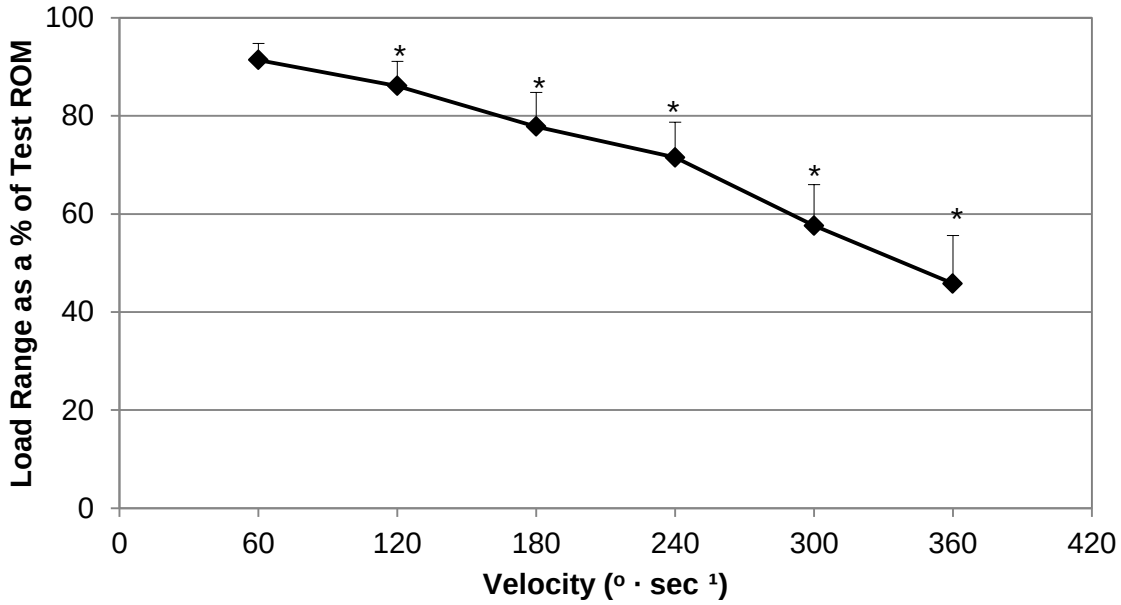


Figure 7.10. Load Range as a Percentage of the Total Test Range of Motion (ROM) During Ankle Inversion (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity.

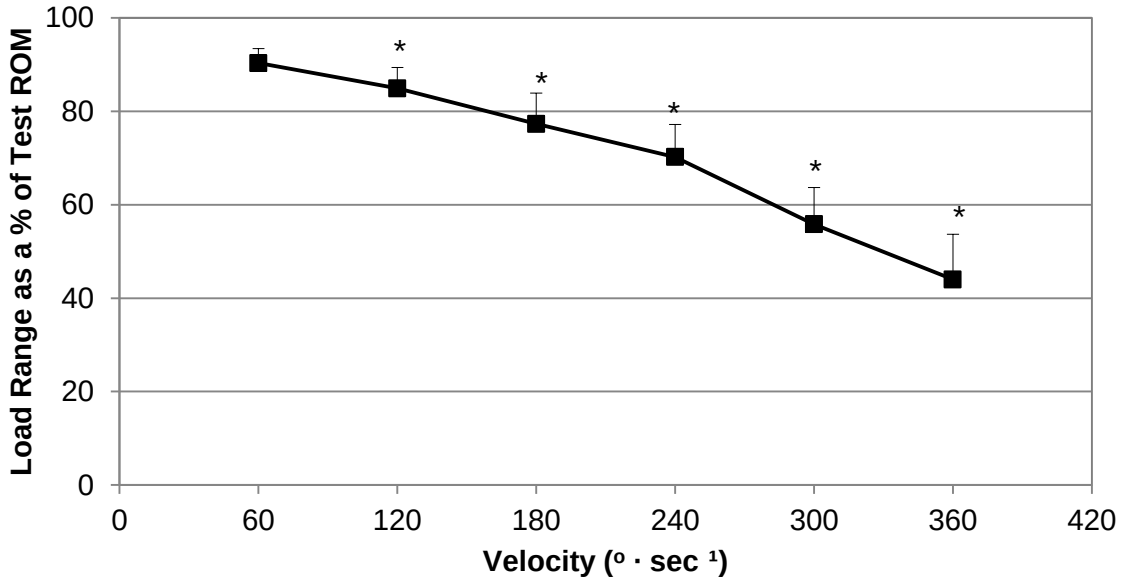


Figure 7.11. Load Range as a Percentage of the Total Test Range of Motion (ROM) During Ankle Eversion (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity.

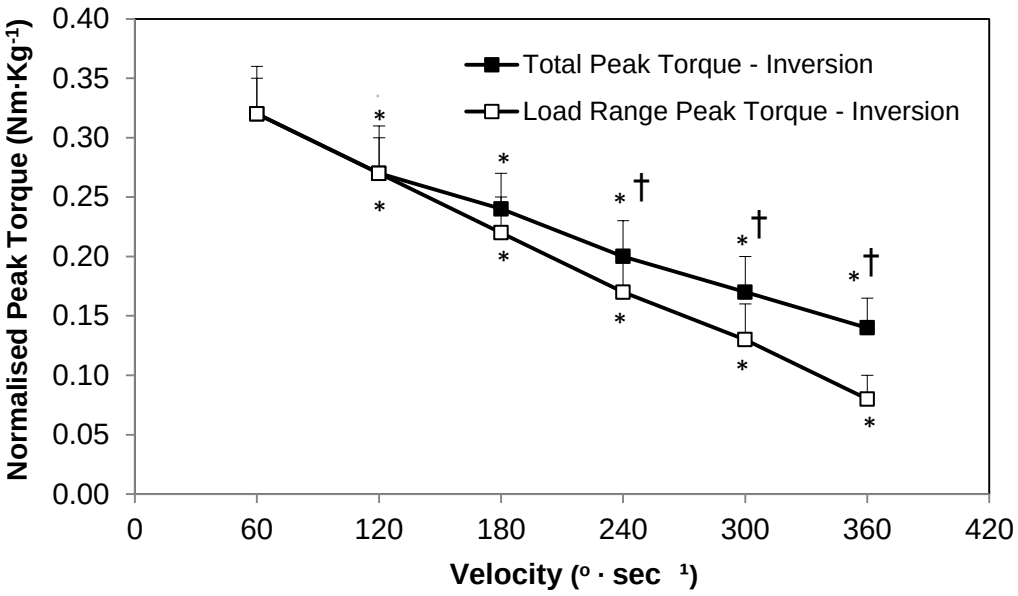


Figure 7.12. Normalised Total and Load Range Peak Torque for Ankle Inversion with Changes in Velocity (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity. † Significant difference (*P*<0.05) between normalised total peak torque and load range peak torque at corresponding velocity.

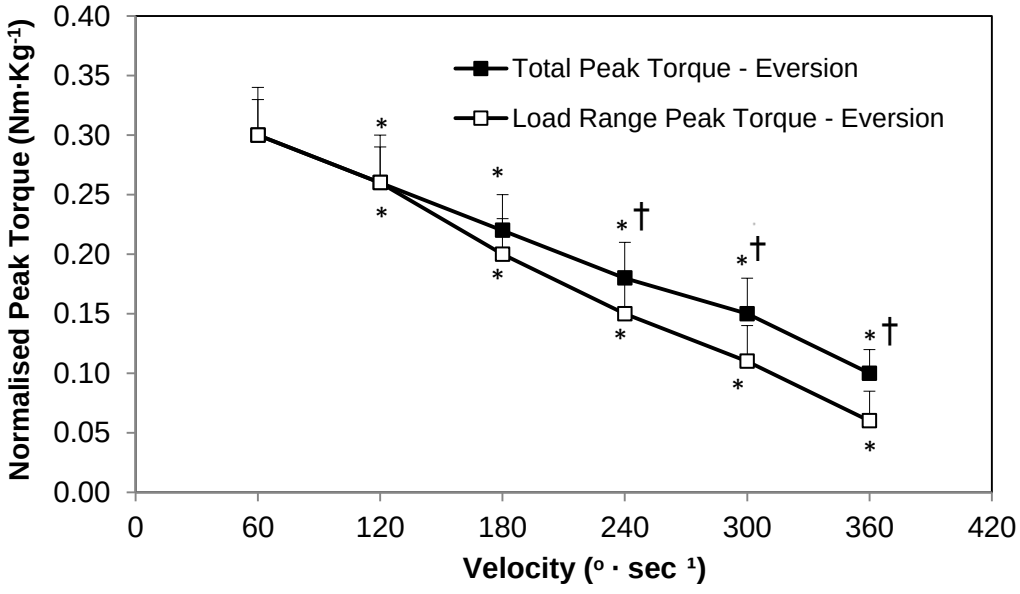


Figure 7.13. Normalised Total and Load Range Peak Torque for Ankle Eversion with Changes in Velocity (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity. † Significant difference (*P*<0.05) between normalised total peak torque and load range peak torque at corresponding velocity.

The 6 x 2 x 2 (speed x analysis type x movement) repeated measures ANOVA results revealed that normalised total work (Figures 7.14 and 7.15) values significantly decreased with each increase in velocity for both ankle inversion and ankle eversion. There was no significant difference found between ankle inversion and eversion results. The normalised total work values significantly decreased from 0.17 $Nm \cdot Kg^{-1}$ to 0.04 $Nm \cdot Kg^{-1}$ for inversion, and from 0.16 $Nm \cdot Kg^{-1}$ to 0.03 $Nm \cdot Kg^{-1}$ for eversion, at 60 through 360° $\cdot$ s^{-1}.

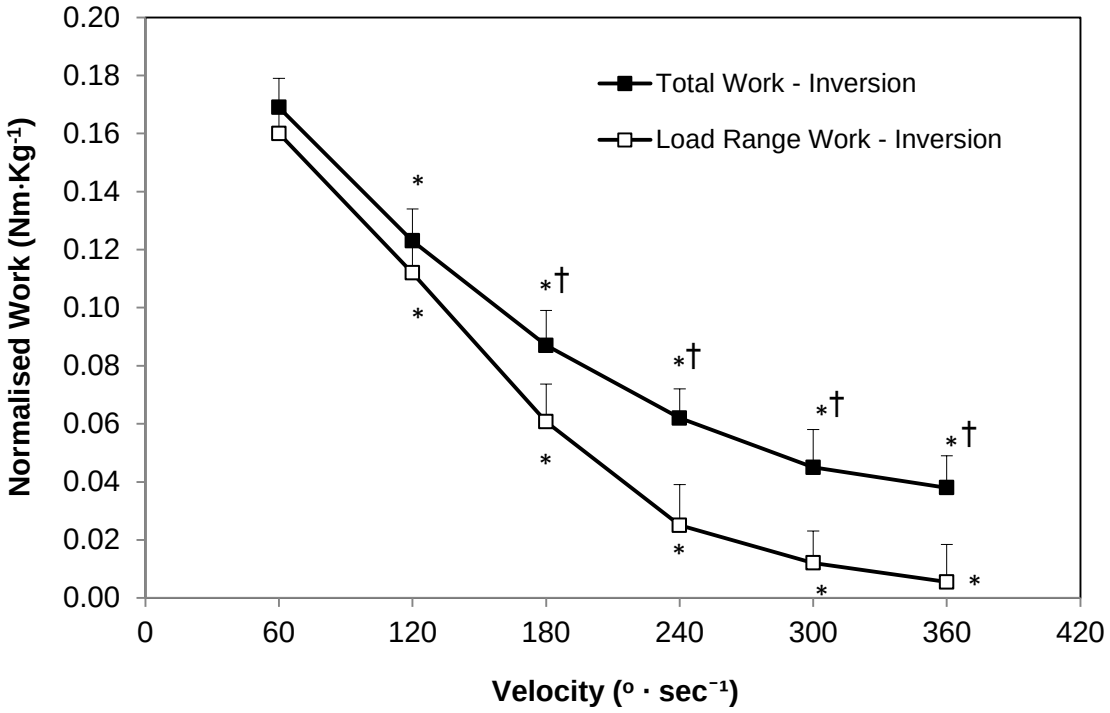


Figure 7.14. Normalised Total and Load Range Work for Ankle Inversion with Changes in Velocity (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity. † Significant difference (*P*<0.05) between normalised total work and load range work at the corresponding velocity.

Normalised load range work (Figures 7.14 and 7.15) values significantly decreased with each increase in velocity for both ankle inversion and ankle eversion. There was no significant difference found between ankle inversion and eversion results. The

normalised load range work values significantly decreased from 0.16 $Nm \cdot Kg^{-1}$ to 0.01 $Nm \cdot Kg^{-1}$ for inversion, and from 0.15 $Nm \cdot Kg^{-1}$ to 0.01 $Nm \cdot Kg^{-1}$ for eversion, at 60 through 360° $\cdot s^{-1}$. The 6 x 2 x 2 repeated measures ANOVA results also showed a significant difference between normalised total work and load range work from speeds of 180° $\cdot s^{-1}$ and above for ankle inversion, and 240° $\cdot s^{-1}$ and above for ankle eversion.

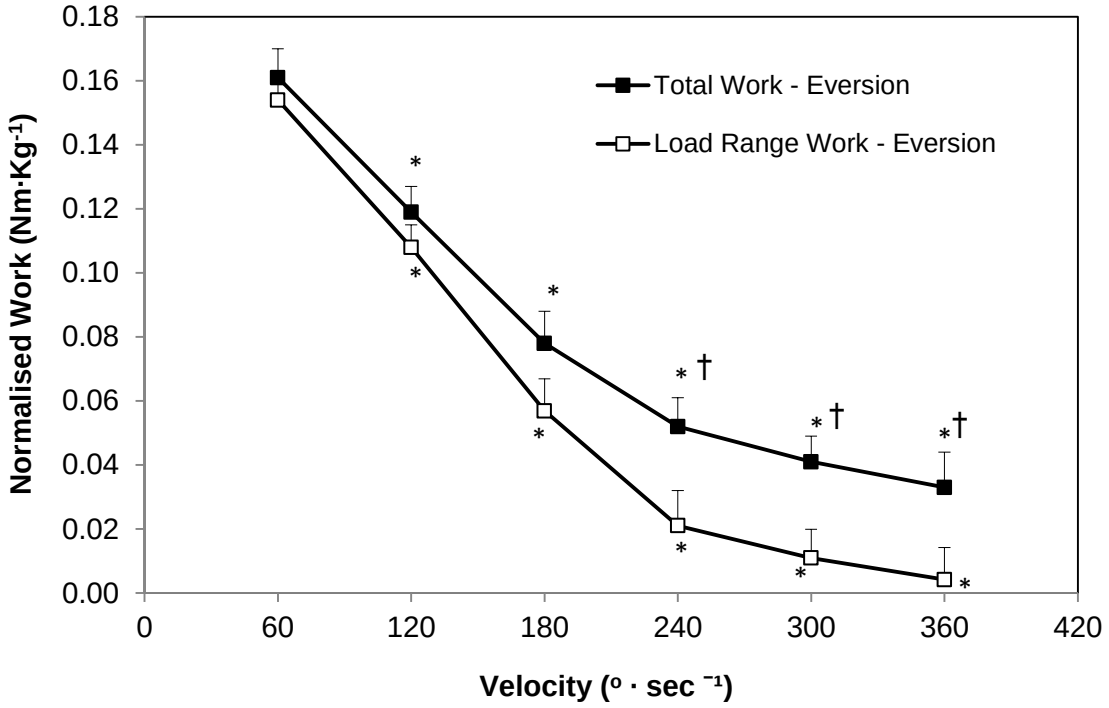


Figure 7.15. Normalised Total and Load Range Work for Ankle Eversion with Changes in Velocity (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity. † Significant difference (*P*<0.05) between normalised total work and load range work at the corresponding velocity.

7.6.5 Discussion

The aim of this study was to quantify the components of load range, acceleration, and deceleration through a velocity spectrum during concentric ankle inversion and eversion

isokinetic exercise. The secondary aim of the study was to investigate the effect of load range on peak torque and work done. It is apparent from the results that load range significantly decreased while acceleration and deceleration ROM significantly increased with each increase in velocity, for both ankle inversion and ankle eversion. Therefore, the hypothesis of this study which stated that with an increase in velocity there will be a significant decrease in the load range component, and a significant increase in the acceleration and deceleration components can be formally accepted. When the total peak torque data was corrected for load range there was a significant decrease in peak torque at velocities of 240° · s^{-1} and above for both ankle inversion and ankle eversion. Load range correction also resulted in a significant decrease in work done at velocities of 180° · s^{-1} and above for ankle inversion, and 240° · s^{-1} and above for ankle eversion. Therefore, the hypothesis of this study which stated the load range corrected peak torque and total work data will be significantly different to the uncorrected data at higher velocities can be formally accepted.

7.6.5 1 Load Range, Acceleration and Deceleration Quantification

The findings of the present study reflected past investigations in which isokinetic constant velocity movement was measured under concentric conditions. Osternig (1986) reported that knee extension load range decreased from 92% to 16% at speeds of 50 through 400° · s^{-1}. Also investigating the knee, Kurdak et al. (2005) found a reduction in load range from 94% to 4% for knee extension at speeds 30 through 390° · s^{-1}, and from 94.5% to 6.5% for knee flexion at speeds 30 through 450° · s^{-1}. Scibelli et al. (1993) demonstrated that bilateral knee extension/flexion load range decreased from 87.8% to 31.8% at speeds from 60 through 360° · s^{-1}. In addition to this, Brown,

Whitehurst, Findley et al. (1995) found that load range decreased from 95.3% to 0% and from 96.3% to 21.8% during shoulder external and internal rotation, respectively, at speeds from 60 through 450° · s^{-1}. Gautrey et al. (2013b) found that hip abduction load range decreased from 92.9% to 48.2%, and hip adduction load range decreased from 93.8% to 49.3%, at 60 through 360° · s^{-1} respectively. The current study found that load range decreased from 91.4% to 45.8% for ankle inversion, and from 90.3% to 44.0% for ankle eversion, at speeds of 60 through 360° · s^{-1}. It is apparent that the results of the present study mirror the findings of the above authors (Brown, Whitehurst, Gilbert et al., 1995; Gautrey et al., 2013b; Kurdak et al., 2005; Osternig, 1986; Scibelli et al., 1993), as they all found an inverse relationship between load range and isokinetic velocity. However, direct comparisons are difficult to make due to the different joints and musculature being tested.

Brown and Whitehurst (2000) highlighted the importance of separating the data into the three phases of acceleration, load range and deceleration. Surprisingly, some authors still fail to do this, and only consider the load range component (Kurdak et al., 2005). However, Brown, Whitehurst, Gilbert et al. (1995) and Gautrey et al. (2013b) did consider the impact of acceleration and deceleration and found that both components significantly increased with each increase in velocity. These results mirror the findings of the present study. However, direct comparisons of the results must be made with caution due to the fact that Brown, Whitehurst, Gilbert et al. (1995) studied the flexors and extensors of the knee, with a ROM of 80˚, and Gautrey et al. (2013b) studied the abductors and adductors at the hip, with a ROM of 45˚, whereas the present study investigated the invertors and evertors of the ankle with a ROM of 35˚.

By quantifying ROM for the load range, acceleration and deceleration components a more complete understanding of a concentric action on the isokinetic dynamometer can be achieved. The results emphasise the need for the clinician to fully understand the inverse relationship between isokinetic velocity and load range, and select the appropriate velocity accordingly. Any strength gains from training on the isokinetic dynamometer may be relative to the total amount of ROM actually sustained at the pre-selected velocity (ie, load range).

The results from the current study also emphasise the variation that exists between different joints. The load range, acceleration and deceleration components have never been previously quantified for ankle inversion and eversion isokinetic exercise. Even though the same general trend was indentified (load range significantly decreased while acceleration and deceleration ROM significantly increased with each increase in velocity), it can be seen that different joints have different levels of acceleration, load range, deceleration and maximum speed. These results further elucidate the findings that it is very important to load range correct data prior to analysis and that one cannot utilise factors from dissimilar joints. Therefore, the results from the present study should only be employed by future researchers if they are investigating the invertors and evertors of the ankle.

7.6.5.2 Load Range Correction for Peak Torque and Work Done

In the present study there was a significant difference between normalised total peak torque and load range peak torque from speeds of $240° \cdot s^{-1}$ and above for both ankle inversion and ankle eversion. There was also a significant difference between

normalised total work and load range work from speeds of 180° · s^{-1} and above for ankle inversion, and 240° · s^{-1} and above for ankle eversion. In agreement with these findings, Kurdak et al. (2005) found that the consideration of load range for peak torque and work calculations resulted in a significant decrease in the data when compared to the data presented by the isokinetic dynamometer. Kurdak et al. (2005) found a significant difference between total peak torque and load range peak torque at speeds above 270° · s^{-1} for knee extension, and above 300° · s^{-1} for knee flexion. They also found a significant difference between total work and load range work at speeds above 90° · s^{-1} for both knee extension and knee flexion (Kurdak et al., 2005). Gautrey et al. (2013b) found that when the total peak torque data was corrected for load range there was a significant decrease in peak torque at velocities of 300° · s^{-1} and above, for both hip abduction and hip adduction. Load range correction also resulted in a significant decrease in work done at velocities of 120° · s^{-1} and above, for both hip abduction and hip adduction. These results highlight the importance of correcting the data for load range, as it is apparent that large errors can occur if this process is not undertaken (Brown & Whitehurst, 2000).

The normalised load range peak torque values and the normalised load range work values in the present study were lower than the results reported by Kurdak et al. (2005). However, this was expected as Kurdak et al. (2005) studied the flexors and extensors of the knee joint and not the invertors and evertors of the ankle joint. Unfortunately the majority of studies investigating peak torque and work of the ankle invertors and ankle evertors do not normalise their data to the subject's body weight (Claiborne, Timmons & Pincivero, 2009; Jacobs, Uhl, Seeley, Sterling & Goodrich, 2005; Laheru Kerr & McGregor, 2007; Piva, Teixeira, Almeida, Gil, DiGioia, Levison et al., 2011). They also

do not indicate whether load range correction was completed (Claiborne et al., 2009; Jacobs et al., 2005; Johnson, Millie, Martinez, Crombie & Rogers, 2004; Laheru et al., 2007; Piva et al., 2011) which makes comparisons of the data difficult. An extensive search of the literature was conducted and no papers identifying that they had load range corrected their ankle isokinetic dynamometry data could be found. This paper highlights the need for past researchers to carefully reconsider the meaningfulness of their data, and in future, it is proposed that load range correction is conducted.

7.6.5.3 Clinical Implications

The results from the present study imply that peak torque and total work values should always be corrected by the clinician to account for load range, as otherwise errors may be present. As the isokinetic dynamometer is often used for training or rehabilitation, the results identify a need for the clinician to carefully consider velocity selection during ankle inversion and ankle eversion exercise. Any strength gains from isokinetic training may be proportional to the amount of time actually spent at the pre-selected velocity.

7.6.5.4 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. It must be remembered that the results are only applicable if the same joint, equipment and protocol is used as in the current study. Future studies may wish to repeat this study but using different joints, different makes of isokinetic dynamometers and varied protocols.

7.6.6 Conclusion

In summary, the results indicate that an inverse relationship exists between load range and velocity during concentric ankle inversion and ankle eversion isokinetic exercise. If the velocity is not reached, the result is in absence of machine offered resistance. In addition, the results emphasise the importance of also considering the acceleration and deceleration components, as these both significantly increased with each increase in velocity, for ankle inversion and ankle eversion.

The results also highlight the importance of correcting the data for load range, as it is apparent that large errors can occur if this process is not undertaken. Both peak torque and work decreased following load range correction. As the isokinetic dynamometer is often used for training or rehabilitation, the results identify a need for the clinician to carefully consider velocity selection during ankle inversion and ankle eversion exercise. Any strength gains from isokinetic training may be proportional to the amount of time actually spent at the pre-selected velocity (ie, load range).

7.7 Development of Research

Pilot Study Six addressed the effect of load range on peak torque and total work values during isokinetic ankle inversion-eversion. The results found that the peak torque and total work values produced by the isokinetic dynamometer should be adjusted to account for load range. Therefore, it was important that this method was adopted during Study Three and Four.

7.8 Pilot Study Seven: The Effect of Isokinetic Testing Speed on the Reliability of Muscle Fatigue Indicators during a Hip Abductor-Adductor Fatigue Protocol

7.8.1 Abstract

Aim: To investigate the reliability of fatigue indicators calculated from peak torque and total work during isokinetic speeds of 60, 90, 120 and 180° · s^{-1} during a hip fatigue protocol. **Method:** Ten males suffering from unilateral FAI and ten male healthy controls performed five maximal concentric contractions on an isokinetic dynamometer. Following a four minute rest period subjects were instructed to perform repeated maximal concentric contractions to fatigue, which was defined as three consecutive repetitions below 50% of the maximum peak torque value. Each testing speed was randomised with 24 hours between speeds. The subjects were asked to return to the laboratory seven days later to repeat the four speeds, with 24 hours between speeds. Muscle fatigue was determined for each testing speed by the fatigue index, the percent decrease in performance and the slope of the regression equation. **Results:** The most reliable fatigue determination method was the slope of the regression equation, when testing at a speed of 120° · s^{-1}. **Conclusion:** It is recommended that future investigators examine and plot their data before choosing the slope of the regression equation as their fatigue indicator, as a linear model is required.

7.8.2 Introduction

Hislop and Perrine (1967) originally introduced the concept of isokinetic dynamometry in 1967. Since then it has become a popular method for the assessment of muscle

performance, using common parameters such as peak torque and total work (Gleeson & Mercer, 1992). Recently, isokinetic dynamometry has been the favoured choice for fatigue assessment (Bosquet et al., 2010; Croisier, Ganteaume, Binet, Genty & Ferret, 2008).

Fatigue has been defined as any reduction in the force generating capacity of the total neuromuscular system regardless of the force required in any given situation (Bigland-Richie & Woods, 1984). The ability to objectively document muscle fatigue has been an area of concern in both research and clinical settings (Gleeson & Mercer, 1992). Researchers and healthcare professionals should adopt a reliable method to quantify this manifestation of exercise. The development of isokinetic dynamometry has provided a stepping stone towards objectively measuring muscle fatigue. It should be kept in mind that the reliability of isokinetic testing for a desired protocol should be sufficient so that measures for training or injury induced changes in muscle ability are not attributed to instrument or testing error.

It has been consistently demonstrated that isokinetic peak torque and total work are reliable measures (Bosquet et al., 2010; Brown et al., 1993; Gross et al., 1991; Pincevero et al., 1997; Sole et al., 2007). In regards to the assessment of muscle fatigue using this modality, research is sparse and questionable reliability values have been demonstrated (Gleeson & Mercer, 1992; Larsson et al., 2003; Pincevero et al., 2001). Furthermore, the vast majority of studies using isokinetic methods have focused on peak torque, rather than total work (Gleeson & Mercer, 1996). Peak torque represents only one point of the moment-angular position curve, the highest one (Bosquet et al., 2010). It may not adequately describe other torques developed

throughout the movement. This is the reason why total work, which represents the area under the curve (Hislop & Perrine, 1967), should also be considered. This parameter accounts for the overall adaptation of the curve, not only its highest value.

There is also a lack of consensus in the literature regarding the techniques used to calculate fatigue. The original recommendations proposed by Thorstensson and Karlsson (1976) stated that muscle fatigue should be measured with the fatigue index (FI), calculated as the ratio of the average peak torque of the last three contractions to the average peak torque of the first three contractions (Bosquet et al., 2010). More recently, Pincivero et al. (2001) acknowledged that given the linear nature of the relationship between the total work of each contraction and the number of maximal concentric contractions, the slope of the regression equation would be more appropriate to determine the rate of decrease of total work and thus to estimate muscle fatigue. Although not specific to isokinetic dynamometry, Glaister et al. (2004) quantified muscle fatigue during repeated maximal sprints using the percent decrement score; they argued the suitability of this method since it considered data from each effort in its calculation.

To date, the knee has been the focus for the majority of the isokinetic dynamometry reliability studies (Bosquet et al., 2010; Brown et al., 1993; Feiring, Ellenbecker & Dershield, 1990; Gross et al., 1991; Sole et al., 2007), with limited studies focusing on the hip (Claiborne et al., 2009; Piva et al., 2011). In addition to this there is negligible research investigating the reliability of different muscle fatigue indicators (Bosquet et al., 2010). At present, there has been no research on the development of a reliable fatigue protocol of the hip musculature.

Reliability studies are frequently performed on healthy populations (Bosquet et al., 2010; Brown et al., 1993; Sole et al., 2007; Taylor et al., 1991), however, isokinetic dynamometry is commonly used to test subjects that are recovering from injury. It is important that equipment is shown to be reliable in healthy populations, but also that it is reliable in persons that have recovered from injury. The incidence of FAI, exhibiting residual symptoms such as feelings of instability, giving way, pain or re-injury, is a common development following an initial ankle sprain. Many sufferers of FAI go months or years without suffering an ankle sprain, and are therefore termed healthy patients but with a history of FAI. Clinicians and health professionals in sport will often use isokinetic dynamometry to test this population, as well as healthy individuals, throughout the sporting season. It is therefore important that the equipment used is reliable in both healthy subjects, and patients with a history of FAI.

It has been suggested that the hip abductors play a critical role in controlling foot placement during ambulation. A deficit at the hip abductors may alter foot placement, causing the foot to contact the ground in a more adducted position (Friel et al., 2006). In patients with FAI this potentially increases the chance of rolling over on the ankle and sustaining a lateral ankle sprain. In addition to this, the added factor of localised hip muscle fatigue in FAI sufferers may further increase the probability of suffering an ankle sprain (Friel et al., 2006).

Therefore, the purpose of this study was to determine the effect of isokinetic testing speed on the relative and absolute reliability of the fatigue index, percent decrease in performance and slope of the regression equation during a hip abduction-adduction fatigue protocol, in FAI subjects and healthy controls.

7.8.3 Method

7.8.3.1 Subjects

The same subjects were used as in Pilot Study One (Section 3.2.3.1)

7.8.3.2 Experimental Design

Subject's age, mass and height were recorded. Testing was performed on the Biodex System 2 Isokinetic Dynamometer (Biodex Medical Systems, Shirley, New York). The Biodex was set up according to the Biodex System 2 Manual, and was calibrated according to manufacturer's specifications prior to testing. The cushion control was set to zero, to allow the subject the greatest availability of velocity attainment prior to deceleration (Brown, Whitehurst, Gilbert et al., 1995). All subjects completed a practice session on the isokinetic dynamometer a week prior to the main testing procedure.

Subjects were required to lie on their side (facing away from the dynamometer power head) with the tested hip (right) superior to the opposite hip. The knee of the tested leg was extended, and the opposite knee was flexed at 90°. The axis of the dynamometer was aligned superior and medial to the greater trochanter of the tested leg. The subject's right leg was attached to the Biodex hip attachment, superior to the lateral knee joint line. The subject's range of motion was set between 0-45° of abduction. The degrees of motion were set based on the average limitations of hip motion in healthy individuals (Emery et al., 1999; Makofsky, Panicker, Abbruzzese, Aridas, Camp, Drakes

et al., 2007; Reid, 1992). The subject's hand grasped the border of the chair (Figure 7.16).

The subject was then instructed to perform five isokinetic maximal repetitions, to determine their maximum peak torque. Each subject was instructed to push their leg upwards (abduction) and pull their leg downwards (adduction) as hard and as fast as possible. The maximum peak torque value was established and subjects were given a four minute rest period (Salavati et al., 2007). Following this, subjects were instructed to abduct and adduct their hip repeatedly as hard and as fast as possible until they reached fatigue. Fatigue was defined as three consecutive repetitions below 50% of the maximum peak torque value (Emery et al., 1999; Gear, 2011; Salavati et al., 2007). The same strong verbal encouragements were given to each subject throughout the test to motivate them to develop maximal torque during each repetition (McNair et al., 1996). Each testing speed (60, 90, 120 and 180° · s^{-1}) was randomised with a minimum of 24 hours between speeds. The subjects were then given a seven day rest period, and were asked to return to the laboratory to repeat the four speeds again with a minimum of 24 hours between speeds. Therefore, each subject visited the laboratory on eight separate occasions to complete all testing sessions. Subjects were asked to refrain from any vigorous exercise during the week, and were tested at the same time of day to reduce the effect of diurnal variation.

7.8.3.3 Data Analysis

The same data analysis was used as in Pilot Study Four (Section 7.2.3.3). Again load ranged showed an inverse relationship to velocity for the hip abductors (Figure 7.17).

Therefore, all peak torque and total work data was reduced for load range prior to analysis. Muscle fatigue was only determined for the hip abductors, as the gluteus medius was the focused muscle for the fatigue protocol, due to its stabilising role in the frontal plane at the hip.

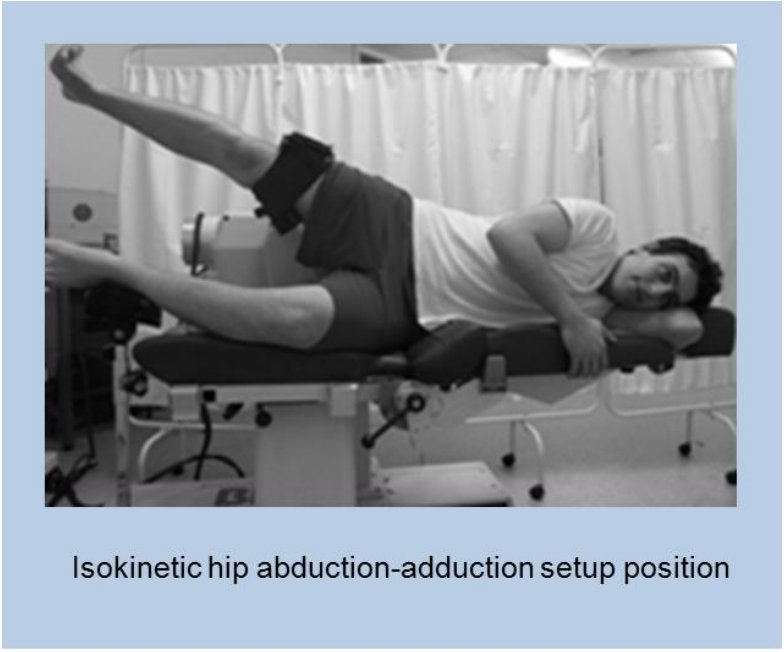


Figure 7.16. Isokinetic Hip Abduction-Adduction Setup Position. Image shows leg abducted.

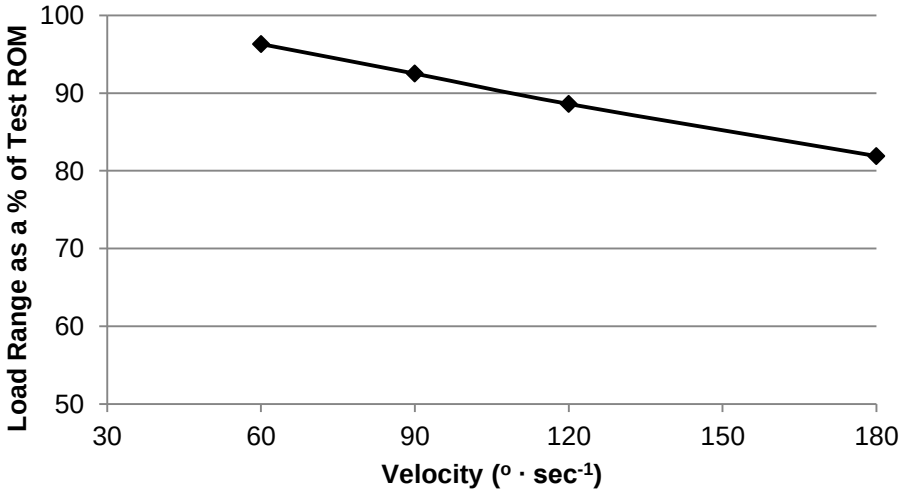


Figure 7.17. Load Range as a Percentage of the Total Test Range of Motion (ROM) During Hip Abduction (Mean ± SD).

7.8.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. Systematic bias, which refers to a difference in measurements in a particular direction between repeated tests, was assessed with seven (peak torque [3 fatigue indicators], total work [3 fatigue indicators] and number of repetitions to fatigue) 2 x 4 x 2 (subjects type [healthy or FAI] × speed [60, 90, 120 and 180° · s^{-1}] × time [first week testing or second week testing]) mixed factorial analysis of variance (ANOVA). The two within-subject factors were speed and time of test, and the between-subject factor was subject type. Sphericity was verified for all data being compared by the Mauchly test. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of homogeneity of intercorrelations. The Multivariate Test box (Wilk's Lambada value) was studied for three-way interactions, then two-way interactions and then main effects, to identify differences for the within-subject factors (speed and time) (P<0.05). The Test of Between-Subject Effects box was observed to identify differences for the between-subject factor (subject type) (P<0.05). The Pairwise Comparisons post-hoc test was used to determine exactly where the significant findings occurred for the within-subject factor when there were more than two conditions (speed). Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.0125.

Relative reliability was assessed by calculating the $ICC_{(2,1)}$. From the ICC value the SEM was calculated, which represented absolute reliability (Section 3.2.3.4, paragraph 2).

7.8.4 Results

7.8.4.1 Peak torque

The mixed factorial ANOVA showed no significant differences for the fatigue indicators (fatigue index, percent decrease in performance and the slope of the regression equation) between the first week and second week of testing. There was also no significant difference between the two groups tested (healthy and FAI). However, there was a significant decrease (P<0.0125) in peak torque with each increase in velocity. When studying the relative reliability results for the healthy subject's peak torque (table 7.8) the values ranged from moderate to very high depending on which fatigue determination method was used. The fatigue index showed moderate to high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.60, 0.74, 0.88 and 0.78, respectively. Absolute reliability for the fatigue index showed SEM values of 0.94%, 1.14%, 1.03% and 1.36%, for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method showed high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.78, 0.85, 0.88 and 0.77, respectively. Absolute reliability for the percent decrease in performance showed SEM values of 0.93%, 0.64%, 0.52% and 0.94%, for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation showed high to very high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.85, 0.91, 0.93

and 0.91, respectively. Absolute reliability for the slope of the regression equation showed SEM values of 0.06%, 0.08%, 0.02% and 0.07% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation was the most reliable method of fatigue determination, and the most reliable testing speed was 120° · s^{-1}.

When studying the relative reliability results for the FAI subjects the peak torque values (table 7.8) ranged from moderate to very high depending on which fatigue indicator was used. The fatigue index showed high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.79, 0.77, 0.71 and 0.75, respectively. Absolute reliability for the fatigue index showed SEM values of 0.41%, 0.41%, 0.44% and 0.77% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method showed moderate to very high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.69, 0.71, 0.93 and 0.80, respectively. Absolute reliability for the percent decrease in performance showed SEM values of 0.40%, 0.35%, 0.26% and 0.82% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation showed high to very high reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.91, 0.80, 0.92 and 0.83, respectively. Absolute reliability for the slope of the regression equation showed SEM values of 0.04%, 0.07%, 0.01% and 0.03% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method had the highest reliability (ICC=0.93, SEM=0.26) when testing at 120° · s^{-1}. This was followed by the slope of the regression equation (ICC=0.92, SEM=0.01) when testing at 120° · s^{-1}.

7.8.4.2 Total work

The mixed factorial ANOVA showed no significant differences for the fatigue indicators (fatigue index, percent decrease in performance and the slope of the regression equation) between the first week and second week of testing. There was also no significant difference between the two groups tested (healthy and FAI). However, there was a significant decrease (P<0.0125) in peak torque with each increase in velocity. When studying the relative reliability results for the healthy subject's total work (table 7.9) the values ranged from moderate to very high depending on which fatigue determination method was used. The fatigue index showed moderate to high reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.65, 0.78, 0.64 and 0.60, respectively. Absolute reliability for the fatigue index showed SEM values of 1.26%, 0.86%, 1.17% and 1.08% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method showed high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.82, 0.81, 0.81 and 0.82, respectively. Absolute reliability for the percent decrease in performance showed SEM values of 0.99%, 0.91%, 0.89% and 0.82% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation showed high to very high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.87, 0.86, 0.92 and 0.92, respectively. Absolute reliability for the slope of the regression equation showed SEM values of 0.04%, 0.05%, 0.02% and 0.03% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The results showed the slope of the regression to be the most reliable method of fatigue determination, when testing at a speed of 120°.

When studying the reliability results for the subjects with FAI subjects the total work values (table 7.9) ranged from moderate to very high depending on which fatigue determination method was used. The fatigue index showed moderate to high reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.74, 0.71, 0.81 and 0.62, respectively. Absolute reliability for the fatigue index showed SEM values of 0.85%, 0.92%, 0.58% and 0.61% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The percent decrease in performance method showed moderate to high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.78, 0.75, 0.80 and 0.60, respectively. Absolute reliability for the percent decrease in performance showed SEM values of 0.19%, 0.25%, 0.26% and 0.73% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The slope of the regression equation showed low to very high relative reliability, with speeds 60, 90, 120 and 180° · s^{-1} showing ICC results of 0.86, 0.84, 0.90 and 0.71, respectively. Absolute reliability for the slope of the regression equation showed SEM values of 0.02%, 0.03%, 0.01% and 0.05% for speeds 60, 90, 120 and 180° · s^{-1}, respectively. The results showed the slope of the regression to be the most reliable method when testing at 120° · s^{-1}.

Table 7.8. Muscle Fatigue Indicators from Peak Torque Data During Hip Abduction. Data are presented as Mean (SD).

PARAMETER	HEALTHY SUBJECTS				FAI SUBJECTS			
	TEST 1	TEST 2	ICC	SEM (%)	TEST 1	TEST 2	ICC	SEM (%)
60° · s^{-1}								
FI (%)	64.25 (2.44)	63.23 (2.34)	0.60	0.94	63.33 (2.26)	63.22 (2.55)	0.79	0.41
DP (%)	41.12 (4.63)	40.21 (4.78)	0.78	0.93	40.07 (4.69)	39.82 (4.55)	0.69	0.40
Slope (Nm·rep^{-1})	-1.45 (0.13)	-1.68 (0.15)	0.85	0.06	-1.41 (0.12)	-1.46 (0.13)	0.91	0.04
N^{o} Reps to Fatigue	53.00	57.00	0.68	1.31	55.00	58.00	0.65	1.09
90° · s^{-1}								
FI (%)	63.18 (2.67)	61.11 (2.75)	0.74	1.41	63.16 (2.36)	62.71 (2.26)	0.77	0.41
DP (%)	39.92 (4.37)	39.03 (4.19)	0.85	0.64	39.96 (4.57)	39.70 (4.64)	0.71	0.35
Slope (Nm·rep^{-1})	-1.07 (0.12)	-0.96 (0.13)	0.91	0.08	-1.30 (0.11)	-1.18 (0.10)	0.80	0.07
N^{o} Reps to Fatigue	53.00	56.00	0.62	1.05	54.00	51.00	0.64	0.98
120° · s^{-1}								
FI (%)	62.05 (2.95)	63.51 (2.90)	0.88	1.03	62.79 (2.41)	63.05 (2.25)	0.71	0.44
DP (%)	38.48 (4.42)	40.71 (4.37)	0.88	1.52	38.86 (4.63)	39.02 (4.73)	0.93	0.26
Slope (Nm·rep^{-1})	-1.17 (0.10)	-1.31 (0.09)	0.93	0.02	-1.21 (0.09)	-1.21 (0.09)	0.92	0.01
N^{o} Reps to Fatigue	54.00	55.00	0.69	1.13	52.00	50.00	0.69	1.04
180° · s^{-1}								
FI (%)	60.60 (2.97)	58.58 (2.89)	0.78	1.36	62.03 (2.32)	61.05 (2.60)	0.75	0.77
DP (%)	37.37 (4.63)	36.34 (4.58)	0.77	0.94	38.60 (4.61)	39.75 (4.63)	0.80	0.82
Slope (Nm·rep^{-1})	-1.09 (0.11)	-1.07 (0.12)	0.91	0.07	-1.15 (0.07)	-1.12 (0.06)	0.83	0.03
N^{o} Reps to Fatigue	48.00	49.00	0.66	1.21	49.00	51.00	0.76	1.09

Table 7.9. Muscle Fatigue Indicators from Total Work Data During Hip Abduction. Data are presented as Mean (SD).

PARAMETER	HEALTHY SUBJECTS				FAI SUBJECTS			
	TEST 1	TEST 2	ICC	SEM (%)	TEST 1	TEST 2	ICC	SEM (%)
60° · s^{-1}								
FI (%)	61.14 (3.78)	62.31 (3.91)	0.65	1.26	59.01 (3.88)	57.84 (3.97)	0.74	0.85
DP (%)	40.39 (2.78)	41.52 (2.72)	0.82	0.99	38.90 (2.83)	38.90 (2.83)	0.78	0.19
Slope (Nm·rep^{-1})	-0.78 (0.11)	-0.73 (0.11)	0.87	0.04	-0.74 (0.11)	-0.71 (0.12)	0.86	0.02
N^{o} Reps to Fatigue	53.00	57.00	0.68	1.31	55.00	58.00	0.65	1.09
90° · s^{-1}								
FI (%)	59.56 (3.73)	60.48 (3.70)	0.78	0.86	57.88 (3.88)	57.88 (3.88)	0.71	0.92
DP (%)	39.36 (2.86)	39.53 (3.12)	0.81	0.91	38.53 (2.86)	38.34 (2.75)	0.75	0.25
Slope (Nm·rep^{-1})	-0.75 (0.10)	-0.76 (0.12)	0.86	0.05	-0.76 (0.12)	-0.80 (0.11)	0.84	0.03
N^{o} Reps to Fatigue	53.00	56.00	0.62	1.05	54.00	51.00	0.64	0.98
120° · s^{-1}								
FI (%)	59.08 (3.82)	58.61 (3.90)	0.64	1.17	56.97 (3.89)	56.76 (3.97)	0.81	0.58
DP (%)	38.25 (2.90)	36.99 (2.92)	0.81	0.89	38.21 (2.89)	37.64 (3.00)	0.80	0.26
Slope (Nm·rep^{-1})	-0.72 (0.11)	-0.71 (0.11)	0.92	0.02	-0.73 (0.12)	-0.71 (0.11)	0.90	0.01
N^{o} Reps to Fatigue	54.00	55.00	0.69	1.13	52.00	50.00	0.69	1.04
180° · s^{-1}								
FI (%)	57.82 (3.91)	57.07 (4.02)	0.60	1.08	56.41 (4.08)	55.43 (3.87)	0.62	0.61
DP (%)	37.48 (3.03)	37.12 (3.01)	0.82	0.82	37.46 (3.12)	37.23 (3.30)	0.60	0.73
Slope (Nm·rep^{-1})	-0.71 (0.10)	-0.67 (0.09)	0.92	0.03	-0.70 (0.12)	-0.67 (0.10)	0.71	0.05
N^{o} Reps to Fatigue	48.00	49.00	0.66	1.21	49.00	51.00	0.76	1.09

7.8.4.3 Number of repetitions to fatigue

The mixed factorial ANOVA showed no significant differences for the number of repetitions to fatigue between the first week and second week of testing. There were also no significant differences between the two groups tested (healthy and FAI), or the four speeds tested (60, 90, 120 and 180° · s^{-1}). The number of repetitions to fatigue was correlated to the ICC reliability values to see if a relationship was present. The healthy subjects produced *r* values of 0.16 and 0.02, for testing session 1 and testing session 2, respectively. The FAI subjects produced *r* values of 0.41 and 0.17, for testing session 1 and testing session 2, respectively. The results showed there was no relationship present when correlating the number of repetitions to fatigue with the ICC.

7.8.5 Discussion

The aim of this study was to examine the test-retest reliability of fatigue measures calculated from peak torque and total work during isokinetic speeds of 60, 90, 120 and 180° · s^{-1} during an isokinetic hip abductor-adductor fatigue protocol. The main findings that emerged from the study were firstly, the slope of the regression equation was the most reliable method of fatigue determination in healthy subjects and FAI subjects, when using peak torque or total work values, and secondly, the most reliable fatigue measures occurred at the speed of 120° · s^{-1}.

The choice of either peak torque or total work to assess average performance during a fatigue test does not seem to influence relative or absolute reliability. The same conclusion applies to the speed of the isokinetic dynamometer as relative and absolute

reliability values were not influenced by a change in speed. When observing the peak torque values at 60, 90, 120 and 180° · s^{-1} the relative reliability values (ICC) for the slope of the line measure were consistently between 0.85-0.93 for the healthy subjects and 0.80-0.92 for the FAI subjects. Absolute reliability (SEM) also produced consistently low values between 0.02%-0.08% for the healthy subjects, and 0.01%-0.07% for the FAI subjects. The same can be observed with the total work values for the different isokinetic dynamometry speeds, as relative reliability for the slope of the line measure were between 0.86-0.92 for healthy subjects, and 0.84-0.90 for the FAI subjects. Absolute reliability values were again consistent with the total work measure producing values between 0.02%-0.05% for the healthy subjects and 0.01%-0.05% for the FAI subjects. It is also apparent from the above results that the type of subjects tested (healthy or FAI) did not influence relative or absolute reliability results.

The different fatigue determination methods did produce large variations in relative and absolute reliability values. The slope of the line measurement consistently produced high relative and absolute reliability values. Whereas, the fatigue index and the percentage decrease in performance produced lower and more variable relative and absolute reliability values (Tables 7.8 and 7.9).

A limited number of studies have looked at the reliability of different fatigue measures. Bosquet et al. (2010) found high relative reliability (peak torque ICC's = 0.82-0.88, total work ICC's = 0.81-0.87) for the slope of the line. Pincivero et al. (2001) studied the reliability of the fatigue index and the slope of the line during isokinetic quadriceps femoris muscle fatigue. The authors found moderate to high ICC's for the non-dominant leg (0.78-0.92) and high ICC's for the dominant leg (0.82-0.89) when analysed by the

slope. These results agree with the present study as we found the slope of the line to be the most reliable method when observing both relative and absolute reliability.

The appropriateness of a method to objectively quantify muscle strength or endurance is dependent upon its reliability and the inherent error associated with that method. Piva et al. (2011) illustrated high test-retest reliability of an isometric hip abduction protocol (ICC = 0.92). Claiborne et al. (2009) showed that peak torque measurements during isokinetic hip abduction at $60° \cdot s^{-1}$ displayed high test-retest reliability (ICC range = 0.81-0.91).

Although not directly related to the hip, Feiring et al. (1990) showed that quadriceps peak torque and total work displayed high test-retest reliability (ICC = 0.96 and ICC = 0.97, respectively). It should be recognised that the ability of reproducing the testing protocol with respect to adequate calibration, gravity correction, and standard patient set up in the current study was likely to have improved accuracy, and should be deemed important components for improving the reliability of a test (Gross et al., 1991; Pincevero et al., 1997; Pincevero et al., 2001; Winter et al., 1981).

The accuracy to which these protocols are reproducible is also a critical factor as determined by the SEM. Although high reliability coefficients (such as ICC's) have been previously reported for isokinetic strength, SEM values have received little attention in the literature. The SEM value in this study was expressed as a percentage in order to allow clinical usage of these measures. As demonstrated by the results of the current study, re-test values for peak torque and total work varied by 0.01 - 1.52% to the initial test. It should therefore, seem appropriate in future studies to attribute differences in

isokinetic results to intervention, training improvements or injury, should they exceed the SEM values outlined in tables 7.8 and 7.9.

There seems to be a lack of consensus in the literature on the most appropriate or reliable speed to be used for isokinetic dynamometry. The knee joint has been well documented with the majority of authors opting for a speed of $180° \cdot s^{-1}$. The hip, however, has very rarely been studied. The sparse literature that has tested the hip joint have either used the isokinetic dynamometer in its isometric mode (Piva et al., 2011), or have tested at speeds of 60 and $90° \cdot s^{-1}$ (Salavati et al., 2007). Ferber, McClay-Davis & Williams (2003) found that during running at 3.65 m/s (13.2 km/hr) over a 25m distance, the peak angular velocity for hip flexion was $103.5° \cdot s^{-1}$. The present study found $120° \cdot s^{-1}$ to be the most reliable testing speed, so even though this speed may be far from 'explosive sporting movement' velocities, it may replicate speeds from more endurance based activities as shown by Ferber et al. (2003).

Both peak torque and total work decreased during the test. Three methods were used to quantify this force reduction: the fatigue index (Kannus, 1994), percentage decrease in performance (Glaister et al., 2004) and the slope (Pincevero et al., 2001). The fatigue index and the percent decrease in performance measure the percentage of force reduction throughout the test. The slope represents the rate of decrease in performance. The main assumption, stated by Bosquet et al. (2010), for using this measure is the linearity of the relationship between peak torque or total work and the number of repetitions.

Previous studies have reported a linear relationship between performance and the number of repetitions during 20 (Maffiuletti et al., 2007) and 30 (Thorstensson & Karlsson, 1976) maximal reciprocal concentric contractions. The slope could therefore be used to quantify muscle fatigue. However, Bosquet et al. (2010) stated that there was a tendency of the line to plateau after 40 repetitions, and suggests that an exponential model would be more appropriate than a linear one to fit performance data measured for longer protocols. Studies by Gerdle and Elert (1994) and Larsson et al. (2003) also agree with the finding of Bosquet et al. (2010). However, the present study did not use a fixed number of repetitions and one subject reached 67 repetitions before 3 contractions were below 50% of their maximum peak torque. This subject still presented with a linear model, rather than an exponential decrease which would contrast with the above literature. The above studies were all performed on the knee, whereas the present study was fatiguing the hip musculature. We would recommend that future investigators examine and plot their data before choosing the slope of the line as their fatigue determination method, as a linear model is required. As a point of interest the number of repetitions to fatigue was correlated to the ICC reliability values to see if a relationship was present. However, the results showed no correlation between these two variables.

Limited research has focused on fatigue protocols of the hip musculature (Claiborne et al., 2009; Salavati et al., 2007). Current theory suggests that the hip abductors play a critical role in controlling foot placement during ambulation. A deficit at the hip abductors may alter foot placement, causing the foot to contact the ground in a more adducted position (Friel et al., 2006). In patients with FAI this potentially increases the chance of rolling over on the ankle, inducing a lateral ankle sprain. It has been suggested that

fatigue of the hip abductors may also cause these deficits at the ankle joint (Friel et al., 2006). Therefore, it was crucial to develop a reliable hip fatiguing protocol for the hip abductors, so that research can continue to investigate this phenomenon.

7.8.5.1 Clinical Implications

The results from the current study showed that the isokinetic dynamometer was a reliable device for testing the fatigability of the hip abductors in both healthy individuals but also individuals with a history of FAI. Many individuals in the sporting population suffer from a history of FAI, and the results from this study conclude that clinicians and other health professionals can perform isokinetic testing protocols on the hip abductors with confidence that the protocol is reliable, in not only healthy individuals, but also the large population of individuals with a history of FAI.

7.8.5.2 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. It must be remembered that the results are only applicable if the same joint, equipment and protocol is used as in the current study. Future studies may wish to repeat this study but using different joints, different makes of isokinetic dynamometers and varied protocols.

7.8.6 Conclusion

In summary, the most reliable fatigue determination method for the hip abductors was the slope of the regression equation, when testing at a speed of 120° · s^{-1}. However, it is recommended that future investigators examine and plot their data before choosing this as their fatigue indicator, as a linear model is required. The choice of either peak torque or total work to assess performance during a fatigue test did not influence relative or absolute reliability. The between-day reliability that was performed in the present study has valuable research and clinical relevance. Many athletic or rehabilitation activities typically involve multiple bouts of testing, sometimes with high-intensity muscle contractions. Protocols and methods used for testing should always be determined as reliable before testing commences, so that any differences that are reported can be reported as true.

7.9 Development of Research

Pilot Study Seven addressed the issue of identifying the most reliable isokinetic speed to be used for Study Three and Four. The results identified 120° · s^{-1} as the most reliable testing speed during the isokinetic hip abduction-adduction fatigue protocol. Therefore, this speed was deemed suitable for use in Study Three and Four.

7.10 Pilot Study Eight: The Effect of Segmental Stabilisation on the Reliability of Peak Torque and Total Work During Isokinetic Hip Abduction-Adduction Testing

7.10.1 Abstract

Aim: To compare the test-retest reliability of two setup positions during isokinetic hip abduction-adduction exercise, and to investigate the effect of setup position on peak torque and total work. **Method:** Sixteen male healthy subjects performed three maximal concentric hip abduction-adduction repetitions, on their right leg, at 60, 120, 180, 240, 300 and 360° · s^{-1}, during two setup positions (non-stabilised setup (NS) vs. stabilised setup (SS)). Setup position was randomised with 24 hours between. Subjects returned to the laboratory 7 days later for repeat testing. **Results:** The results indicated that the SS produced significantly greater (P<0.05) peak torque and total work values. The SS also produced marginally higher ICC results and lower SEM variance at speeds 60 though 240° · s^{-1}. However, speeds 300 and 360° · s^{-1} only showed moderate reliability. In the field of research where reliable protocols are a necessity, the use of three simple, and easy to apply additional straps during the SS, improved the reliability and magnitude of peak torque and total work measures. **Conclusion:** The results suggest that researchers and clinicians should opt for speeds between 60 and 240° · s^{-1} if they are conducting repeated tests, and require a reliable protocol.

7.10.2 Introduction

In recent years, the role of the hip abductors and hip adductors have become of great interest (Ekstrand & Hilding, 1999; Emery et al., 1999; Fredericson, Cookingham,

Chaudhari, Dowdell, Oestreicher & Sahrmann, 2000; Friel et al., 2006; Gautrey et al., 2013b; Holmich, 1998; Inman, 1947; Laheru et al., 2007; Nadler, Malanga, DePrince, Stitik & Feinberg, 2000; Nicholas & Tyler , 2002; O'Connor , 2004; Tyler, Nicholas, Campell & McHugh, 2001). Many clinicians and sports injury professionals have recognised that the hip musculature plays an essential role, especially in relation to stabilisation of the pelvis (Friel et al., 2006; Gautrey et al., 2013b; Laheru et al., 2007). Strength testing of the hip musculature is often undertaken by sports injury professionals and is of great importance for screening, rehabilitation and injury prevention purposes. Consequently, it is paramount that a reliable method is available for the assessment of the hip abductor and adductor muscles (Laheru et al., 2007).

Injuries of the hip musculature are a common occurrence in sport. Holmich (1998) and Ekstrand and Hilding (1999) reported that the frequency of adductor strains was 8-18% of all injuries in football players. Rugby league have also reported that 10.6% of injuries to players were adductor strains (O'Connor, 2004). These injuries have been linked to muscle weakness, muscle imbalance and a history of previous injury (Nicholas & Tyler, 2002; Tyler et al., 2001). Such injuries may be preventable if the risk factors can be addressed, however, the success of such screening procedures depends on the accuracy and reproducibility of the methods used (Laheru et al., 2007).

Nadler et al. (2000) proposed that screening of hip strength prior to sports performance may be important in the prevention of lower limb injury in athletes. Preseason hip strengthening exercises have been shown to be effective in reducing the incidence of adductor strains (Nicholas & Tyler, 2002). However, such interventions rely on objective and repeatable test protocols.

One of the most commonly used methods for muscle strength testing in the clinical and research environment is isokinetic dynamometry. This is most likely due to the easily repeatable patient setup position, effortless selection of velocity, and simple range of motion (ROM) settings. The reliability of the Biodex System 2 isokinetic dynamometer has been shown to be high, with ICC's $_{(2,1)}$ ranging from 0.92-0.98 for peak torque and 0.88-0.97 for total work (Brown et al., 1993). However, the majority of reliability studies have focused on the knee (Bosquet et al., 2010; Brown et al., 1993; Feiring et al., 1990; Gross et al., 1991; Sole et al., 2007), with only a few focusing on the hip (Burnett, Betts & King, 1990; Emery et al., 1999).

A variety of stabilisation procedures have been reported in the literature when investigating the knee (Arnold & Perrin, 1993; Bohannon & Smith, 1989; Burdett & Van Swearingen, 1987; Johnson & Siegel, 1978; Montgomery, Douglas & Deuster, 1989; Patterson, Nelson & Duncan, 1984). However, limited studies (Laheru et al., 2007) have investigated the effects of stabilising the pelvis during a side lying hip abduction and adduction isokinetic protocol. Laheru et al. (2007) measured side lying hip abduction and adduction on a Cybex dynamometer, and found that reducing pelvic rotation did not enhance reproducibility and did not affect torque production. Laheru et al. (2007) stated that to test the maximal strength of a group of muscles, it is necessary to limit segmental body movement, however, this is very difficult to achieve voluntarily when side lying. It is often found that during abduction and adduction movements the pelvis rotates away from the original set up position (Laheru et al., 2007), which may lead to the recruitment of different muscle groups to facilitate the movement, and may contribute to inaccurate results. From these findings it can be hypothesised that in the present study the stabilised setup (SS) position will not produce increased reliability or

higher peak torque and total work values, in comparison to the non-stabilised setup (NS).

The primary aim of this study was therefore to compare the relative and absolute reliability of two different setup positions (non-stabilised setup (NS) vs. stabilised setup (SS)) during isokinetic hip abduction-adduction exercise, across a velocity spectrum of 60 to 360° · s^{-1}. The secondary aim of this study was to investigate the effect of setup position on peak torque and total work.

7.10.3 Method

7.10.3.1 Subjects

The same subjects were used as in Pilot Study Five (Section 7.4.3.1).

7.10.3.2 Experimental Design

Subject's age, mass and height were recorded. A five minute warm-up was performed on a Monark cycle ergometer (Monark, Varberg, Sweden) at 50rpm with a resistance of 50 Watts. Testing was performed on the Biodex System 2 Isokinetic Dynamometer (Biodex Medical Systems, Shirley, New York). The reliability of the System 2 Biodex dynamometer has been shown to be high, with ICC's $_{(2,1)}$ ranging from 0.92-0.98 for peak torque and 0.88-0.97 for total work (Brown et al., 1993). Taylor et al. (1991) also demonstrated the mechanical validity of the Biodex isokinetic dynamometer in relation to human torque, joint position and limb velocity.

Apparatus Setup

The Biodex was set up according to the Biodex System 2 Manual, and was calibrated according to manufacturer's specifications prior to testing. The cushion control was set to zero, to allow the subject the greatest availability of velocity attainment prior to deceleration (Brown, Whitehurst, Gilbert et al., 1995; Taylor et al., 1991). All subjects completed a practice session on the isokinetic dynamometer a week prior to the main testing procedure.

Non-stabilised Setup (NS) (Biodex)

Subjects were positioned in accordance with manufacturer's instructions. Subjects were side lying, facing away from the dynamometer power head. The right hip was superior with the knee fully extended, and the left knee was flexed to 90°. The axis of the dynamometer was aligned superior and medial to the greater trochanter of the right leg. The subject's right leg was attached to the Biodex hip attachment, superior to the lateral joint line of the knee. The subject's top hand grasped the border of the chair (Figure 7.16 and 7.18).

Stabilised Setup (SS) (Modified Biodex)

This setup was developed to try and restrict rotational movement of the subject. The subject was positioned identical to the NS, however, three additional straps at the level of the thigh, pelvis and chest were attached to the subject. Each strap was looped around the chair base, and tightened to secure the subject in place. The thigh strap was fastened at the midpoint between the medial joint line of the knee and the anterior superior iliac spine (ASIS). The pelvis strap was applied at the level of the ASIS, and the chest strap was fastened at the level of the nipples (Figure 7.18).

Testing Protocol

The subject's range of motion (ROM) was set at 0-45° of abduction. The ROM was based on the average limitations of hip abduction motion in healthy individuals (Emery et al., 1999; Makofsky et al., 2007; Reid, 1992). Warm-up on the isokinetic device consisted of three sub maximal reciprocal concentric abduction and adduction repetitions with increasing intensity (i.e. first repetition at 25% perceived effort, second repetition at 50% perceived effort, and third repetition at 75% perceived effort) (Brown, Whitehurst, Gilbert et al., 1995), at $60° \cdot s^{-1}$ through $360° \cdot s^{-1}$ (Brown et al., 1993; Timm & Fyke, 1993). In addition the subject completed two maximal intensity repetitions at each speed (Brown, Whitehurst, Gilbert et al., 1995; Findley et al., 2006).

Testing began with the subject's leg at 0° of abduction and consisted of three maximal concentric reciprocal hip abduction and adduction gravity corrected repetitions in a fixed order at 60, 120, 180, 240, 300 and $360° \cdot s^{-1}$ (Gautrey et al., 2013b), with a 30 second rest between velocities (Timm & Fyke, 1993). Each subject was encouraged to contact the mechanical end stops during both abduction and adduction movements. The same verbal encouragement was given to each subject throughout the test to motivate them to develop maximal torque during each repetition (McNair et al., 1996) but no visual feedback of torque generation was provided.

Subjects were randomised to the setup they would undertake first (NS or SS), and following a 24 hour rest period subjects returned to the laboratory to complete the remaining setup. Subject's then returned to the laboratory 7 days later for repeat testing, identical to the first week.

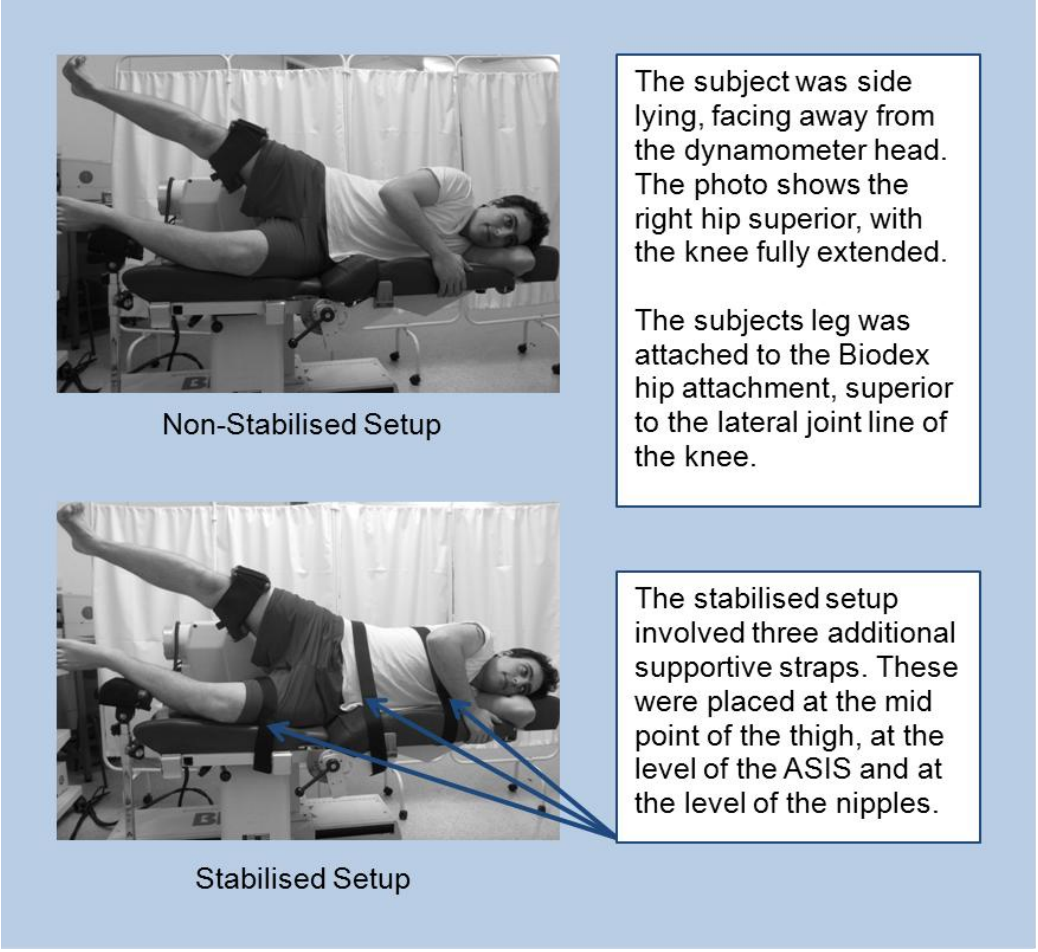


Figure 7.18. Isokinetic Hip Abduction-Adduction Setup Positions.

7.10.3.3 Data Analysis

The same data analysis was used as in Pilot Study Five (Section 7.4.3.3).

7.10.3.4 Statistical Analysis

Using SPSS (version 19) normal Gaussian distribution of the data was verified by the Shapiro-Wilk test. Systematic bias, which refers to a difference in measurements in a particular direction between repeated tests, was assessed with four (peak torque [hip abduction and hip adduction] and total work [hip abduction and hip adduction]) 2 x 6 x 2

(setup position [NS or SS] x speed [60, 120, 180, 240, 300 and 360° · s^{-1}] × time [first week testing or second week testing]) repeated measures analysis of variance (ANOVA). Sphericity was verified for all data being compared by the Mauchly test. The Multivariate Test box (Wilk's Lambada value) was studied for three-way interactions, then two-way interactions and then main effects (P<0.05). The Pairwise Comparisons post-hoc test was used to determine exactly where the significant findings occurred when there were more than two conditions (speed). Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.008.

Relative reliability was assessed by calculating the $ICC_{(2,1)}$. From the ICC value the SEM was calculated, which represented absolute reliability (Section 3.2.3.4, paragraph 2).

7.10.4 Results

7.10.4.1 Peak Torque

The peak torque relative reliability results for hip abduction (Table 7.10) and hip adduction (Table 7.11) during the two setup positions showed that the SS was the most reliable setup. Currier (1990) suggested that an ICC value > 0.80 was acceptable for clinical work, and therefore speeds 60 through 240° · s^{-1} would be adequate during the SS. When observing the hip abduction results (Table 7.10) for the NS only speeds 60 and 120° · s^{-1} produced ICC's above the acceptable level (>0.80). When observing the hip adduction results (Table 7.11) for the NS only speeds 60 through 180° · s^{-1} produced ICC's above the acceptable level (>0.80).

When observing the SEM results for hip abduction (Table 7.10) and hip adduction (Table 7.11) the results show that the SS had the lowest, and therefore the least variable results. It has been stated that SEM values below 10% are an acceptable level of variance. When observing the hip abduction results (Table 7.10) the SS for speeds 60 through 240° · s^{-1} all had SEM values below 10% (range: 3.9-9.8%) and therefore all fall within the recommended level of variance. When observing the hip adduction results (Table 7.11) the SS for speeds 60 through 360° · s^{-1} all had SEM values below 10% (range: 3.1-8.6%) and therefore all fall within the recommended level of variance. When observing the NS only speeds 60 through 180° · s^{-1} produced SEM values below the recommended 10% threshold, for both hip abduction and hip adduction. The results also highlight that during the two setup positions, peak torque relative and absolute reliability decreased with each increase in velocity (Table 7.10 and 7.11). It can therefore be seen that the SS was the most reliable setup position, with the highest ICC results and the lowest SEM variance during both hip abduction and hip adduction.

Results from the repeated measures ANOVA showed significantly greater (P<0.05) peak torque values for the SS during both hip abduction (Figure 7.19) and hip adduction (Figure 7.20) for speeds 60, 120, 180 and 240° · s^{-1}. The two fastest speeds of 300 and 360° · s^{-1} still showed that the SS had a greater peak torque value, but the results were not statistically significant. The results therefore show that the SS enabled the greatest peak torque values to be produced. The repeated measures ANOVA also showed a significant decrease (P<0.008) in normalised peak torque values with each increase in velocity (Table 7.10 and 7.11).

Table 7.10. Normalised Peak Torque Values for Hip Abduction During the Nonstabilised Setup and the Stabilised Setup.

VELOCITY ($° \cdot s^{-1}$)	NONSTABILISED SETUP - BIODEX				STABILISED SETUP - MODIFIED BIODEX			
	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)
60	0.90 (0.09)	0.92 (0.11)	0.91	4.3	1.02 (0.10)	0.98 (0.08)	0.95	3.9
120	0.78 (0.11) *	0.76 (0.10) *	0.86	6.4	0.84 (0.10) *	0.85 (0.11) *	0.89	6.1
180	0.59 (0.09) *	0.61 (0.07) *	0.78	9.8	0.67 (0.08) *	0.66 (0.12) *	0.84	8.9
240	0.40 (0.08) *	0.44 (0.11) *	0.71	11.3	0.46 (0.10) *	0.50 (0.09) *	0.83	9.8
300	0.28 (0.08) *	0.24 (0.09) *	0.50	14.8	0.29 (0.08) *	0.31 (0.07) *	0.60	11.9
360	0.14 (0.04) *	0.16 (0.05) *	0.43	16.7	0.18 (0.05) *	0.19 (0.04) *	0.53	11.5

Data are presented as Mean (SD). * Significantly different (P<0.008) from previous velocity

Table 7.11. Normalised Peak Torque Values for Hip Adduction During the Nonstabilised Setup and the Stabilised Setup.

VELOCITY ($° \cdot s^{-1}$)	NONSTABILISED SETUP - BIODEX				STABILISED SETUP - MODIFIED BIODEX			
	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)
60	1.06 (0.10)	1.12 (0.13)	0.90	4.5	1.25 (0.12)	1.27 (0.09)	0.94	3.1
120	0.89 (0.13) *	0.92 (0.11) *	0.87	4.3	0.96 (0.11) *	0.99 (0.13) *	0.93	4.0
180	0.77 (0.10) *	0.79 (0.08) *	0.82	6.3	0.82 (0.09) *	0.85 (0.10) *	0.88	5.8
240	0.56 (0.09) *	0.58 (0.10) *	0.75	10.3	0.64 (0.09) *	0.62 (0.08) *	0.81	7.8
300	0.35 (0.11) *	0.39 (0.08) *	0.51	10.2	0.41 (0.11) *	0.43 (0.10) *	0.59	6.9
360	0.17 (0.06) *	0.19 (0.07) *	0.48	13.7	0.21 (0.10) *	0.23 (0.07) *	0.55	8.6

Data are presented as mean (SD). * Significantly different (P<0.008) from previous velocity

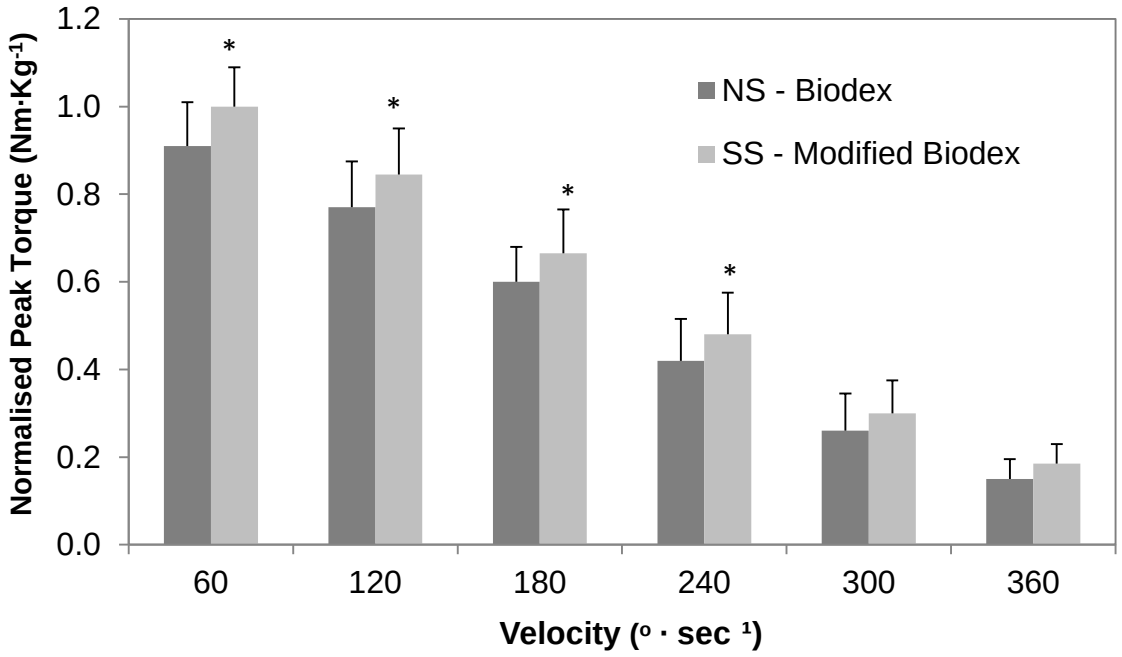


Figure 7.19. Normalised Peak Torque for Hip Abduction During the Nonstabilised Setup (NS) and the Stabilised Setup (SS) Across a Velocity Spectrum (Mean ± SD). * SS significantly (*P*<0.05) higher than NS.

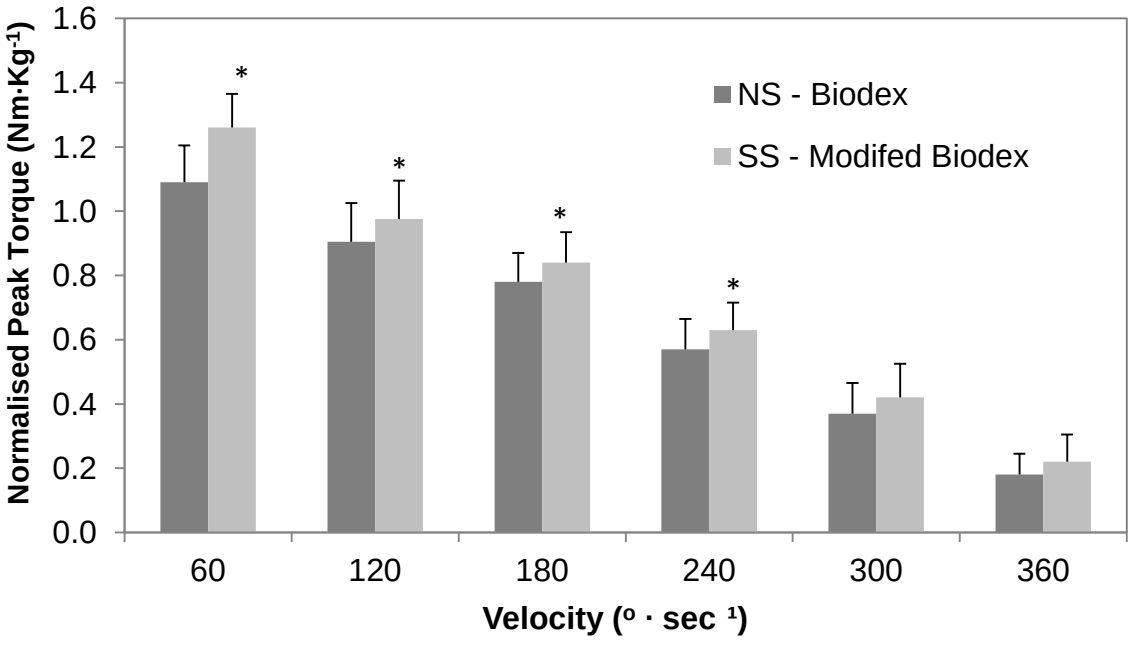


Figure 7.20. Normalised Peak Torque for Hip Adduction During the Nonstabilised Setup (NS) and the Stabilised Setup (SS) Across a Velocity Spectrum (Mean ± SD). * SS significantly (*P*<0.05) higher than NS.

7.10.4.2 Total Work

The total work relative reliability results for hip abduction (Table 7.12) and hip adduction (Table 7.13) during the two setup positions showed that the SS was the most reliable setup. Currier (1990) suggested that an ICC value > 0.80 was acceptable for clinical work, and therefore speeds 60 through 240° · s^{-1} would be adequate during the SS. When observing the hip abduction (Table 7.12) and hip adduction (Table 7.13) results for the NS only speeds 60 through 180° · s^{-1} produced ICC's above the acceptable level (>0.80).

When observing the SEM results for hip abduction (Table 7.12) and hip adduction (Table 7.13) the results show that the SS had the lowest, and therefore the least variable results. As previously mentioned, 10% has been stated as an acceptable SEM value. The SS for speeds 60 through 240° · s^{-1} all had SEM values below 10% (range: 5.6-10.0%), for both hip abduction and hip adduction, and therefore all fall within the recommended level of variance. When observing the NS only speeds 60 through 180° · s^{-1} produced SEM values below the recommended 10% threshold, for both hip abduction and hip adduction. The results also highlight that during the two setup positions, total work relative and absolute reliability decreased with each increase in velocity (Table 7.12 and 7.13). It can therefore be seen that the SS was the most reliable setup position, with the highest ICC results and the lowest SEM variance during both hip abduction and hip adduction.

Results from the repeated measures ANOVA showed significantly greater (P<0.05) total work values during hip abduction (Figure 7.21) for the SS across all velocities.

Significantly greater (P<0.05) results were also demonstrated for the SS during hip adduction (Figure 7.22) for speeds 60, 120 and 180° · s^{-1}. The three faster speeds of 240, 300 and 360° · s^{-1} showed that the SS had a greater total work value, but the results were not statistically significant. The results therefore show that the SS enabled the greatest total work values to be produced. The repeated measures ANOVA also showed a significant decrease (P<0.008) in normalised total work values with each increase in velocity (Table 7.12 and 7.13).

Table 7.12. Normalised Total Work Values for Hip Abduction During the Nonstabilised Setup and the Stabilised Setup.

VELOCITY ($° \cdot s^{-1}$)	NONSTABILISED SETUP - BIODEX				STABILISED SETUP - MODIFIED BIODEX			
	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)
60	0.77 (0.09)	0.74 (0.10)	0.92	6.7	0.80 (0.11)	0.82 (0.09)	0.95	6.1
120	0.70 (0.10) *	0.71 (0.09) *	0.90	7.0	0.75 (0.09) *	0.78 (0.08) *	0.93	6.6
180	0.55 (0.09) *	0.54 (0.09) *	0.81	9.0	0.58 (0.09) *	0.59 (0.07) *	0.84	8.3
240	0.30 (0.08) *	0.32 (0.09) *	0.74	15.6	0.38 (0.08) *	0.40 (0.08) *	0.82	10.0
300	0.19 (0.06) *	0.21 (0.05) *	0.54	15.1	0.24 (0.05) *	0.26 (0.07) *	0.59	13.8
360	0.10 (0.03) *	0.13 (0.04) *	0.42	14.3	0.16 (0.02) *	0.18 (0.04) *	0.50	14.1

Data are presented as Mean (SD). * Significantly different (P<0.008) from previous velocity

Table 7.13. Normalised total work values for hip adduction during the nonstabilised setup and the stabilised setup.

VELOCITY ($° \cdot s^{-1}$)	NONSTABILISED SETUP - BIODEX				STABILISED SETUP - MODIFIED BIODEX			
	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)	TEST 1 ($Nm \cdot Kg^{-1}$)	TEST 2 ($Nm \cdot Kg^{-1}$)	ICC	SEM (%)
60	0.82 (0.08)	0.84 (0.11)	0.91	6.0	0.90 (0.10)	0.87 (0.08)	0.94	5.6
120	0.77 (0.09) *	0.79 (0.08) *	0.92	7.1	0.85 (0.10) *	0.83 (0.09) *	0.92	7.0
180	0.60 (0.08) *	0.64 (0.10) *	0.83	8.1	0.69 (0.08) *	0.67 (0.08) *	0.86	5.8
240	0.39 (0.09) *	0.40 (0.08) *	0.75	10.7	0.40 (0.08) *	0.43 (0.07) *	0.81	9.8
300	0.24 (0.07) *	0.26 (0.06) *	0.56	14.4	0.27 (0.05) *	0.27 (0.06) *	0.61	10.5
360	0.15 (0.04) *	0.15 (0.03) *	0.44	13.2	0.16 (0.03) *	0.18 (0.02) *	0.52	11.1

Data are presented as mean (SD). * Significantly different (P<0.008) from previous velocity.

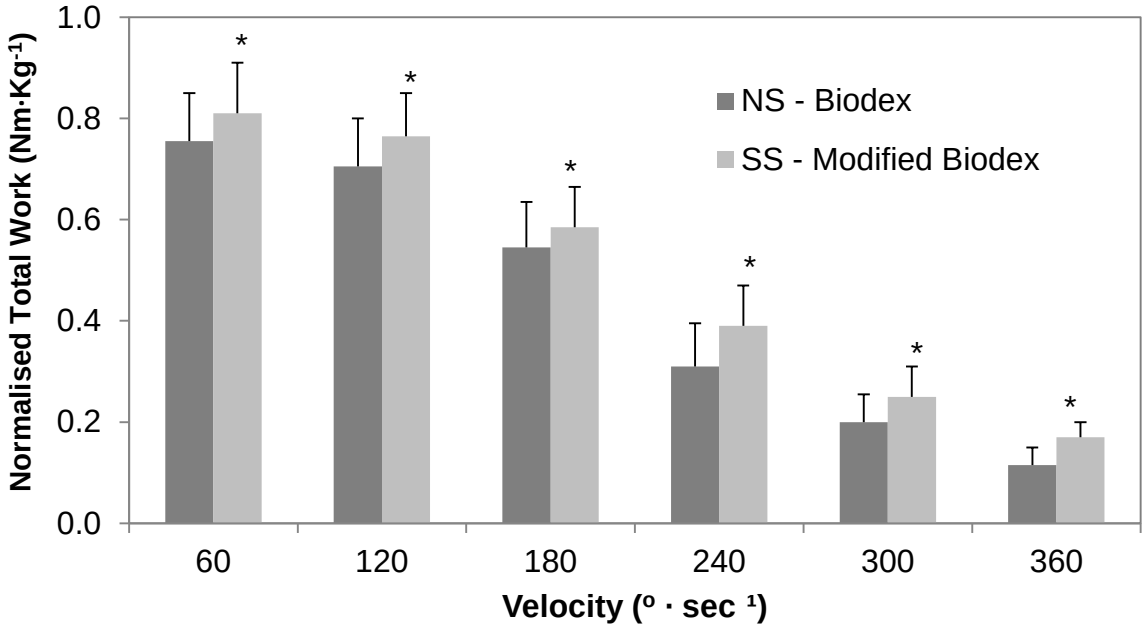


Figure 7.21. Normalised Total Work for Hip Abduction During the Nonstabilised Setup (NS) and the Stabilised Setup (SS) Across a Velocity Spectrum (Mean ± SD). * SS significantly (*P*<0.05) higher than NS.

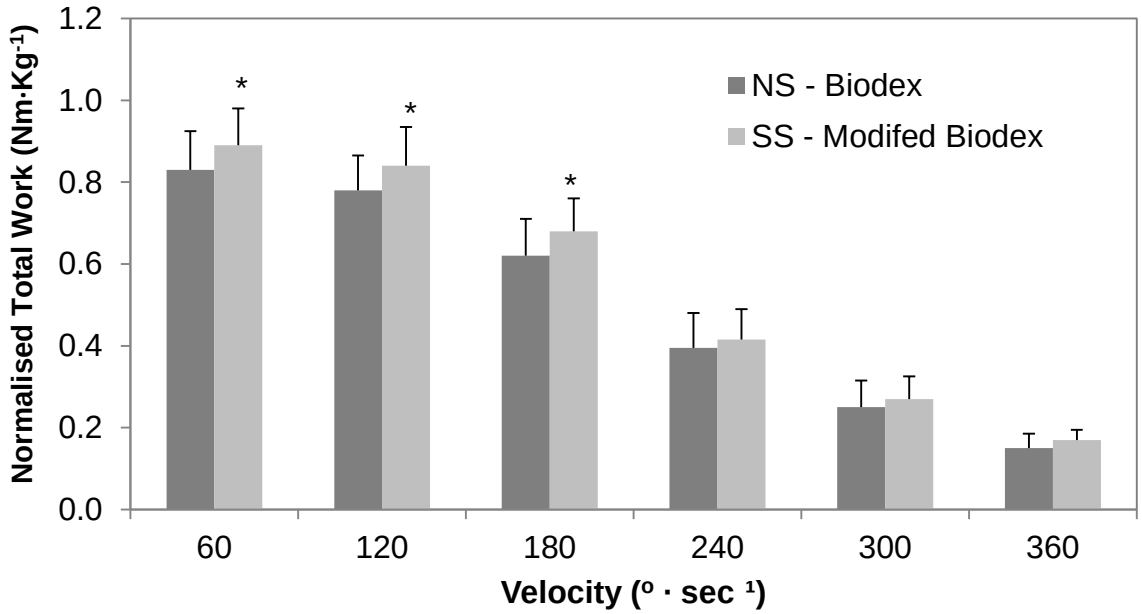


Figure 7.22. Normalised Total Work for Hip Adduction During the Nonstabilised Setup (NS) and the Stabilised Setup (SS) Across a Velocity Spectrum (Mean ± SD). * SS significantly (*P*<0.05) higher than NS.

7.10.5 Discussion

The primary aim of this study was to compare the relative and absolute reliability of two different setup positions (non-stabilised setup (NS) vs. stabilised setup (SS)) during isokinetic hip abductor-adductor exercise. The results highlighted that the SS was the most reliable setup position, with the highest ICC results and the lowest SEM variance during ankle inversion and ankle eversion. The secondary aim of this study was to investigate the effect of setup position on the magnitude of peak torque and total work. From the results it was apparent that there were significantly greater (P<0.05) peak torque values for the SS during both hip abduction and hip adduction for speeds 60, 120, 180 and 240° · s^{-1}. There were also significantly greater (P<0.05) total work values during hip abduction for the SS across all velocities. Significantly greater (P<0.05) results were also demonstrated for the SS during hip adduction for speeds 60, 120 and 180° · s^{-1}. Therefore, the hypothesis that the stabilised setup (SS) position will not produce increased reliability or higher peak torque and total work values, in comparison to the non-stabilised setup (NS) can be formally rejected.

7.10.5.1 Peak Torque and Total Work Reliability

The peak torque and total work relative reliability results for hip abduction and adduction during the two setup positions show that the SS was the most reliable setup. Speeds 60 through 240° · s^{-1} demonstrated ICC values > 0.80, which has been suggested as acceptable for clinical work (Currier, 1990). Previous authors have found that isokinetic testing of the hip abductors and hip adductors has been associated with only moderate repeatability (Burnett et al., 1990; Emery et al., 1999). One causal factor that has been

related to this moderate repeatability is excess pelvic rotation (Laheru et al., 2007). No study to date has investigated the effect of increased stabilisation on reliability values during hip abduction-adduction exercise. The present study found that using three additional straps during the patient setup on the isokinetic dynamometer, led to an improvement in the relative and absolute reliability values.

The current study is the only one obtainable that has tested through a velocity spectrum ranging from 60 to 360° · s^{-1}. The results from the present study found that with each increase in velocity, there was a decrease in reliability (Tables 7.10 to 7.13). This is possibly due to the subjects finding it more difficult to obtain the higher velocities, as shown by the inverse relationship between load range and increased velocity (Feiring et al., 1990). As the velocity of the dynamometer increases, the subject finds it more difficult to achieve this velocity, as a result of this the peak torque and total work values become more variable, and therefore lower reliability values are produced. The slower speeds of 60 through 240° · s^{-1} all demonstrate ICC's above 0.80, however speeds 300 and 360° · s^{-1} only show moderate reliability. These results may suggest that researchers, clinicians and sports injury professionals should opt for a speed between 60 and 240° · s^{-1} if they are conducting repeated tests, and require a reliable protocol.

The accuracy to which these protocols are reproducible is also a critical factor as determined by the SEM. The SEM value in this study was expressed as a percentage in order to allow clinical usage of these measures. As demonstrated by the results of the current study, re-test values for peak torque and total work, for the NS and the SS varied by 3.1 – 16.6% to the initial test. It should therefore, seem appropriate in future

studies to attribute differences in isokinetic results to intervention, training improvements or injury, should they exceed the SEM values outlined in tables 7.10 to 7.13.

The relevance of the reliability findings in the present study lies predominantly in the research field. It may be argued that the increase in reliability is marginal between the SS and the NS. For example, when observing the hip abductor peak torque results at $60° \cdot s^{-1}$ (Table 7.10) the relative reliability improved from 0.91 with the NS to 0.95 with the SS. The SEM variance was also improved from 4.3% with the NS to 3.9% with the SS. These changes may seem small, but in the field of research where reliable protocols are a necessity, the use of three simple, and easy to apply additional straps during the SS, improved the reliability of the protocol.

Another important difference between the setup positions, is that the SS peak torque results show reliable measures for speeds 60 through $240° \cdot s^{-1}$ for both hip abduction and adduction. However, the NS only show reliable peak torque results for speeds 60 and $120° \cdot s^{-1}$ for abduction, and 60 through $180° \cdot s^{-1}$ for adduction. For the total work results the SS show reliable results for speeds 60 through $240° \cdot s^{-1}$ for both hip abduction and adduction. However, the NS only show reliable total work values for speeds 60 through $180° \cdot s^{-1}$ for hip abduction and adduction. If researchers, clinicians or sports injury professionals wish to test patients at velocities between 180 and $240° \cdot s^{-1}$, they should opt for the SS as this notably increased the reliability in comparison to the NS.

7.10.5.2 Peak Torque and Total Work Magnitude

The second main finding of the present study was that increased stabilisation (SS) led to an increase in peak torque and total work measures during both hip abduction and hip adduction. Several other studies have investigated the effect of stabilising body segments on peak torque and total work (Kovaleski et al., 1995; Lentell et al., 1990; Porter & Kaminski, 2004; Weir, 2005). Hart, Stobbe, Till and Plummer (1984) reported increased knee extension torque with trunk stabilisation. However, in contrast to this, Hanten and Ramberg (1988) found no significant differences in concentric and eccentric torque between a stabilisation protocol that included both trunk and pelvis stabilisation vs. simply holding onto the sides of the testing table. Patterson et al. (1984) also found no significant differences with a stabilised vs. nonstabilised test on isokinetic knee extension and flexion torque.

Limited studies (Porter & Kaminski, 2004) have investigated the effects of stabilising the pelvis during a side lying hip abduction and adduction isokinetic protocol. One study was found by Laheru et al. (2007) that measured hip abduction and adduction on a Cybex dynamometer, and found that reducing pelvic rotation did not enhance reproducibility and did not affect torque production. The results from the present study challenge the results of Laheru et al. (2007) as an increase in both peak torque and total work during the SS during hip abduction and hip adduction were found.

The results from the present study may show that the increased stabilisation, led to a decrease in pelvic rotation, which then facilitated the generation of greater peak torque measurements. Pelvic rotation was not objectively measured in the present study, but

on subjective observation it could be seen that there was less movement around the pelvis with the SS. The subjects also generally reported feeling more secure with the three additional straps. Leheru et al. (2004) used a different method of stabilisation in comparison to the present study, and even though they found no significant increase in peak torque, they did measure pelvic rotation during hip abduction and adduction at 60° $\cdot$ s^{-1}. Leheru et al. (2004) reported that the stabilised condition showed an overall mean reduction from 22.3° to 14.8°, in pelvic rotation in the transverse plane, however this did not reach statistical significance (Laheru et al., 2007). Future studies investigating the effect of pelvic stabilisation during hip abduction and adduction should include measures of pelvic rotation, so it can be determined if a relationship exists between reduced pelvic motion and increased torque measurements.

7.10.5.3 Clinical Implications

In the present study the SS led to an increase in reliability and magnitude of peak torque and total work measurements, therefore, clinicians in the future should opt for this setup over the NS. The results also indicate that the SS is the strongest setup position for the hip abductors and adductors, and this may have implications for therapeutic and testing protocols. For example, facilitation of very weak hip abductors and adductors might be better accomplished with the SS position.

7.10.5.4 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. It must be

remembered that the results are only applicable if the same equipment and protocol is used as in the current study. Future studies may wish to repeat this study but using different makes of isokinetic dynamometers and varied protocols.

7.10.6 Conclusion

It is often found that during side lying isokinetic hip abduction and adduction the pelvis rotates away from the original set up position, which may lead to the recruitment of different muscle groups to assist the movement, and contribute towards inaccurate results. In the present study, the SS led to an increase in reliability and magnitude of peak torque and total work measurements. However, speeds 300 and 360° $\cdot$ s^{-1} only showed moderate reliability during the SS. The results suggest that researchers, clinicians and sports injury professionals should opt for the SS at speeds between 60 and 240° $\cdot$ s^{-1} if they are conducting repeated tests, and require a reliable protocol.

7.11 Development of Research

Pilot Study Eight addressed the issue of identifying the most reliable hip setup position to be used in Study Three and Four. The results indicated that the addition of three extra stabilisation straps increased the reliability of peak torque measures, and also enabled the subject to produce significantly higher peak torque values. Therefore, this setup position was deemed appropriate for Study Three and Four.

7.12 Pilot Study Nine: The Effect of Velocity on Load Range during Isokinetic Hip Abduction-Adduction Exercise

7.12 1 Abstract

Aim: To quantify the components of acceleration, load range and deceleration through a velocity spectrum during concentric hip abduction and adduction isokinetic exercise, and to investigate the effect of load range on peak torque and work done. **Method:** Sixteen male healthy subjects performed three maximal concentric reciprocal hip abduction and adduction gravity corrected repetitions in a fixed order at 60, 120, 180, 240, 300 and 360° · s^{-1}, with a 30 second rest between velocities. **Results:** Hip abduction and adduction results revealed that load range significantly decreased while acceleration and deceleration ROM significantly increased (P<0.008) with each increase in velocity. When the total peak torque data was corrected for load range there was a significant decrease (P<0.05) in peak torque at velocities of 300° · s^{-1} and above, for both hip abduction and adduction. Load range correction also resulted in a significant decrease (P<0.05) in work done at velocities of 120° · s^{-1} and above, for both hip abduction and adduction. **Conclusion:** The results demonstrate an inverse relationship between isokinetic velocity and load range during concentric hip abduction and adduction, and suggest a need for the clinician to carefully consider velocity selection when performing exercise on an isokinetic device.

7.12.2 Introduction

The isokinetic dynamometer has commonly been used for rehabilitation or training purposes (Brown et al., 2003; Hamdoun-Kahlaoui et al., 2010; Hammami et al., 2012; Murray et al., 2007; Nickols-Richardson et al., 2007; Osternig, 2000). As the isokinetic dynamometer only offers resistance once the pre-set velocity is attained, any strength gains achieved from isokinetic exercise may be proportional to the total amount of range of motion (ROM) actually sustained at the pre-set isokinetic velocity (Brown, Whitehurst, Findley et al., 1995). It is therefore of great interest to investigate what percentage of the ROM of a concentric action is actually spent at the pre-selected velocity, over a velocity spectrum.

A concentric action performed on an isokinetic device involves three main components: acceleration, sustained velocity, and deceleration (Brown, Whitehurst, Gilbert et al., 1995; Osternig, 1975; Taylor et al., 1991). The acceleration component has been defined as the individual's ability to "catch" the dynamometer (Davies, 1992; Osternig, 1986). The "catch" phase is completed once the individual attains the pre-set velocity, and the resistance is met, which then prevents any further acceleration (Davies, 1992; Osternig, 1986). The sustained velocity component of the repetition has also been termed load range (Brown, Whitehurst, Gilbert et al., 1995; Findley et al., 2006; Kurdak et al., 2005). To be more precise the concept of load range has been described as external machine resistance encountered through a pre-set sustained velocity within a defined range of motion (ROM) (Brown, Whitehurst, Findley et al., 1995). The final component, mechanical deceleration, offers resistance while the isokinetic dynamometer decreases speed at the end of the defined ROM. However, Brown,

Whitehurst, Gilbert et al. (1995) has argued that this phase is neither directly governed by the tester nor quantifiable as torque produced under controlled isokinetic conditions, and therefore ceases to be isokinetic (Brown, Whitehurst, Gilbert et al., 1995).

Earlier research has shown that torque patterns are significantly affected when the load range phase of the motion is taken into consideration (Brown & Whitehurst, 2000; Brown, Whitehurst, Gilbert et al., 1995; Kovaleski et al., 1995). In short, this means that actual torque may differ by a large magnitude if evaluated outside of the load range (Findley et al., 2006). Kurdak et al. (2005) found a significant decrease when comparing load range peak torque to total peak torque at speeds above $270° \cdot s^{-1}$ for knee extension and above $300° \cdot s^{-1}$ for knee flexion. The authors also found a significant decrease when comparing load range work and total work at speeds above $90° \cdot s^{-1}$ for both knee extension and knee flexion. These results highlight the importance of correcting the data for load range as it is apparent that large errors can occur if this process is not undertaken.

Increased angular velocity results in a reduction in load range, thus data from the measurements that were performed at higher angular velocities may not actually reflect load range values (Kurdak et al., 2005). This is in agreement with the classic force – velocity curve, which explains the relationship between skeletal muscle contraction velocity and torque production (Widrick et al., 1996): as velocity increases, torque decreases (Brown & Whitehurst, 2000). Therefore extra caution is required to make correct interpretation of isokinetic results (Brown & Whitehurst, 2000).

Load range has been investigated previously, however, only during unilateral knee flexion/extension (Brown, Whitehurst, Gilbert et al., 1995; Osternig, 1986; Taylor et al., 1991; Wilk et al., 1994), bilateral knee flexion/extension (Scibelli et al., 1993) and shoulder external/internal rotation (Brown, Whitehurst, Findley et al., 1995). Each study found an inverse relationship between load range and velocity, yet the primary focus of these studies was load range, apart from Brown, Whitehurst, Gilbert et al. (1995) who also considered the impact of the acceleration and deceleration components. Therefore, quantification of each component may lead to a more complete understanding of load range magnitude and position within the exercised ROM. This information may better equip the clinician in more accurate velocity prescription during isokinetic exercise. From the findings of previous literature it can be hypothesised that with each increase in velocity there will be a decrease in the load range component, and an increase in the acceleration and deceleration components. It was also hypothesised that load range corrected peak torque and total work data will be significantly different to the uncorrected data at higher velocities.

Recently the investigation into the musculature around the hip has become of interest, especially in regards to patients with a history of FAI (Gautrey et al., 2013a). It has been suggested that patients with a history of FAI may have a weakness in muscles surrounding the hip, primarily the gluteus medius, which results in a more adducted foot placement during the gait cycle (Friel et al., 2006; O'Dwyer, Sainsbury & O'Sullivan, 2011). This adducted foot position results in an increased chance of the individual contacting the floor with the lateral aspect of the foot, which could potentially lead to an increased chance of 'rolling over' on the ankle and sustaining a lateral ankle sprain (Friel et al., 2006). O'Dwyer et al. (2011) stated that dysfunction of the gluteus medius is

commonly implicated in lower limb pathologies. It has been stated that the gluteus medius muscle should be evaluated in healthy participants, to try and identify individuals with a possible pre-disposition to ankle sprains (Beckman & Buchanan, 1995).

The primary aim of this study was therefore to quantify the components of load range, acceleration, and deceleration through a velocity spectrum during concentric hip abduction and adduction isokinetic exercise. The secondary aim of this study was to investigate the effect of load range on peak torque and work done.

7.12.3 Method

7.12.3.1 Subjects

The same subjects were used as in Pilot Study Five (Section 7.4.3.1).

7.12.3.2 Experimental Design

The same experimental design was used as in Pilot Study Eight (Section 7.10.3.2); apart from only the SS was used, and subjects were not required to return to the laboratory seven days later to repeat the procedure.

7.12.3.3 Data Analysis

The same data analysis was used as in Pilot Study Six (Section 7.6.3.3); apart from results were determined for hip abduction and adduction, instead of ankle inversion and

eversion. Load range, acceleration and deceleration were determined for hip abduction and adduction (Figures 7.23 and 7.24).

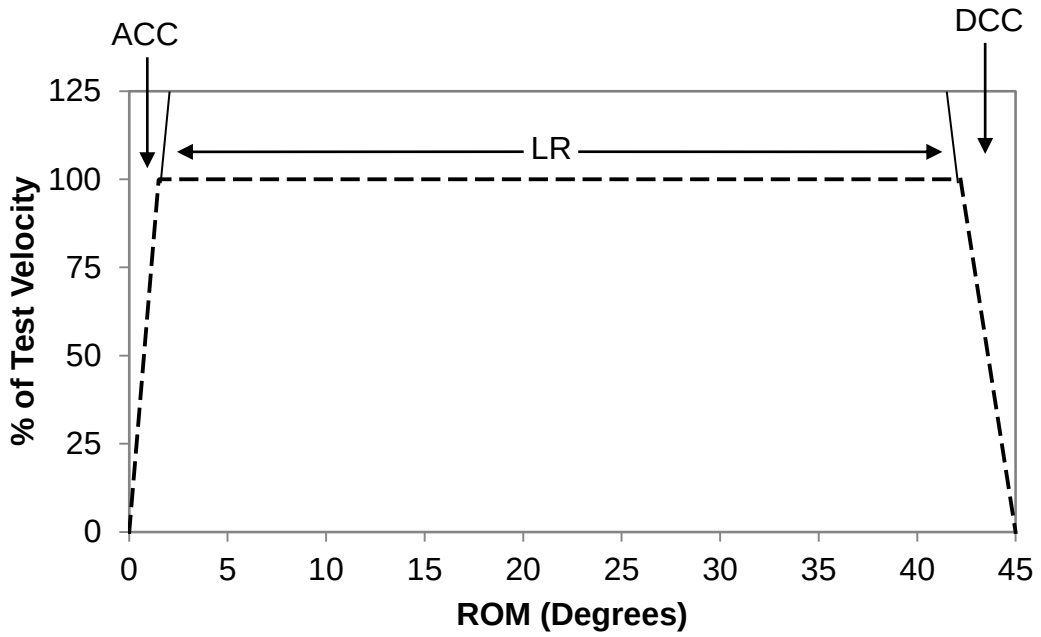


Figure 7.23. Example of a Mean Velocity Tracing at 60° · s^{-1} Showing Acceleration (ACC), Load Range (LR), and Deceleration (DCC) Range of Motion (ROM) During Hip Abduction.

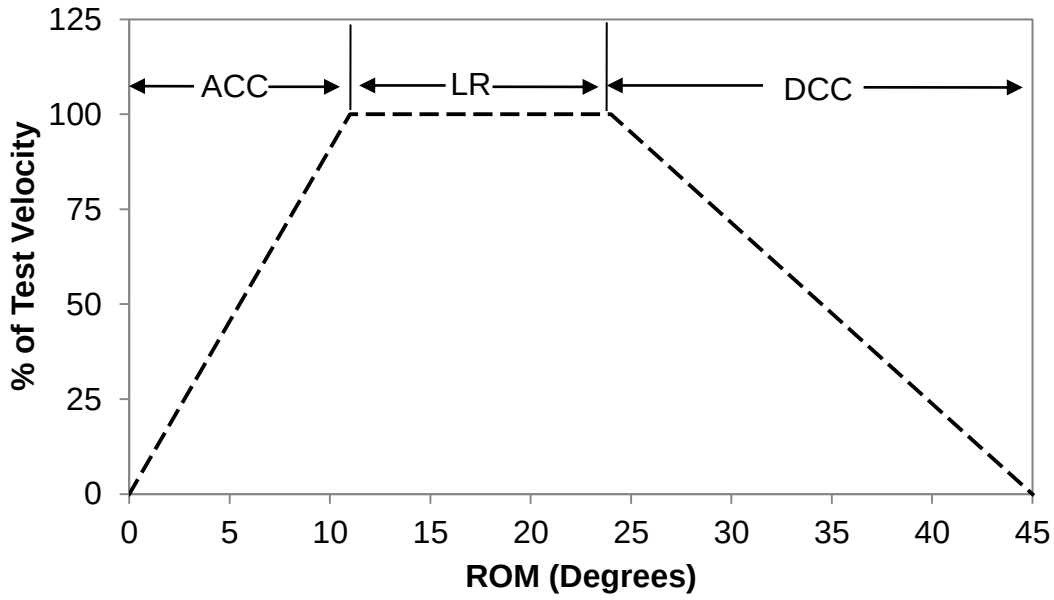


Figure 7.24. Example of a Mean Velocity Tracing at 360° · s^{-1} Showing Acceleration (ACC), Load Range (LR), and Deceleration (DCC) Range of Motion (ROM) During Hip Abduction.

7.12.3.4 Statistical Analysis

Using SPSS (version 19) a 6 x 2 (speed [60, 120, 180, 240, 300 and 360° · s^{-1}] x movement [hip abduction and hip adduction]) repeated measures analysis of variance (ANOVA) was performed for the acceleration, load range and deceleration data. Sphericity was verified for all data being compared by the Mauchly test. The Multivariate Test box (Wilk's Lambada value) was studied for two-way interactions and then main effects (P<0.05). The Pairwise Comparisons post-hoc test was used to determine exactly where the significant findings occurred when there were more than two conditions (speed). Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.008.

A 6 x 2 x 2 (speed [60, 120, 180, 240, 300 and 360° · s^{-1}] x analysis type [total values and load range values] x movement [hip abduction and hip adduction]) repeated measures ANOVA was performed for peak torque and work data. Sphericity was verified for all data being compared by the Mauchly test. The Multivariate Test box (Wilk's Lambada value) was studied for three-way interactions, then two-way interactions and then main effects (P<0.05). The Pairwise Comparisons post-hoc test was used to determine exactly where the significant findings occurred when there were more than two conditions (speed). Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at P<0.008.

7.12.4 Results

The 6 x 2 (speed x movement) repeated measures ANOVA results revealed that load range significantly decreased while acceleration and deceleration ROM significantly increased with each increase in velocity, for both hip abduction and hip adduction (Table 7.14). There was no significant difference found between abduction and adduction results. The amount of ROM spent in load range significantly decreased from 41.8˚ to 21.7˚ for abduction, and from 42.2˚ to 22.2˚ for adduction, at 60 through 360° · s^{-1}. The amount of ROM spent in acceleration significantly increased from 1.1˚ to 11.1˚ for abduction, and from 0.9˚ to 10.7˚ for adduction, at 60 through 360° · s^{-1}. The amount of ROM spent in deceleration significantly increased from 2.1˚ to 12.2˚ for abduction, and from 1.9˚ to 12.1˚ for adduction, at 60 through 360° · s^{-1}. Observing the results as a percentage of the total test ROM the abduction load range (Figure 7.25) significantly decreased from 92.9% to 48.2%, and adduction load range (Figure 7.26) significantly decreased from 93.8% to 49.3%, at 60 through 360° · s^{-1} respectively.

Table 7.14. Hip Abduction and Adduction Acceleration, Load Range and Deceleration Range of Motion Across Velocities.

Velocity ($° \cdot s^{-1}$)	Acceleration (Degrees)	Load Range (Degrees)	Deceleration (Degrees)
Abduction			
60	1.1 (0.2)	41.8 (0.3)	2.1 (0.1)
120	2.0 (0.2) *	39.0 (0.3) *	4.0 (0.2) *
180	3.2 (0.5) *	35.7 (0.5) *	6.1 (0.1) *
240	4.1 (0.8) *	32.7 (0.9) *	8.2 (0.4) *
300	7.6 (0.8) *	27.3 (1.2) *	10.1 (0.7) *
360	11.1 (1.0) *	21.7 (1.9) *	12.2 (0.7) *
Adduction			
60	0.9 (0.1)	42.2 (0.2)	1.9 (0.1)
120	1.9 (0.2) *	39.2 (0.4) *	3.9 (0.1) *
180	3.0 (0.4) *	36.1 (0.4) *	5.9 (0.2) *
240	4.0 (0.4) *	33.0 (0.7) *	8.0 (0.3) *
300	7.2 (0.8) *	27.9 (1.0) *	9.9 (0.7) *
360	10.7 (0.8) *	22.2(1.7) *	12.1 (0.7) *

Data are presented as mean (SD).* Significantly (P<0.008) different from previous velocity

The 6 x 2 x 2 (speed x analysis type x movement) repeated measures ANOVA results revealed that normalised total peak torque (Figures 7.27 and 7.28) values significantly decreased with each increase in velocity for both hip abduction and hip adduction. There was no significant difference found between hip abduction and adduction results. The normalised total peak torque values significantly decreased from 1.2 $Nm \cdot Kg^{-1}$ to 0.4 $Nm \cdot Kg^{-1}$ for abduction, and from 1.0 $Nm \cdot Kg^{-1}$ to 0.37 $Nm \cdot Kg^{-1}$ for adduction, at 60 through 360° $\cdot s^{-1}$.

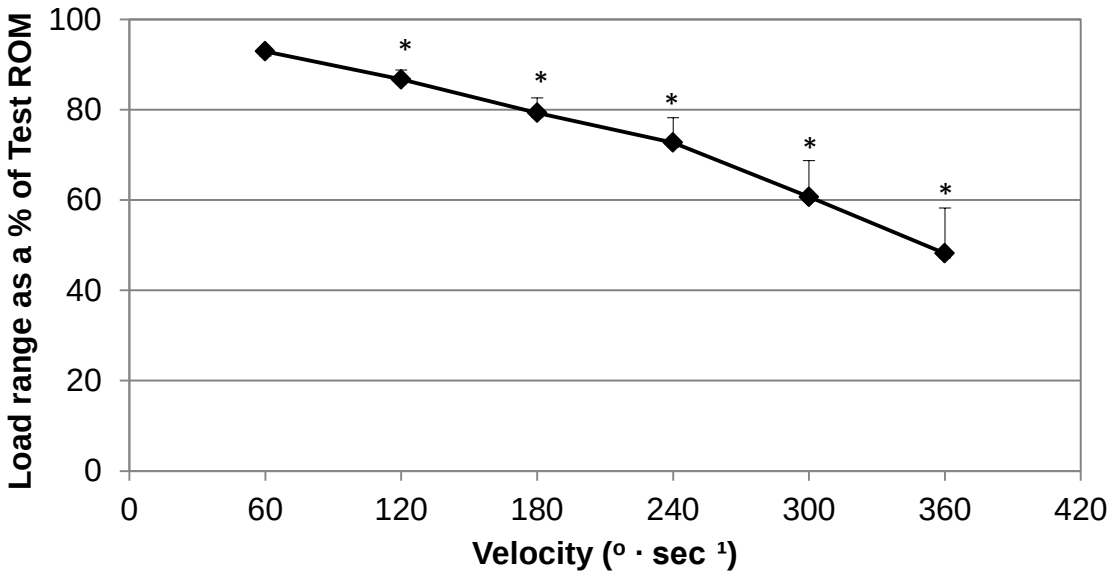


Figure 7.25. Load Range as a Percentage of the Total Test Range of Motion (ROM) During Hip Abduction (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity.

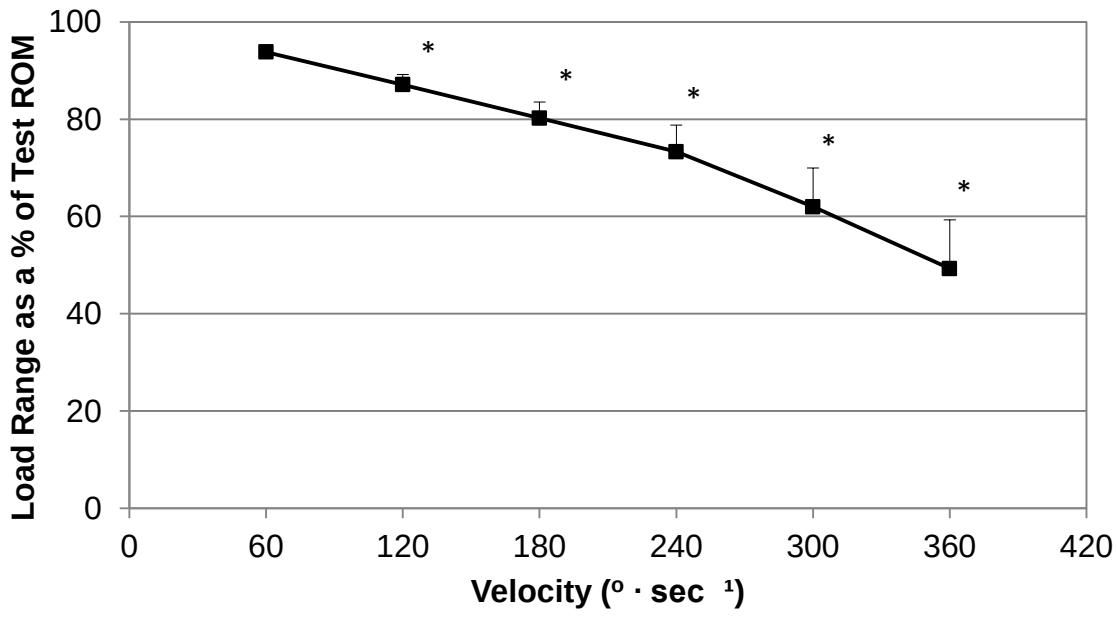


Figure 7.26. Load Range as a Percentage of the Total Test Range of Motion (ROM) During Hip Adduction (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity.

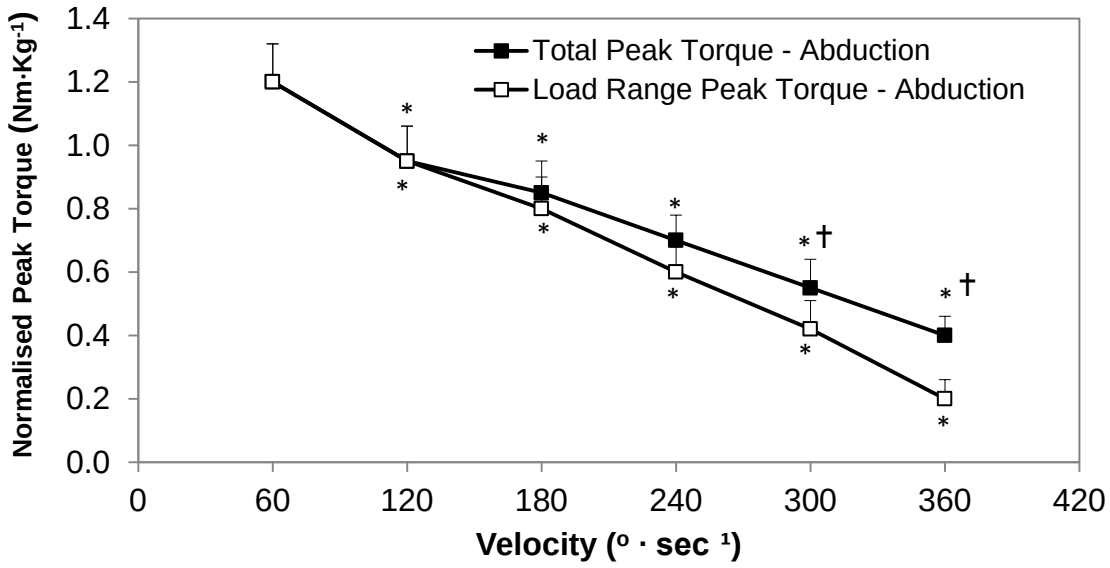


Figure 7.27. Normalised Total and Load Range Peak Torque for Hip Abduction with Changes in Velocity (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity. † Significant difference (*P*<0.05) between normalised total peak torque and load range peak torque at corresponding velocity.

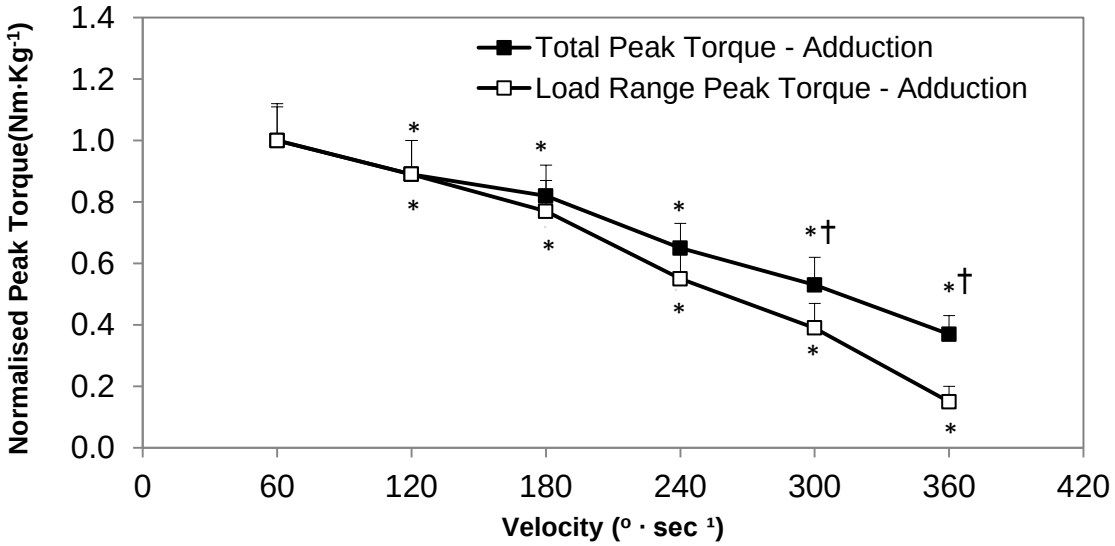


Figure 7.28. Normalised Total and Load Range Peak Torque for Hip Adduction with Changes in Velocity (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity. † Significant difference (*P*<0.05) between normalised total peak torque and load range peak torque at corresponding velocity.

Normalised load range peak torque (Figures 7.27 and 7.28) values significantly decreased with each increase in velocity for both hip abduction and hip adduction. There was no significant difference found between hip abduction and adduction results. The normalised load range peak torque values significantly decreased from 1.2 $Nm \cdot Kg^{-1}$ to 0.2 $Nm \cdot Kg^{-1}$ for abduction, and from 1.0 $Nm \cdot Kg^{-1}$ to 0.1 $Nm \cdot Kg^{-1}$ for adduction, at 60 through 360° $\cdot s^{-1}$. The 6 x 2 x 2 repeated measures ANOVA results also showed a significant difference between normalised total peak torque and load range peak torque from speeds of 300° $\cdot s^{-1}$ and above for both hip abduction and hip adduction.

The 6 x 2 x 2 (speed x analysis type x movement) repeated measures ANOVA results revealed that normalised total work (Figures 7.29 and 7.30) values significantly decreased with each increase in velocity for both hip abduction and hip adduction. There was no significant difference found between hip abduction and adduction results. The normalised total work values significantly decreased from 0.85 $Nm \cdot Kg^{-1}$ to 0.5 $Nm \cdot Kg^{-1}$ for abduction, and from 0.84 $Nm \cdot Kg^{-1}$ to 0.49 $Nm \cdot Kg^{-1}$ for adduction, at 60 through 360° $\cdot s^{-1}$.

Normalised load range work (Figures 7.29 and 7.30) values significantly decreased with each increase in velocity for both hip abduction and hip adduction. There was no significant difference found between hip abduction and adduction results. The normalised load range work values significantly decreased from 0.82 $Nm \cdot Kg^{-1}$ to 0.11 $Nm \cdot Kg^{-1}$ for abduction, and from 0.79 $Nm \cdot Kg^{-1}$ to 0.09 $Nm \cdot Kg^{-1}$ for adduction, at 60 through 360° $\cdot s^{-1}$. The 6 x 2 x 2 repeated measures ANOVA results also showed a significant difference between normalised total work and load range work from speeds of 120° $\cdot s^{-1}$ and above for both hip abduction and hip adduction.

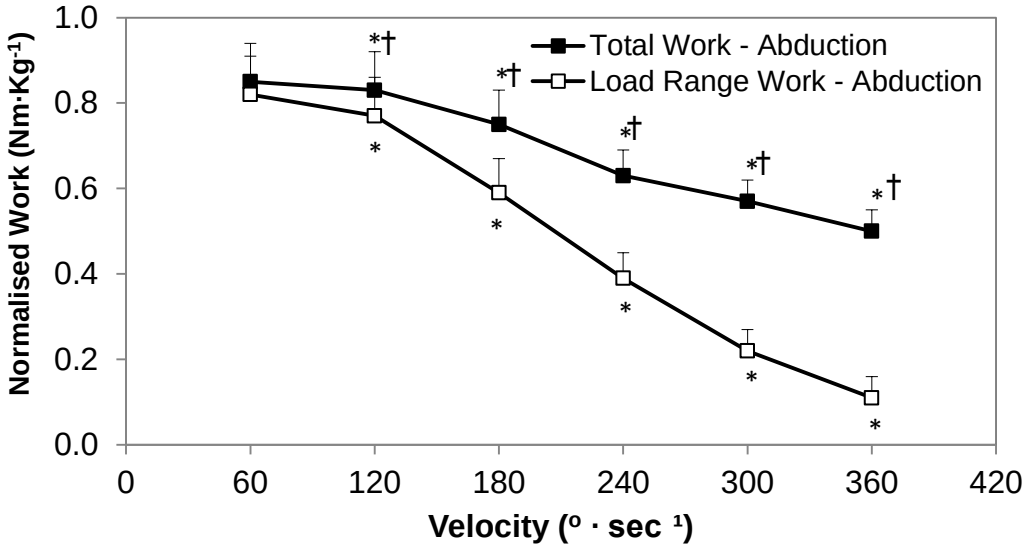


Figure 7.29. Normalised Total and Load Range Work for Hip Abduction with Changes in Velocity (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity. † Significant difference (*P*<0.05) between normalised total work and load range work at the corresponding velocity.

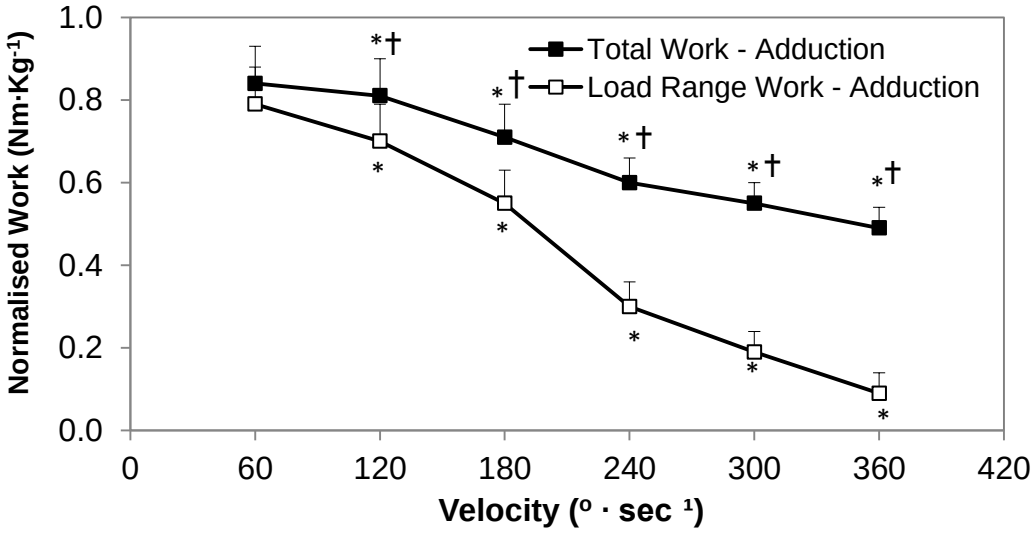


Figure 7.30. Normalised Total and Load Range Work for Hip Adduction with Changes in Velocity (Mean ± SD). * Significantly (*P*<0.008) different from previous velocity. † Significant difference (*P*<0.05) between normalised total work and load range work at the corresponding velocity.

7.12.5 Discussion

This study aimed to quantify the components of load range, acceleration, and deceleration through a velocity spectrum during concentric hip abduction and adduction isokinetic exercise. The secondary aim of the study was to investigate the effect of load range on peak torque and work done. It is apparent from the results that load range significantly decreased while acceleration and deceleration ROM significantly increased with each increase in velocity, for both hip abduction and hip adduction. Therefore, the hypothesis of this study which stated that with an increase in velocity there will be a significant decrease in the load range component, and a significant increase in the acceleration and deceleration components can be formally accepted

When the total peak torque data was corrected for load range there was a significant decrease in peak torque at velocities of $300° \cdot s^{-1}$ and above for both hip abduction and hip adduction. Load range correction also resulted in a significant decrease in work done at velocities of $120° \cdot s^{-1}$ and above for both hip abduction and hip adduction. Therefore, the hypothesis of this study which stated the load range corrected peak torque and total work data will be significantly different to the uncorrected data at higher velocities, can be formally accepted.

7.12.5.1 Load Range, Acceleration and Deceleration

The findings of the present study reflected past investigations in which isokinetic constant velocity movement was measured under concentric conditions. Osternig (1986) reported that knee extension load range decreased from 92% to 16% at speeds

of 50 through 400° · s^{-1}. Wilk et al. (1994) also described a torque range decrease of 87% to 19% from speeds of 180 through 450° · s^{-1} during knee extension and flexion. Also investigating the knee, Kurdak et al. (2005) found a reduction in load range from 94% to 4% for knee extension at speeds 30 through 390° · s^{-1}, and from 94.5% to 6.5% for knee flexion at speeds 30 through 450° · s^{-1}. Scibelli et al. (1993) demonstrated that bilateral knee extension/flexion load range decreased from 87.8% to 31.8% at speeds from 60 through 360° · s^{-1}. In addition to this, Brown, Whitehurst, Gilbert et al. (1995) found that load range decreased from 95.3% to 0% and from 96.3% to 21.8% during shoulder external and internal rotation, respectively, at speeds from 60 through 450° · s^{-1}. The current study found that load range decreased from 92.9% to 48.2% for hip abduction, and from 93.8% to 49.3% for hip adduction, at speeds of 60 through 360° · s^{-1}.

It is apparent that the results of the present study mirror the findings of the above authors (Brown, Whitehurst, Gilbert et al., 1995; Kurdak et al., 2005; Osternig, 1986; Scibelli et al., 1993; Wilk et al., 1994), as they all found an inverse relationship between load range and isokinetic velocity. Brown and Whitehurst (2000) highlighted the importance of separating the data into the three phases of acceleration, load range and deceleration. Surprisingly, some authors still fail to do this, and only consider the load range component (Kurdak et al., 2005). However, Brown, Whitehurst, Gilbert et al. (1995) did consider the impact of acceleration and deceleration and found that both components significantly increased with each increase in velocity. These results mirror the findings of the present study. However, direct comparisons of the results must be made with caution due to the fact that Brown, Whitehurst, Gilbert et al. (1995) studied

the flexors and extensors of the knee, with a ROM of 80˚, whereas the present study investigated the abductors and adductors of the hip with a ROM of only 45˚.

If the results by Brown, Whitehurst, Gilbert et al. (1995) are converted to a percentage it can be seen that acceleration increased from 1.3% to 18.1% for knee extension, and from 1.3% to 19.1% for knee flexion, at 60 through 360° · s^{-1}. Deceleration also increased from 2.5%% to 27.8% for knee extension, and from 2.1% to 28.0% for knee flexion, at 60 through 360° · s^{-1}. In comparison, the results of the present study found that acceleration significantly increased from 2.4% to 24.7% for hip abduction, and from 2.0% to 23.8% for hip adduction, at 60 through 360° · s^{-1}. Deceleration also significantly increased from 4.7% to 27.1% for hip abduction, and from 4.2% to 26.9% for hip adduction, at 60 through 360° · s^{-1}. It is clear to see there is an increase in the acceleration and deceleration components with an increase in isokinetic velocity in both studies.

In comparison to the results of Brown, Whitehurst, Gilbert et al. (1995) the present study showed a higher percentage of the ROM being spent in acceleration, but found a similar percentage of the ROM being spent in deceleration. Possible reasons for the differences in acceleration may be due to the different joint and musculature being tested between studies. The deceleration component was shown to be similar between studies, this may be due to using the same cushioning level as the Brown, Whitehurst, Gilbert et al. (1995) study, and the isokinetic dynamometer, rather than the subject themselves, was in control of this factor. However, once again direct comparisons should be made with caution due to the previously identified differences between the studies.

By quantifying ROM for the load range, acceleration and deceleration components a more complete understanding of a concentric action on the isokinetic dynamometer can be achieved. The results emphasise the need for the clinician to fully understand the inverse relationship between isokinetic velocity and load range, and select the appropriate velocity accordingly. Any strength gains from training on the isokinetic dynamometer may be relative to the total amount of ROM actually sustained at the pre-selected velocity (ie, load range).

The results from the current study also emphasise the variation that exists between different joints. Even though the same general trend was indentified (load range significantly decreased while acceleration and deceleration ROM significantly increased with each increase in velocity), it can be seen that different joints have different levels of acceleration, load range, deceleration and maximum speed. These results further elucidate the findings that it is very important to load range correct data prior to analysis and that one cannot utilise factors from dissimilar joints. Therefore, the results from the present study should only be employed by future researchers if they are investigating the abductors and adductors of the hip.

7.12.5.2 Load Range Correction for Peak Torque and Work Done

In the present study there was a significant difference between normalised total peak torque and load range peak torque from speeds of $300° \cdot s^{-1}$ and above for both hip abduction and hip adduction. There was also a significant difference between normalised total work and load range work from speeds of $120° \cdot s^{-1}$ and above for both hip abduction and hip adduction. In agreement with these findings Kurdak et al. (2005)

found that the consideration of load range for peak torque and work calculations resulted in a significant decrease in the data when compared to the data presented by the isokinetic dynamometer. The authors found a significant difference between total peak torque and load range peak torque at speeds above 270° · s^{-1} for knee extension, and above 300° · s^{-1} for knee flexion. They also found a significant difference between total work and load range work at speeds above 90° · s^{-1} for both knee extension and knee flexion (Kurdak et al., 2005). These results highlight the importance of correcting the data for load range as it is apparent that large errors can occur if this process is not undertaken (Brown & Whitehurst, 2000).

The normalised load range peak torque values and the normalised load range work values in the present study were lower than the results reported by Kurdak et al. (2005). However, this was expected as Kurdak et al. (2005) studied the flexors and extensors of the knee joint and not the abductors and adductors of the hip joint. Unfortunately the majority of studies investigating peak torque and work of the hip abductors and hip adductors do not normalise their data to the subject's body weight (Claiborne et al., 2009; Jacobs et al., 2005; Laheru et al., 2007; Piva et al., 2011). They also do not indicate whether load range correction was completed (Claiborne et al., 2009; Jacobs et al., 2005; Johnson et al., 2004; Laheru et al., 2007; Piva et al., 2011) which unfortunately makes comparisons of the data difficult. Only one study by Johnson et al. (2004) reported the data in terms of normalised peak torque values. The authors found normalised peak torque values of 0.93 Nm·Kg^{-1} for the hip abductors and 1.01 Nm·$Kg^{-1}$for the hip adductors, at an isokinetic velocity of 60° · s^{-1} (Johnson et al., 2004). These results are similar to the results reported in the present study which found values of 1.2 Nm·Kg^{-1} and 1.0 Nm·Kg^{-1} for the hip abductors and hip adductors, respectively at 60° ·

s^{-1}. However, no faster speeds were tested by Johnson et al. (2004) so only the comparison at 60° · s^{-1} can be made. Johnson et al. (2004) also did not indicate if the data was reduced for load range, therefore comparisons should be made with caution as inconsistencies may be present.

The results from the present study indicate that load range corrected results are significantly different from the 'total' results produced by the isokinetic dynamometer at speeds of 300° · s^{-1} and above for peak torque data, and 120° · s^{-1} and above for work data, for both hip abduction and hip adduction. This trend is different to what has been found at other joints, and emphasises the fact that it is vital to load range correct data prior to analysis and that one cannot employ factors from dissimilar joints. For that reason, the results from the present study should only be utilised by future researchers if they are investigating the abductors and adductors of the hip.

In terms of velocity prescription for the hip joint, there seems to be a lack of consensus in the literature on the most appropriate speed. The hip has very rarely been studied, and the sparse literature that is available have tested the hip in the isometric mode (Piva et al., 2011), or have tested at speeds of 60 and 90° · s^{-1} (Salavati et al., 2007). Ferber et al. (2003) found that during running at 3.65 m/s (13.2 km/hr), the peak angular velocity for the hip was 103.5° · s^{-1}. Even though this speed may be far from 'explosive sporting movement' velocities, it may replicate speeds from more endurance based activities, The present study investigated a velocity spectrum from 60 to 360° · s^{-1}, but the results from Ferber et al. (2003) possibly indicate that speeds close to 103.5° · s^{-1} are most relevant and should be chosen when investigating athletes from more endurance based sports.

7.12.5.3 Clinical Implications

The results from the present study imply that peak torque and total work values should always be corrected by the clinician to account for load range, as otherwise errors may be present. As the isokinetic dynamometer is often used for training or rehabilitation, the results identify a need for the clinician to carefully consider velocity selection during hip abduction and adduction exercise. Any strength gains from isokinetic training may be proportional to the amount of time actually spent at the pre-selected velocity (ie, load range).

7.12.5.4 Limitations and Recommendations for Future Research

Only young male subjects were recruited for this study. A similar study should be repeated investigating female subjects, but also different age groups. It must be remembered that the results are only applicable if the same joint, equipment and protocol is used as in the current study. Future studies may wish to repeat this study but using different joints, different makes of isokinetic dynamometers and varied protocols.

7.12.6 Conclusion

In summary, the results indicate that an inverse relationship exists between load range and velocity during concentric hip abduction and hip adduction isokinetic exercise. If the velocity is not reached, the result is in absence of machine offered resistance. In addition, the results emphasise the importance of also considering the acceleration and

deceleration components, as these both significantly increased with each increase in velocity, for hip abduction and hip adduction.

The results also highlight the importance of correcting the data for load range, as it is apparent that large errors can occur if this process is not undertaken. Both peak torque and work decreased following load range correction. As the isokinetic dynamometer is often used for training or rehabilitation, the results identify a need for the clinician to carefully consider velocity selection during hip abduction and hip adduction exercise. Any strength gains from isokinetic training may be proportional to the amount of time actually spent at the pre-selected velocity (ie, load range).

7.13 Development of Research

Pilot Study Nine addressed the effect of load range on peak torque and total work values during isokinetic hip abduction-adduction. The results found that the peak torque and total work values given by the isokinetic dynamometer had to be adjusted to account for load range. Therefore, it was very important that this method was adopted during Study Three and Four.

Chapter Eight

The Effect of Localised and Globalised Fatigue on Muscle Latency in Healthy versus Functionally Unstable Subjects

8.1 Study Three: The Effect of Localised and Globalised Fatigue on Muscle Latency in Healthy and Functionally Unstable Subjects Following a Simulated Ankle Sprain

8.1.1 Abstract

Aim: To research muscle latency in the unilateral FAI subject's UA and SA, compared to a healthy control group's DA and NDA, both before and immediately after a) ankle inversion-eversion isokinetic fatigue, b) hip abduction-adduction isokinetic fatigue, c) treadmill exercise simulating football match play, and d) a control. **Method:** Twenty males suffering from a unilateral FAI and 20 male healthy controls were subjected to six inversion and plantarflexion tilt perturbations, three on each leg, both before and immediately after each protocol. Electromyographic signals were recorded for the peroneus longus, tibialis anterior and gluteus medius muscles of both limbs. **Results:** The results indicate that the fatigue conditions when compared to the pre-test and control conditions showed no significant difference in muscle latency for all muscles tested, in all groups (UA, SA, DA and NDA). **Conclusion:** It has previously been suggested that muscle fatigue can lead to injury, as reflected by the increased injury risk in the second half, especially during the last quarter of the match. However, results from the present study suggest that fatigue does not lead to increased muscle latencies, and therefore, other factors must be present that lead to this increased injury rate.

8.1.2 Introduction

Recurrent sprains have been reported in over 70% of patients who had previously sustained an inversion ankle sprain (Kent-Braun, 1999; Yeung et al., 1994). Repeated sprains, residual disability, a feeling of “giving way”, and a sensation of joint weakness characterise functional ankle instability, a condition that often arises secondary to inversion trauma (Beckman & Buchanan, 1995; Fernandes et al., 2000; Konradsen et al., 1998; Konradsen & Ravn, 1991). Due to the significant amount of time lost from sport and work, research on the factors that contribute to ankle injuries is warranted.

Neuromuscular control can be defined as the interaction between the nervous and musculoskeletal systems to produce a desired effect, specifically in response to a stimulus (Hertel, 2000). In the ankle specifically, the lateral ligaments are highly innervated by mechanoreceptors (Myers, Riemann, Hwang, Fu, Lephart, 2003), which when stretched sensitise the muscle spindles in the peroneal muscles, subsequently causing a reflex contraction to oppose the stretch (Johansson, Sjolander & Sojka, 1991). Many studies have investigated muscle latencies of the peroneus longus (Isakov et al., 1986; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990) and the tibialis anterior muscles in healthy and functionally unstable subjects (Ebig et al., 1997; Lofvenberg et al., 1995; Mitchell et al., 2008a). However, there is very limited research on the muscle latency of the gluteus medius muscle (Beckman & Buchanan, 1995). Weakness in a stabilising muscle, such as the gluteus medius, may produce deviations in joint motion, a subsequent loss of stability and may contribute towards a repeated injury at the ankle (Friel et al., 2006; Riemann, 2002).

Some authors suggest that fatigue plays a significant role in the occurrence of ankle injuries (Gribble & Hertel, 2004a; Huston et al., 2005; Ochsendorf et al., 2000; Pasquet et al., 2000). Fatigue is defined as any exercise-induced reduction in force generating capacity of a muscle (Bigland-Ritchie & Woods, 1984). Anecdotally, many injuries occur during the latter stages of activity when fatigue is present (Hawkins et al., 2001). Whether the onset of fatigue occurs centrally or peripherally, many researchers have documented decreases in the neuromuscular feedback system of the joint around which the fatigued muscles are located (Gribble & Hertel, 2004a; Harkins et al., 2005; Yaggie & McGregor, 2002; Yeung et al., 1999). No studies have evaluated the muscular latency times of the ankle musculature to an ankle inversion and plantarflexion perturbation before and immediately after localised and globalised fatigue protocols.

Localised fatigue is usually induced through isokinetic protocols. Isokinetic fatigue has often been defined when the peak torque falls below 50% of the maximum voluntary contraction (Ochsendorf et al., 2000; Wikstrom et al., 2004; Yaggie & McGregor, 2002). It has previously been shown that under fatigued conditions, concentric muscle actions result in a greater loss of force than eccentric actions (Pasquet at al., 2000). Research has also found that isokinetic fatigue has led to increased (delayed) muscle latencies in healthy subjects (Cools et al., 2002). However, functionally unstable subjects have not been investigated, and the majority of studies have only examined the effect of fatiguing the ankle musculature (Jackson et al., 2009; Mora, Quinteiro-Blondin & Perot, 2003), no studies have investigated the more proximal stabilizing muscles, such as the gluteus medius.

Most studies of fatigue have evaluated isokinetic contractions (Jackson et al., 2009; Mora et al., 2003; Wikstrom et al., 2004; Yaggie & McGregor, 2002). These types of contractions may not be representative of muscle activity and fatigue development during participation in sports (Green, 1995). Intermittent exercise of the type that occurs in a football game is characterized by a variety of muscle activities. The movements that make up the majority of activities in football are locomotor movements such as running, jogging and walking (Rahnama, Reilly & Lees, 2005). These movements involve the use of the major joint flexors and extensors of the ankle, knee and hip. It is difficult to investigate muscle fatigue in response to a competitive football game, due to practical difficulties, and standardisation. Therefore, laboratory based protocols have been developed, such as the Drust protocol (Drust et al., 2000), which can be used to simulate the competitive event. There is limited research on the effect of football-specific fatigue protocols on muscle latency times; however, if fatigue has a detrimental effect on muscle latency, this could lead to an increased risk of injury. It has been previously suggested that muscle fatigue can lead to injury (Davis & Bailey, 1997), as reflected in the increased risk of injury in the second half, especially during the last quarter of the match (Hawkins et al., 2001).

While several studies have evaluated muscle latencies in healthy versus functionally unstable subjects (Beckman & Buchanan, 1995; Ebig et al., 1997; Johnson & Johnson, 1993; Konradsen et al., 1998; Konradsen & Ravn, 1991), a better understanding of the ankle musculature's responses to an inversion-plantarflexion stress in a fatigued state may help to clear up discrepancies in the literature, and identify if fatigue is a risk factor that may lead to an ankle sprain in healthy subjects, or lead to repeated sprains in FAI subjects.

The aim of this study was therefore to research muscle latency in the unilateral FAI subject's UA and SA, compared to a healthy control group's DA and NDA, both before and immediately after a) ankle inversion-eversion isokinetic fatigue, b) hip abduction-adduction isokinetic fatigue, c) treadmill exercise simulating football match play, and d) a control. It was hypothesised that the FAI subjects would have increased (delayed) muscle latencies in comparison to the healthy control group, across all conditions. It was also hypothesised that the fatigue protocols will further amplify the effect of increased (delayed) muscle latencies in the FAI group in comparison to the pre test and control conditions.

8.1.3 Method

8.1.3.1 Subjects

Forty male subjects were recruited for this study; twenty subjects suffered from functional ankle instability (age = 23.08 $\pm$ 5.05 years, height = 179.20 $\pm$ 5.78 cm, and mass = 79.85 $\pm$ 8.35 kg) and twenty subjects served as healthy controls (age = 22.5 $\pm$ 4.31 years, height = 181.23 $\pm$ 6.15 cm, and mass = 81.07 $\pm$ 11.17 kg). Institutional ethical approval was granted for this study. All subjects read the subject briefing document (Appendix One) and provided written informed consent (Appendix Two) before participation.

Refer to Pilot Study One (Section 3.2.2.1, paragraphs 2, 3 and 4) for inclusion and exclusion criteria.

8.1.3.2 Experimental Design

The same experimental design as Pilot Study One was used; apart from the EMG signal and digitals sampled at 1000 Hz, and subjects were not required to return to the laboratory seven days later to repeat the procedure (Section 3.2.3.2).

Following EMG setup and tilt perturbations at rest, the subject randomly undertook the first of four fatigue procedures. Each procedure was performed with seven days in between, to ensure that one procedure did not have an effect on another. The four procedures were a) ankle inversion-eversion isokinetic fatigue, b) hip abduction-adduction isokinetic fatigue, c) football-specific fatigue or d) 105 minutes quiet rest (control).

Ankle Inversion-Eversion Isokinetic Fatigue

The same isokinetic ankle inversion-eversion fatigue protocol was used as in Pilot Study Four (Section 7.2.3.2), apart from the following; a speed of $120° \cdot s^{-1}$ was used and subjects were not required to return to the laboratory for repeat testing. Immediately following the ankle fatigue protocol, three tilt perturbations were performed randomly on each leg, but only trials from the right limb (fatigued limb) were recorded. The fatigue procedure was then repeated on the left limb.

Hip Abduction-Adduction Isokinetic Fatigue

The same isokinetic hip abduction-adduction fatigue protocol was used as in Pilot Study Seven (Section 7.8.3.2), apart from the following; a speed of 120° · s^{-1} was used, and subjects were not required to return to the laboratory for repeat testing. Immediately following the hip fatigue protocol, three tilt perturbations were performed randomly on each leg, but only trials from the right limb (fatigued limb) were recorded. The fatigue procedure was then repeated on the left limb.

Treadmill Exercise Simulating Football Match Play

The football-specific intermittent exercise protocol was used to provide fatiguing exercise estimated to be equivalent in intensity to playing a game of football (Van Gool, Van Gervan & Boutmans, 1988). The football-specific protocol was performed on a programmable motorized treadmill (Pulsar, HP Cosmos, Nussforf-Traunstein, Germany) and consisted of the different exercise intensities that are observed during football match play (e.g. walking, jogging, running and sprinting).

The pattern of activities in the protocol was similar to that observed by Reilly and Thomas (1976) and the percentage of the total time spent in each activity approximated data collected with time-motion analysis (Bangsbo, 1994; Bangsbo et al., 1991; Reilly & Thomas, 1976; Van Gool et al., 1988; Yamanaka, Haga, Shindo, Narita, Koeski, Matsuura & Eda, 1988). The speeds of each activity in the protocol were 6 km · hr^{-1} (walking), 12 km · hr^{-1} (jogging), 15 km · hr^{-1} (running/cruising) and 21 km · hr^{-1} (sprinting), and were varied in order and duration following the procedure employed by

Drust et al. (2000) and in accordance with the observations by Van Gool et al. (1988) during football match play. Backwards movements, sideways movements, and actions with the ball were not included in the protocol because of the technical impracticalities and safety when using a motorized treadmill.

Each half of the football-specific intermittent protocol was structured as two parts, each 22 minutes in duration separated by one minute static rest, leading to a total of 45 minutes. There was an intermission of 15 minutes between the halves, where the subjects rested. The same protocol was then replicated for the second half. The procedure for this protocol on the treadmill was determined by Drust (1997) to be reliable and repeatable, with a reported coefficient of variation of 4.8% and 95% ratio limits of agreement of 9.4%. Immediately following the football-specific fatigue protocol three tilt perturbations were performed randomly on each leg, and averages of these were used for analysis

Control - 105 Minutes Rest

The subject was required to remain seated for 105 minutes. This quiet rest was used as a control for the football-specific treadmill protocol (total 105 minutes), so that any differences that may occur following the treadmill protocol, are not merely down to test-retest differences. Immediately following the 105 minutes rest, three tilt perturbations were performed randomly on each leg, and averages of these were used for analysis

Following the completion of each procedure the surface electrodes were removed from the subject's lower limbs. The subject then performed a five minute cool down on the cycle ergometer, at 50 rpm with a resistance of 50 Watts.

8.1.3.3 Data Analysis

The same data analysis procedure was used as in Pilot Study One; apart from the EMG trace was processed using the RMS method and was smoothed by 2 ms.

8.1.3.4 Statistical Analysis

Firstly, using SPSS (version 19) statistical tests were performed to identify differences between the ankles tested (DA, NDA, UA and SA) in each of the five conditions (pre-test, ankle isokinetic fatigue, hip isokinetic fatigue, football specific fatigue or control) for each of the muscles tested, when acting as a tilt and support limb. Both univariate normality (Shapiro-Wilk) and multivariate normality (Mahalanobis distances) were verified. Linearity was confirmed by generating a matrix of scatterplots between each pair of variables, separately for each group. A multivariate analysis of variance (MANOVA) was used to explore the differences in muscle latency between the ankles tested in each condition for each of the muscles tested, when acting as a tilt and support limb. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of homogeneity of variance-covariance matrices. The Multivariate Test box (Wilk's Lambada value) was studied for significant differences between the ankles ($P<0.05$).

The Test of Between-Subject Effects box was then observed to identify differences between the ankles for each condition ($P<0.05$). Tukey's post-hoc test was used to determine exactly where the significant findings occurred between the ankles. Due to multiple comparisons being made between groups, a Bonferroni adjustment was carried out, and the alpha level was set at $P<0.0125$.

Secondly, statistical tests were performed to identify differences between the five conditions in each ankle tested for each of the muscles tested, when acting as a tilt and support limb. Both univariate normality (Shapiro-Wilk) and multivariate normality (Mahalanobis distances) were verified. Linearity was confirmed by generating a matrix of scatterplots between each pair of variables, separately for each group. A multivariate analysis of variance (MANOVA) was used to explore the differences in muscle latency between the fatigue conditions in each ankle tested for each of the muscles tested, when acting as a tilt and support limb. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of homogeneity of variance-covariance matrices. The Multivariate Test box (Wilk's Lambada value) was studied for significant differences ($P<0.05$). The Test of Between-Subject Effects box was then observed to identify differences for each of the fatigue conditions ($P<0.05$).

8.1.4 Results

Results from the MANOVA for the tilting limb showed a significant ($P<0.0125$) increase (delay) in muscle latency when comparing the UA of the FAI group to both the DA and

NDA of the control group, across all five conditions, in the peroneus longus (Figure 8.1), tibialis anterior (Figure 8.2) and gluteus medius (Figure 8.3) muscles. The results also showed a significant (P<0.0125) increase (delay) in muscle latency when comparing the SA of the FAI group to both the DA and NDA of the control group, across all conditions in the peroneus longus (Figure 8.1), tibialis anterior (Figure 8.2) and gluteus medius (Figure 8.3) muscles. No significant differences were found between the UA and SA of the FAI group, across all five conditions, for any of the muscles tested (Figures 8.1 to 8.3). In addition to this, no significant differences were found between the DA and NDA of the control group, across the five conditions, for any of the muscles tested (Figures 8.1 to 8.3).

The MANOVA results also showed that when observing the tilting limb results, the fatigue conditions (ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue) when compared to the pre-test and control conditions showed no significant difference in muscle latency for all muscles tested, in all ankles (Figures 8.1 to 8.3).

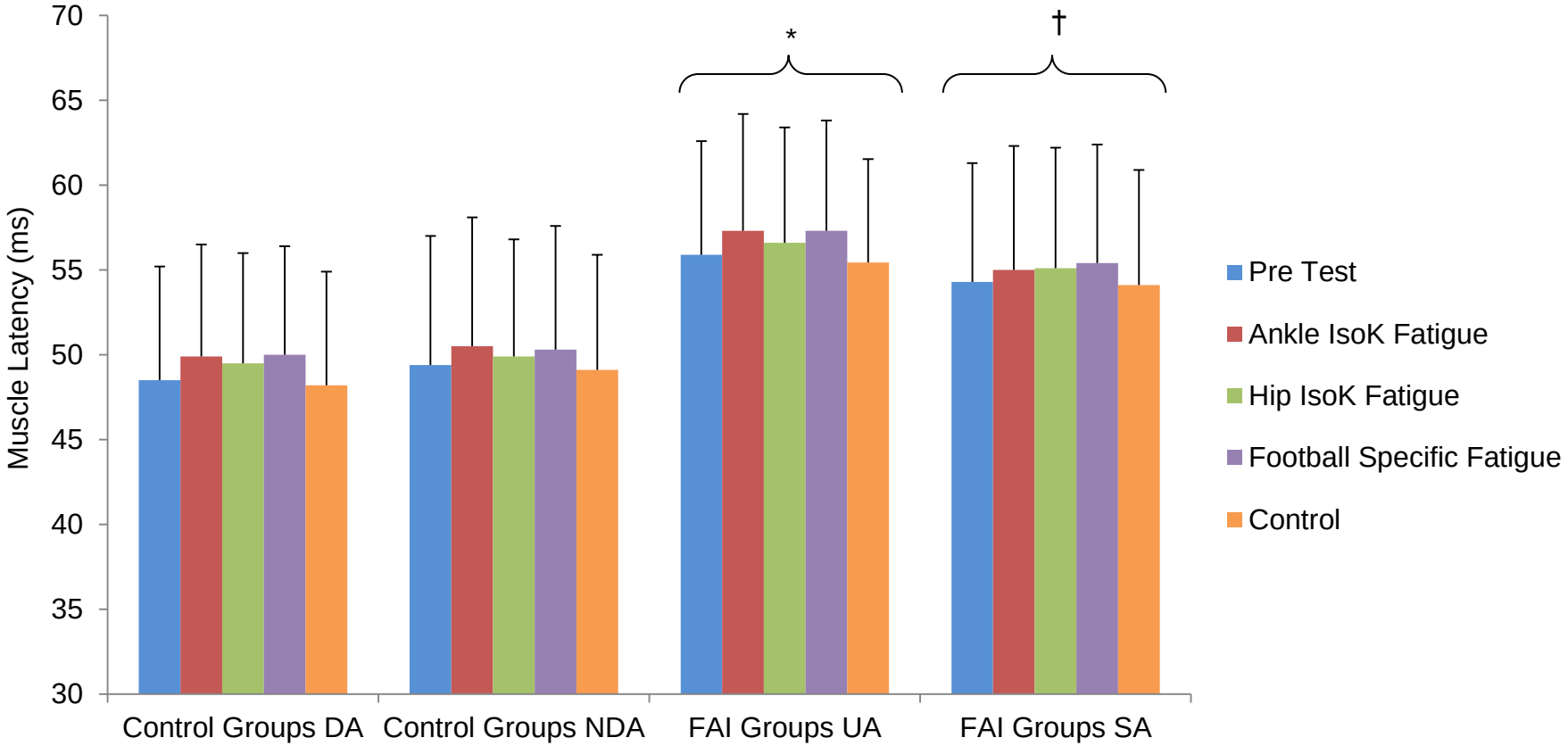


Figure 8.1. Muscle Latencies for the Peroneus Longus when Acting as a Tilting Limb, in the Control Groups DA and NDA and the FAI Groups UA and SA, Across all Five Conditions (Mean ± SD). * FAI groups UA significantly (P<0.0125) slower for each corresponding condition, than the control groups DA and NDA. † FAI groups SA significantly (P<0.0125) slower for each corresponding condition, than the control groups DA and NDA.

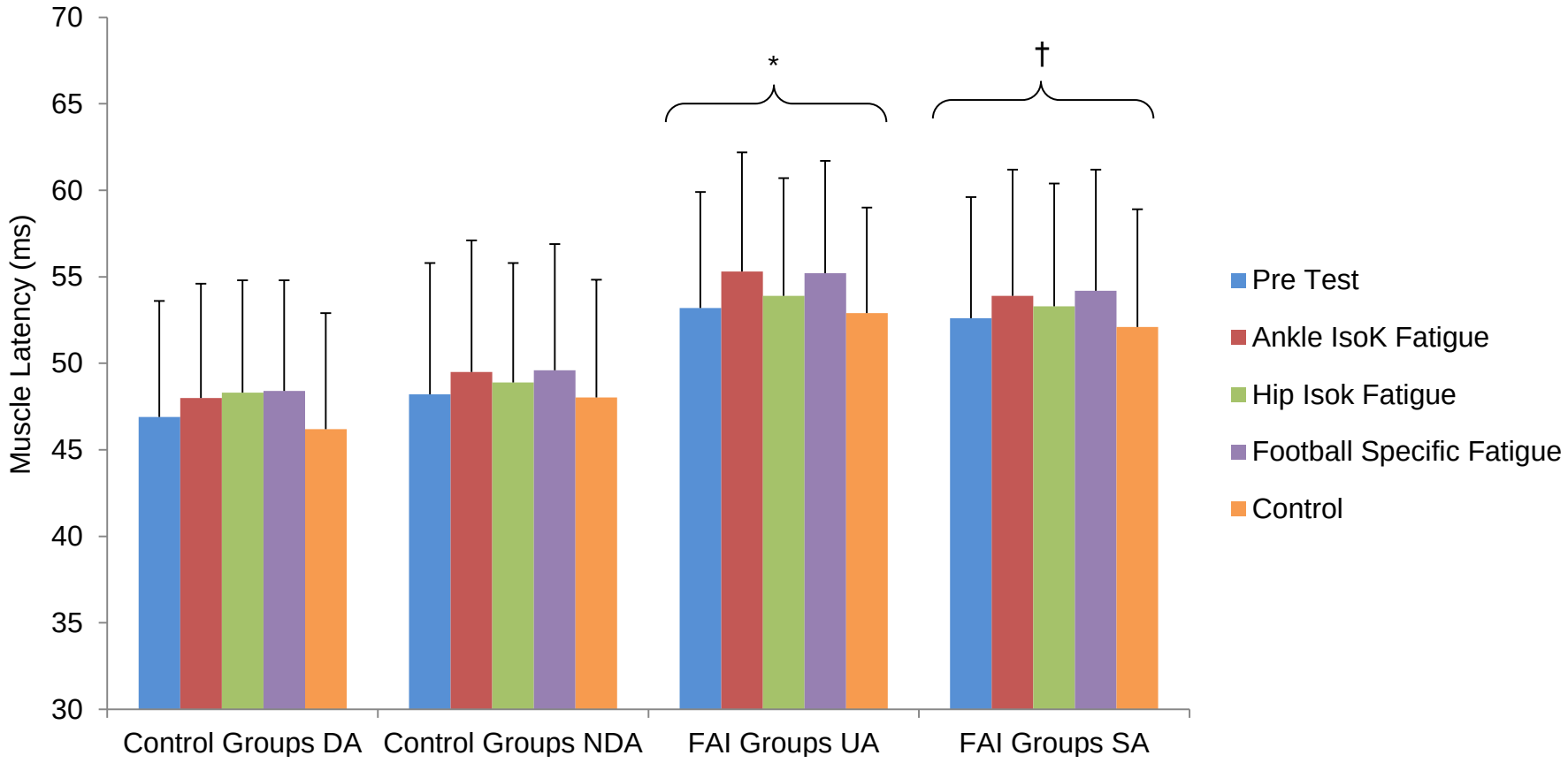


Figure 8.2. Muscle Latencies for the Tibialis Anterior when Acting as a Tilting Limb, in the Control Groups DA and NDA and the FAI Groups UA and SA, Across all Five Conditions (Mean ± SD). * FAI groups UA significantly (*P*<0.0125) slower for each corresponding condition, than the control groups DA and NDA. † FAI groups SA significantly (*P*<0.0125) slower for each corresponding condition, than the control groups DA and NDA.

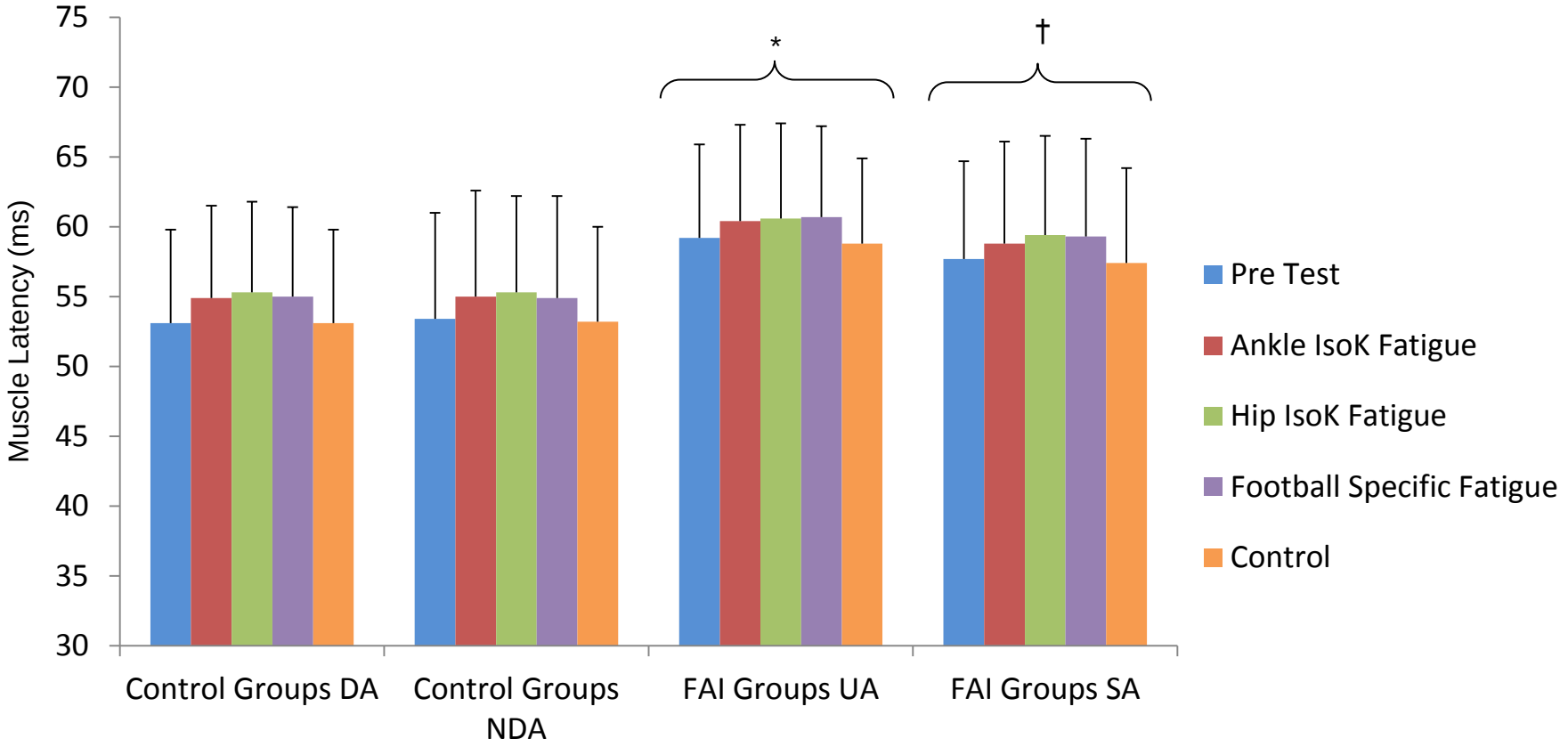


Figure 8.3. Muscle Latencies for the Gluteus Medius when Acting as a Tilting Limb, in the Control Groups DA and NDA and the FAI Groups UA and SA, Across all Five Conditions (Mean ± SD). * FAI groups UA significantly (P<0.0125) slower for each corresponding condition, than the control groups DA and NDA. † FAI groups SA significantly (P<0.0125) slower for each corresponding condition, than the control groups DA and NDA.

Results from the MANOVA for the support limb showed no significant difference in muscle latency when comparing the UA of the FAI group to both the DA and NDA of the control group, across all five conditions, in the peroneus longus (Figure 8.4), tibialis anterior (Figure 8.5) and gluteus medius (Figure 8.6) muscles. The results also showed no significant difference in muscle latency when comparing the SA of the FAI group to both the DA and NDA of the control group, across all five conditions, in the peroneus longus (Figure 8.4), tibialis anterior (Figure 8.5) and gluteus medius (Figure 8.6) muscles. No significant differences were found between the UA and SA of the FAI group, across all five conditions, for any of the muscles tested (Figures 8.4 to 8.6). In addition to this, no significant differences were found between the DA and NDA of the control group, across the five conditions, for any of the muscles tested (Figures 8.4 to 8.6).

The MANOVA also showed that when observing the support limb results, the fatigue conditions (ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue) when compared to the pre-test and control conditions showed no significant difference in muscle latency for all muscles tested, in all ankles (Figures 8.4 to 8.6).

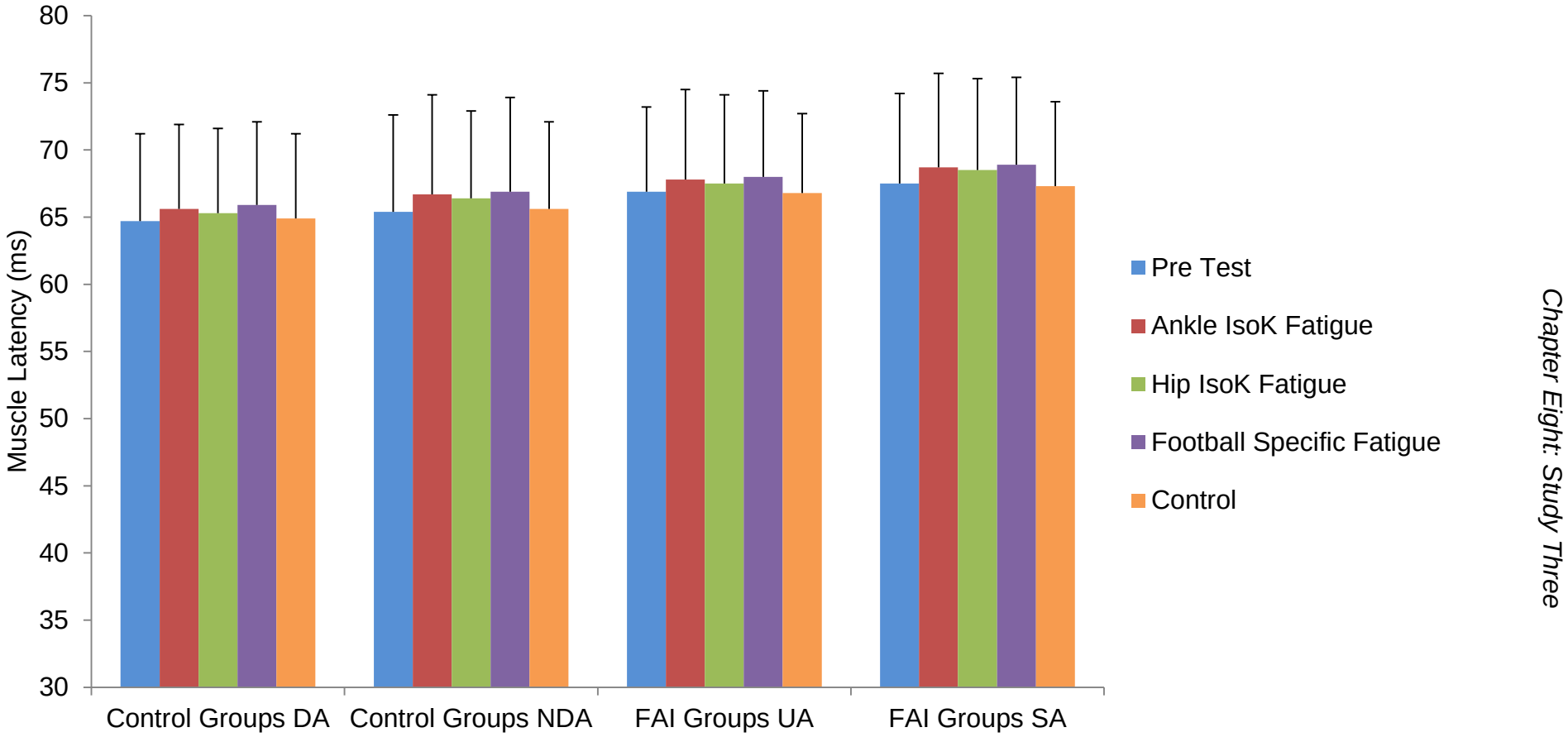


Figure 8.4. Muscle Latencies for the Peroneus Longus when Acting as a Support Limb, in the Control Groups DA and NDA and the FAI Groups UA and SA, Across all Five Conditions (Mean ± SD).

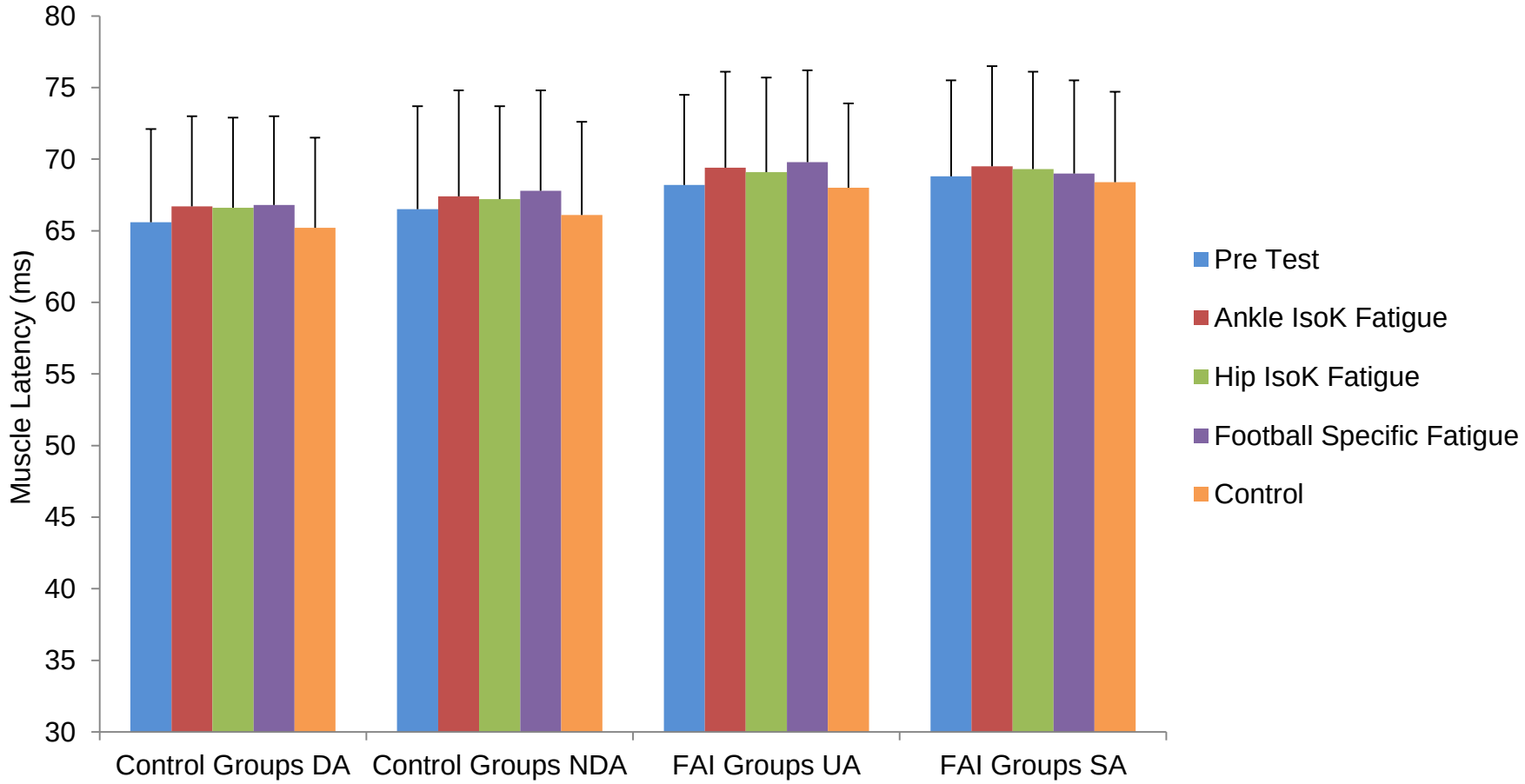


Figure 8.5. Muscle Latencies for the Tibialis Anterior when Acting as a Support Limb, in the Control Groups DA and NDA and the FAI Groups UA and SA, Across all Five Conditions (Mean ± SD).

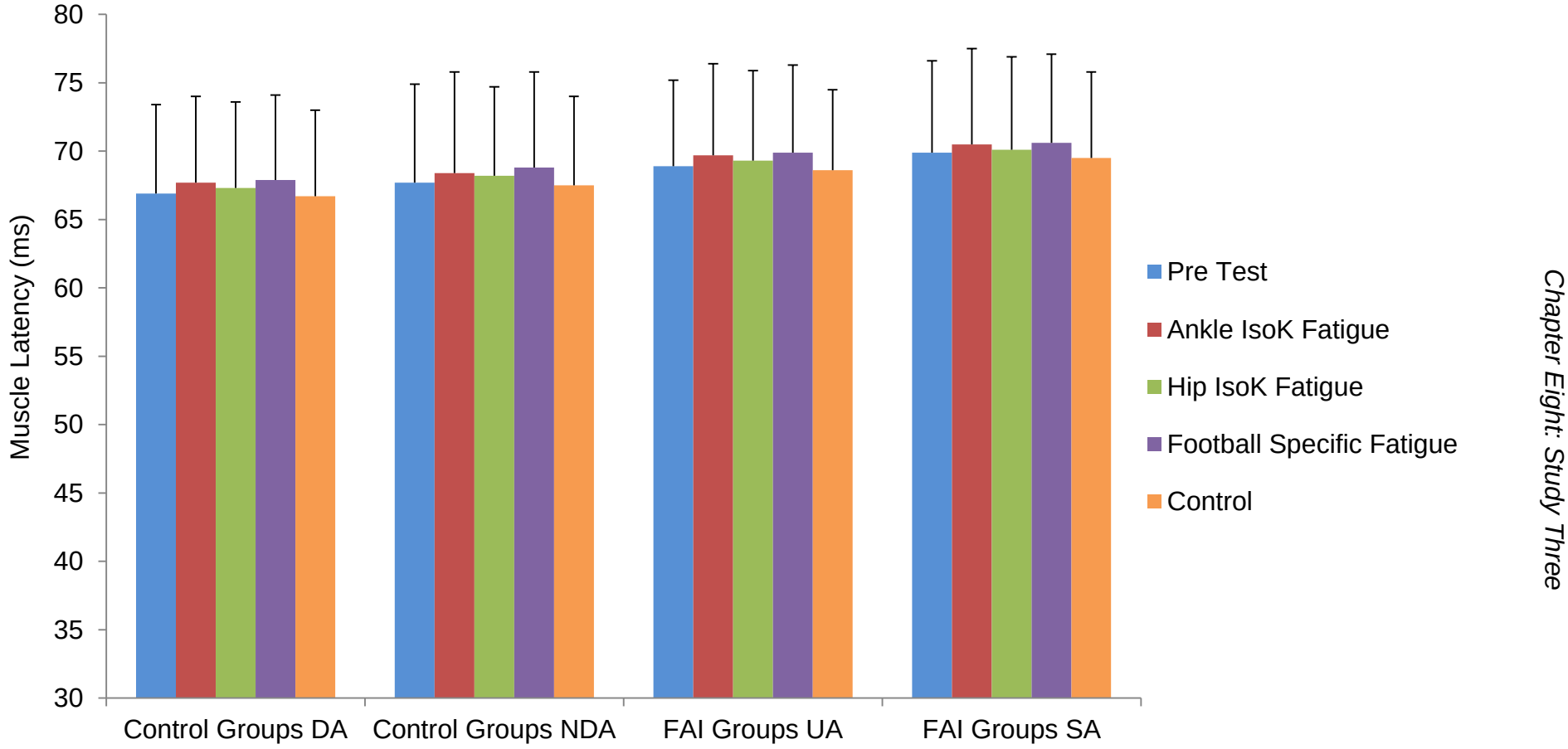


Figure 8.6. Muscle Latencies for the Gluteus Medius when Acting as a Support Limb, in the Control Groups DA and NDA and the FAI Groups UA and SA, Across all Five Conditions (Mean ± SD).

8.1.5 Discussion

The aim of this study was to research muscle latency in the unilateral FAI subject's UA and SA, compared to a healthy control group's DA and NDA, both before and immediately after a) ankle inversion-eversion isokinetic fatigue, b) hip abduction-adduction isokinetic fatigue, c) treadmill exercise simulating football match play, and d) a control. The results of the current study showed that in the tilt limb there was a significant increase (delay) in muscle latency when comparing both the UA and SA of the FAI group to both the DA and NDA of the control group, across all five conditions, in the three muscles tested. Therefore, the hypothesis that the FAI subjects would have increased (delayed) muscle latencies in comparison to the healthy control group, across all conditions was formally accepted.

Again in the tilt limb the fatigue conditions when compared to the pre-test and control conditions showed no significant difference in muscle latency for the three muscles tested, in all ankles. Therefore, the hypothesis that the fatigue protocols will further amplify the effect of increased (delayed) muscle latencies in the FAI group in comparison to the pre test and control conditions was formally rejected.

In the support limb there was no significant difference in muscle latency when comparing both the UA and SA of the FAI group to both the DA and NDA of the control group, across all five conditions, in the three muscles tested. Therefore, the hypothesis that the FAI subjects would have increased (delayed) muscle latencies in comparison to the healthy control group, across all conditions can be formally rejected.

There was also no significant difference in muscle latency when comparing the fatigued conditions to the pre-test and control conditions, for the three muscles tested, in all ankles. Therefore, the hypothesis that the fatigue protocols will further amplify the effect of increased (delayed) muscle latencies in the FAI group in comparison to the pre test and control conditions can be formally rejected.

8.1.5.1 Isokinetic Fatigue

To date, only one previous study has investigated the effect of isokinetic fatigue on muscle latency in response to a tilt platform perturbation. Jackson et al. (2009) hypothesised that isokinetic fatigue would cause an increase (delay) in muscle latency. However, their results showed that isokinetic fatigue lead to a significant decrease (improvement) in muscle latency in the peroneus longus and peroneus brevis muscles. These results contrast with the results from the present study as there was no significant difference in muscle latencies following ankle or hip isokinetic fatigue. Jackson et al. (2009) found no Group x Test interactions, and therefore put their results down to a possible learning effect, in which all subjects became more comfortable on the tilt perturbation device throughout the testing, which resulted in a facilitation of the reflex and therefore an improvement in muscle latency.

A possible reason for the differences in results may be due to the methods used. Jackson et al. (2009) perturbed both limbs in the pre-test, whilst only the dominant (fatigued) leg was tested in the post-test. Jackson et al. (2009) stated that this was to allow post-testing to take place as soon as possible following the fatiguing event. However, this implies that the subjects would therefore know which limb would be

perturbed in the post-test, but not in the pre-test, which may have lead to a condition of increased inhibition in the pre-test, due to the increased uncertainty, resulting in higher (slower) muscle latencies. The present study differed to Jackson et al. (2009) as it perturbed both limbs pre and post-test, therefore, keeping the circumstance of uncertainty in both conditions. This may be a reason why no significant differences were found in muscle latency following both the ankle and hip fatigue protocols.

Cools et al. (2002) also studied muscle latencies, but in the deltoid and the three sections of the trapezius during a sudden downward falling movement of the arm. Their results showed that following isokinetic fatigue there was a significant increase (delay) in muscle latencies in all muscles tested. The present study found no significant differences in muscle latencies following ankle and hip isokinetic fatigue, therefore contrasting with the results from Cools et al. (2002). A reason for the differences in results may be due to the large differences in methods used. Cools et al. (2002) studied the muscles of the shoulder, and only investigated healthy subjects. The isokinetic fatigue protocol was also different to the fatigue protocol in the present study.

The results from the present study would suggest that an ankle and hip isokinetic fatigue protocol has no effect on muscle latencies, in healthy or FAI subjects. It has previously been suggested that muscle fatigue can lead to injury (Davis & Bailey, 1997), as reflected in the increased risk of injury in the second half, especially during the last quarter of the match (Hawkins et al., 2001). However, the results from the present study show that fatigue does not lead to increased (delayed) muscle latencies, and therefore, other factors must be present that lead to this increase in injury rate.

8.1.5.2 Football Specific Fatigue

No study to date has investigated the effect of a football specific fatigue protocol on muscle latencies in response to a tilt platform perturbation. Rahnama et al. (2006) studied the effect of intermittent football specific exercise on EMG activity at various running speeds and found a decrease in muscle activity as a result of fatigue. The results of Rahnama et al. (2006) are consistent with those of Oda and Kida (2001) who found a significant decrease in RMS values of the biceps brachii muscle following maximal concurrent hand grip and elbow extension. The present study found that following a 90 minute football specific treadmill protocol there was no significant difference in muscle latency, when compared to the pre-test and control condition.

The football specific protocol used in this study imposes an unusual constraint on subjects because as they become fatigued they are required to perform the same levels of locomotor activity. In most other studies of fatigue, performance has been evaluated after a fatiguing protocol that uses a self determined effort, typically on isometric equipment. This type of protocol has generally been associated with reduced muscle strength capacity and reduced muscle activation levels (Gabriel & Basford, 2001; Kent-Braun, 1999; Lepers, Maffiuletti, Rochette, Brugniaux & Millet, 2002; Michaut, Pousson, Babault & Van Hoecke, 2002). Under these conditions reduced central drive and failure in contractile properties of a muscle can be directly associated with the reduced muscle strength recorded (Greig et al., 2006). In football, the same phenomenon is observed as the game progresses, with fatigue inhibiting voluntary actions of players causing them to run less far and more slowly. The simulated football protocol used to induce fatigue in the present study has previously been found to reduce muscle strength (Rahnama et

al., 2003), but in the present study did not lead to a significant delay in muscle reaction times. The dilemma in the present study is that the exercise protocol demands that the locomotor activity levels remain the same even when fatigue has developed. Similarly, it is not possible to determine whether the increased (delayed) muscle latencies reflect reduced central drive, changes in excitation-contraction coupling or peripheral factors such as reduced substrate or change in muscle fibre recruitment, or a combination of them all, because these factors were beyond the scope of this research.

It should be noted that even though the treadmill protocol was chosen as it was a more 'sport specific' fatigue protocol, there will still be both kinetic and kinematic variation between overground and treadmill running. Wank, Frick, Schmidtbleicher (1998) observed a shortened stride length, a compensatory higher stride frequency, and lower vertical displacement of the centre of mass in treadmill running. Such alterations will necessitate altered muscular recruitment strategies. However, the subjects in the present study were selected as being familiar with the activity pattern and fully habitualised to the treadmill protocol. Football players, due to the demands of the game, typically have altered kinematics to runners (Wank et al. 1998), making comparisons with studies on distance runners difficult.

As already stated, there was no significant difference in muscle latency following the football specific fatigue protocol in the healthy or FAI subjects. It has previously been found that with fatigue there is a decrease in strength, as measured by isokinetic dynamometry (Rahnama et al., 2003), which would affect the player's ability to perform their skills towards the end of the game when they will be able to run, sprint, jump and tackle less vigorously than they would at the start of the game. Rahnama et al. (2003)

stated that it will also lead to more errors, which will affect a player's susceptibility to injury as the game progresses. It has also been suggested that muscle fatigue can lead to injury (Davis & Bailey, 1997), as reflected in the increased risk of injury in the second half, especially during the last quarter of the match (Hawkins et al., 2001). However, the results from the present study show that fatigue does not lead to significantly increased (delayed) muscle latencies, and therefore, other factors must be present that lead to this increase in injury rate.

8.1.5.3 Clinical Implications

The main clinical implications that have arisen from the findings are that fatigue does not lead to increased (delayed) muscle latencies. Therefore, in terms of muscle latency individuals that participate in sports, as well as sports clinicians and coaches, should not be concerned about the theorised relationship between the onset of fatigue and an increased injury risk at the ankle.

8.1.5.4 Limitations and Recommendations for Future Research

There is an argument to whether or not isokinetic fatigue can simulate the "real life" functional fatigue that occurs during sports participation. However, the benefits of isokinetic protocols are that they are standardised and easily repeatable. The present study also investigated the effect of a football specific protocol on muscle latencies, but due to technical impracticalities backwards movements, sideways movements and actions with the ball were not included. Future studies may wish to investigate muscle

latencies directly following a sports game, to see if ‘real life’ sporting situations have a greater effect on muscle latency than found in the present study.

8.1.6 Conclusion

In summary, the fatigue conditions when compared to the pre-test and control conditions showed no significant difference in muscle latency for all muscles and ankles tested. It has previously been suggested that muscle fatigue may predispose an individual to injury. However, the results from the present study suggest that fatigue does not lead to increased (delayed) muscle latencies, and therefore, factors other than fatigue must be present that lead to this increase in injury rate.

Chapter Nine

The Effect of Localised and Globalised Fatigue on Postural Sway in Healthy versus Functionally Unstable Subjects

9.1 Study Four: The Effect of Localised and Globalised Fatigue on Postural Sway in Healthy and Functionally Unstable Subjects Following a Single Leg Drop Jump Landing

9.1.1 Abstract

Aim: To research postural sway following a single leg drop jump over i) 3 seconds, and ii) 200 ms, in the unilateral FAI group's UA and SA compared to a healthy control group's DA and NDA, both before and immediately after a) ankle inversion-eversion isokinetic fatigue, b) hip abduction-adduction isokinetic fatigue, c) treadmill exercise simulating football match play, and d) a control. **Method:** Twenty males suffering from unilateral FAI and 20 male healthy controls performed 6 single leg drop jumps, 3 on each leg, onto a force platform and remained balanced for 3 seconds. This task was performed both before and immediately after each protocol. **Results:** The results indicated that during the 3 second analysis there was a significant increase (P<0.0125) in lateral and mediolateral sway following each fatigue protocol in the UA and SA of the FAI subjects, in comparison to the DA and NDA of the healthy subjects. During the 200 ms analysis there was a significant increase (P<0.0125) in lateral and mediolateral sway in the FAI subjects UA and SA in the pre-test and control conditions, in comparison to the DA and NDA of the healthy subjects. These findings were further increased under the influence of each fatigue protocol. The football specific fatigue protocol caused the greatest significant increase in medial, lateral and mediolateral sway in both ankles of the FAI and healthy subjects, with the FAI subjects results still being significantly increased (P<0.0125) in comparison to the healthy subjects. **Conclusion:** Clinically, the results show that the fatigued individual may be at greater

risk of ankle inversion injury, especially during more globalised prolonged exercise involving multiple joints, such as a football match.

9.1.2 Introduction

Lateral ankle sprains are one of the most common injuries among athletes (Garrick & Requa, 1989). The disruption of the lateral ligament complex often leads to mechanical instability, peroneal weakness, and a decrease in the neuromuscular control mechanisms about the joint, leaving it particularly susceptible to further injury (Benesch et al. 2000; Fernandes et al., 2000; Hertel, 2000; Konradsen, 2002; Mora et al., 2003). The rate of recurrence has been reported to be as high as 80% among active individuals (Yeung et al., 1994). Recurrent sprains, residual disability, a feeling of "giving way", and a sensation of joint weakness characterise FAI, a condition that often arises secondary to inversion trauma (Beckman & Buchanan, 1995; Fernandes et al., 2000; Konradsen et al., 1998; Konradsen & Ravn, 1991).

Anecdotally, it has been reported that most of these injuries occur at the end of an activity when the participant is fatigued (Hawkins et al., 2001). There appears to be a relationship between muscle fatigue and altered neuromuscular control (Gribble & Hertel, 2004a). One way of quantifying neuromuscular control is through measures of postural stability. Fatigue and deficits in postural control may be predispositions to musculoskeletal injury (Gribble & Hertel, 2004). There is evidence to support a relationship between fatigue and impaired static postural control (Gribble & Hertel, 2004a; Johnston et al., 1998; Lundin et al., 1993; Miller & Bird, 1976). Lundin et al. (1993) found that fatigue to the plantarflexors and dorsiflexors of the ankle created

significant increases in postural sway in the mediolateral direction. Yaggie and McGregor (2002) found similar increases in postural sway when the plantarflexors and dorsiflexors as well as the invertors and evertors of the ankle were fatigued. In contrast to these studies Alderton and Moritz (1996) found no relationship between fatigue to calf musculature and single leg balance 5 and 10 minutes after a continuous heel raising task.

Miller and Bird (1976) investigated performance on a dynamic postural control task following fatigue to the ankle dorsiflexors and plantarflexors, knee and hip flexors and extensors, and abdominals. They found that fatigue to the movers of the knee and hip created significant increases in stabilisation time compared to other muscle groups. The lack of research investigating the effects of fatigue on dynamic postural control tasks warrants further investigation.

The majority of studies examining fatigue have investigated the effects of localised muscle fatigue on postural stability. When muscles have been fatigued locally using an isokinetic dynamometer (50% of the maximal voluntary contraction), subjects showed a loss of stability when attempting to maintain their equilibrium on a balance device (Johnston et al., 1998). However, very few previous studies have attempted to compare the differential effects of fatigue of the ankle and the hip. Winter, Prince, Frank, Powell and Zabjek (1996) has previously explained that the ankle dorsiflexors and plantarflexors play a large role in minimising anteroposterior movements, while the hip abductors and adductors seem to control mediolateral sway of COP. By systematically fatiguing the muscle about the hip and ankle and measuring postural control, it may be

possible to elicit the specific contributions that each joint offers in maintaining postural control.

Nelson and Johnson (1973) examined the effects of local and general muscle fatigue on static balance. Self-reported local and generalised fatigue were induced by performing heel raises and squat thrusts, respectively. Both the general and local fatigue models indicated a decline in static balance, but the generalised mode of fatigue exhibited a greater amount of sway velocity within subjects. More recently, generalised fatigue has been induced through strenuous aerobic physical exercise. Generally, these authors reported a mild effect when vision is available (Lepers, Bigard, Diard, Gouteyron & Guezennec, 1997; Nardone et al., 1998). Bove et al. (2007) used maximal treadmill exercise to induce fatigue and found short-lasting body destabilisation. However, the aerobic physical exercise prescribed is often not specific to a 'real' sporting situation. The effect of more sports specific protocols, such as those employed by Drust et al. (2000) should be investigated.

The aim of this study was therefore to evaluate postural sway following a single leg drop jump over i) 3 seconds, and ii) 200 ms, in the unilateral FAI group's UA and SA compared to a healthy control group's DA and NDA, both before and immediately after a) ankle inversion-eversion isokinetic fatigue (local fatigue), b) hip abduction-adduction isokinetic fatigue (local fatigue), c) football-specific fatigue (global fatigue), and d) a control. It was hypothesised that the FAI subjects would have increased levels of postural sway in comparison to the healthy control group, across all conditions. It was also hypothesised that the fatigue protocols will further increased postural sway in the FAI group in comparison to the pre-test and control conditions.

9.1.3 Method

9.1.3.1 Subjects

The same subjects were used as in Study Three (Section 8.1.3.1).

9.1.3.2 Experimental Design

The same experimental design as Pilot Study Three was used; apart from the force plate sampled at a rate of 200 Hz, and subjects were not required to return to the laboratory seven days later to repeat the procedure (Section 5.2.3.2).

Following postural sway measurements at rest, the subjects randomly undertook the first of four fatigue procedures. Each procedure was performed with seven days in between, to ensure that one procedure did not have an effect on another. The four procedures were a) ankle inversion-eversion isokinetic fatigue, b) hip abduction-adduction isokinetic fatigue, c) football-specific fatigue or d) 105 minutes quiet rest (control).

<u>Ankle Inversion-Eversion Isokinetic Fatigue</u>

The same isokinetic ankle inversion-eversion fatigue protocol was used as in Pilot Study Four (Section 7.2.3.2), apart from the following; a speed of $120° \cdot s^{-1}$was used, and subjects were not required to return to the laboratory for repeat testing. Immediately following the ankle fatigue protocol, three trials of the single leg drop jump were

performed on the right leg (fatigued limb). The fatigue procedure was then repeated on the subjects left limb.

Hip Abduction-Adduction Isokinetic Fatigue

The same isokinetic hip abduction-adduction fatigue protocol was used as in Pilot Study Seven (Section 7.8.3.2), apart from the following; a speed of $120° \cdot s^{-1}$was used, and subjects were not required to return to the laboratory for repeat testing. Immediately following the hip fatigue protocol, three trials of the single leg drop jump were performed on the right leg (fatigued limb). The fatigue procedure was then repeated on the subjects left limb.

Treadmill Exercise Simulating Football Match Play

The same treadmill football-specific fatigue protocol was used as in Study Three (Section 8.1.3.2). Immediately following the football-specific fatigue protocol three single leg drop jumps were performed randomly on each leg, and averages of these were used for analysis

Control - 105 Minutes Rest

The same control procedure was used as in Study Three (Section 8.1.3.2). Immediately following the 105 minutes rest, three single leg drop jumps were performed randomly on each leg and averages of these were used for analysis.

Following the completion of each procedure the subject performed a five minute cool down on the cycle ergometer, at 50 rpm with a resistance of 50 Watts.

9.1.3.3 Data Analysis

The same data analysis as Pilot Study Three was used (Section 5.2.3.3).

9.1.3.4 Statistical Analysis

Firstly, using SPSS (version 19) statistical tests were performed to identify differences between the ankles tested (DA, NDA, UA and SA) in each of the five conditions (pre-test, ankle isokinetic fatigue, hip isokinetic fatigue, football specific fatigue or control) for each of the sway directions, for the 3 second and 200 ms data. Both univariate normality (Shapiro-Wilk) and multivariate normality (Mahalanobis distances) were verified. Linearity was confirmed by generating a matrix of scatterplots between each pair of variables, separately for each group. A multivariate analysis of variance (MANOVA) was used to explore the differences in postural sway between the ankles tested in each of the five conditions for each of the sway directions, for the 3 second and 200 ms data. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of homogeneity of variance-covariance matrices. The Multivariate Test box (Wilk's Lambada value) was studied for significant differences between the ankles ($P<0.05$). The Test of Between-Subject Effects box was then observed to identify differences between the ankles for each condition ($P<0.05$). Tukey's post-hoc test was used to

determine exactly where the significant findings occurred between the ankles. Due to multiple comparisons being made between groups, a Bonferroni adjustment was carried out, and the alpha level was set at $P<0.0125$.

Secondly, statistical tests were performed to identify differences between each of the five conditions in each ankle tested for each of the sway directions, for the 3 second and 200 ms data. Both univariate normality (Shapiro-Wilk) and multivariate normality (Mahalanobis distances) were verified. Linearity was confirmed by generating a matrix of scatterplots between each pair of variables, separately for each group. A multivariate analysis of variance (MANOVA) was used to explore the differences in postural sway between the five conditions in each ankle tested for each of the sway directions, for the 3 second and 200 ms data. The Levene's Test of Equality of Error Variances box was inspected to confirm the assumption of homogeneity of variances across groups. The Box's Test of Equality of Covariance Matrices was also examined to verify the assumption of homogeneity of variance-covariance matrices. The Multivariate Test box (Wilk's Lambada value) was studied for significant differences between conditions ($P<0.05$). The Test of Between-Subject Effects box was then observed to identify differences for each of the fatigue conditions ($P<0.05$). Tukey's post-hoc test was used to determine exactly where the significant findings occurred between the conditions. Due to multiple comparisons being made, a Bonferroni adjustment was carried out, and the alpha level was set at $P<0.01$.

9.1.4 Results

9.1.4.1 3 Second Postural Sway Data

Results from the MANOVA found a significant increase (P<0.0125) in lateral (Table 9.1) and mediolateral (Figure 9.2) sway between the UA in the FAI subjects and both the DA and NDA of the control group, following the ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue protocols. The results also found a significant increase (P<0.0125) in lateral (Table 9.1) and mediolateral (Figure 9.2) sway between the SA in the FAI subjects and both the DA and NDA of the control group, following the ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue protocols. However, there was no significant difference in lateral or mediolateral sway between the UA and SA of the FAI group when compared to both the DA and NDA of the control group during the pre-test and control conditions. The results also showed no significant differences when comparing the sway distances for the anterior, posterior, medial (Table 9.1) and anteroposterior (Figure 9.1) directions when comparing the DA, NDA, UA and SA, across all five conditions.

The MANOVA also found a significant increase (P<0.01) in lateral (Table 9.1) and mediolateral (Figure 9.2) sway when comparing the ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue protocols to the pre-test and control condition in the UA and SA of the FAI subjects, however, no differences were found in the DA and NDA of the control group. The results found no significant difference in sway in the anterior, posterior, medial (Table 9.1) and anteroposterior (Figure 9.1) directions when comparing all five conditions, in all ankles tested.

Table 9.1. Sway Distance (cm) During the 3 Second Analysis for the Control and FAI group, Across all Five Conditions.

Condition	**Control Group**		**FAI Group**	
	DA	**NDA**	**UA**	**SA**
***ANTERIOR SWAY* (cm)**				
Pre Test	7.83 (0.76)	8.32 (0.80)	8.43 (0.86)	7.99 (0.79)
Ankle IsoK Fatigue	7.89 (0.80)	8.25 (0.78)	8.49 (0.92)	8.04 (0.81)
Hip IsoK Fatigue	7.80 (0.78)	8.35 (0.87)	8.51 (0.95)	8.10 (0.86)
Football Specific Fatigue	7.94 (0.83)	8.38 (0.85)	8.52 (0.92)	8.15 (0.91)
Control	7.91 (0.80)	8.30 (0.79)	8.37 (0.83)	8.11 (0.85)
***POSTERIOR SWAY* (cm)**				
Pre Test	12.52 (1.18)	12.76 (1.13)	13.21 (1.28)	12.87 (1.21)
Ankle IsoK Fatigue	12.57 (1.23)	12.70 (1.04)	13.23 (1.32)	12.94 (1.27)
Hip IsoK Fatigue	12.60 (1.31)	12.82 (1.20)	13.29 (1.36)	12.97 (1.32)
Football Specific Fatigue	12.64 (1.42)	12.86 (1.37)	13.26 (1.32)	12.95 (1.23)
Control	12.43 (1.03)	12.67 (1.07)	13.25 (1.34)	12.92 (1.28)
***MEDIAL SWAY* (cm)**				
Pre Test	5.31 (0.50)	5.60 (0.55)	5.62 (0.61)	5.42 (0.53)
Ankle IsoK Fatigue	5.42 (0.60)	5.63 (0.59)	5.68 (0.67)	5.50 (0.58)
Hip IsoK Fatigue	5.44 (0.64)	5.67 (0.63)	5.67 (0.69)	5.53 (0.60)
Football Specific Fatigue	5.48 (0.68)	5.72 (0.67)	5.71 (0.72)	5.57 (0.64)
Control	5.38 (0.56)	5.58 (0.59)	5.68 (0.65)	5.49 (0.57)
***LATERAL SWAY* (cm)**				
Pre Test	6.23 (0.47)	6.32 (0.51)	6.54 (0.58)	6.50 (0.62)
Ankle IsoK Fatigue	6.30 (0.53)	6.38 (0.57)	6.89 (0.71)*†	6.82 (0.67)*†
Hip IsoK Fatigue	6.32 (0.57)	6.36 (0.52)	6.98 (0.78)*†	6.90 (0.76)*†
Football Specific Fatigue	6.29 (0.59)	6.38 (0.56)	7.11 (0.83)*†	7.15 (0.87)*†
Control	6.29 (0.51)	6.35 (0.54)	6.50 (0.53)	6.52 (0.57)

Results presented as Mean (SD). * Significantly (P<0.01) higher than pre-test and control condition. † Significantly (P<0.0125) higher than the control groups DA and NDA.

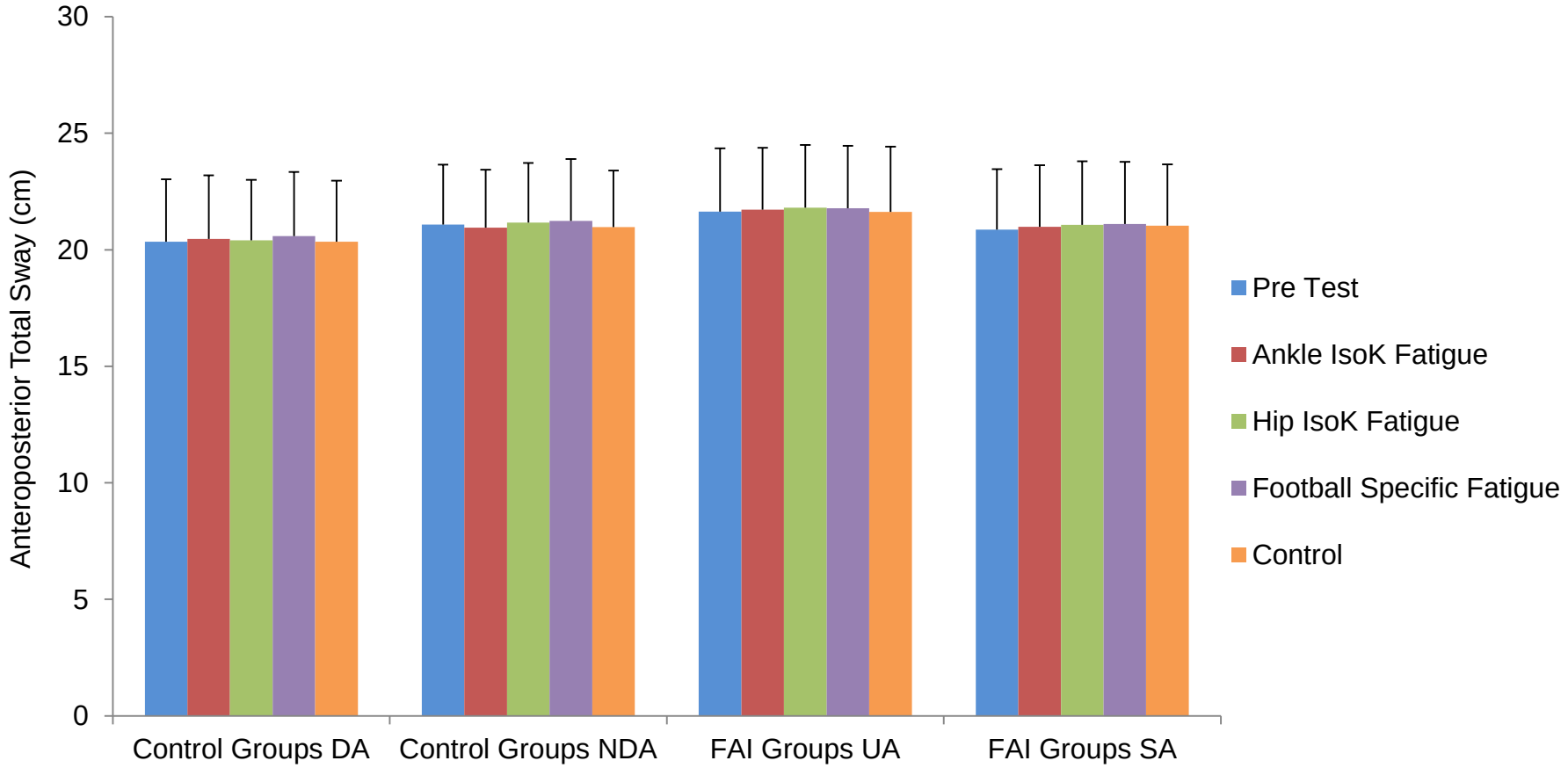


Figure 9.1. Anteroposterior Sway Distance During the 3 Second Analysis for the Control Group's Dominant and Non-Dominant Ankle and the Functional Ankle Instability Group's Unstable and Stable Ankle, Across all Five Conditions (Mean ± SD).

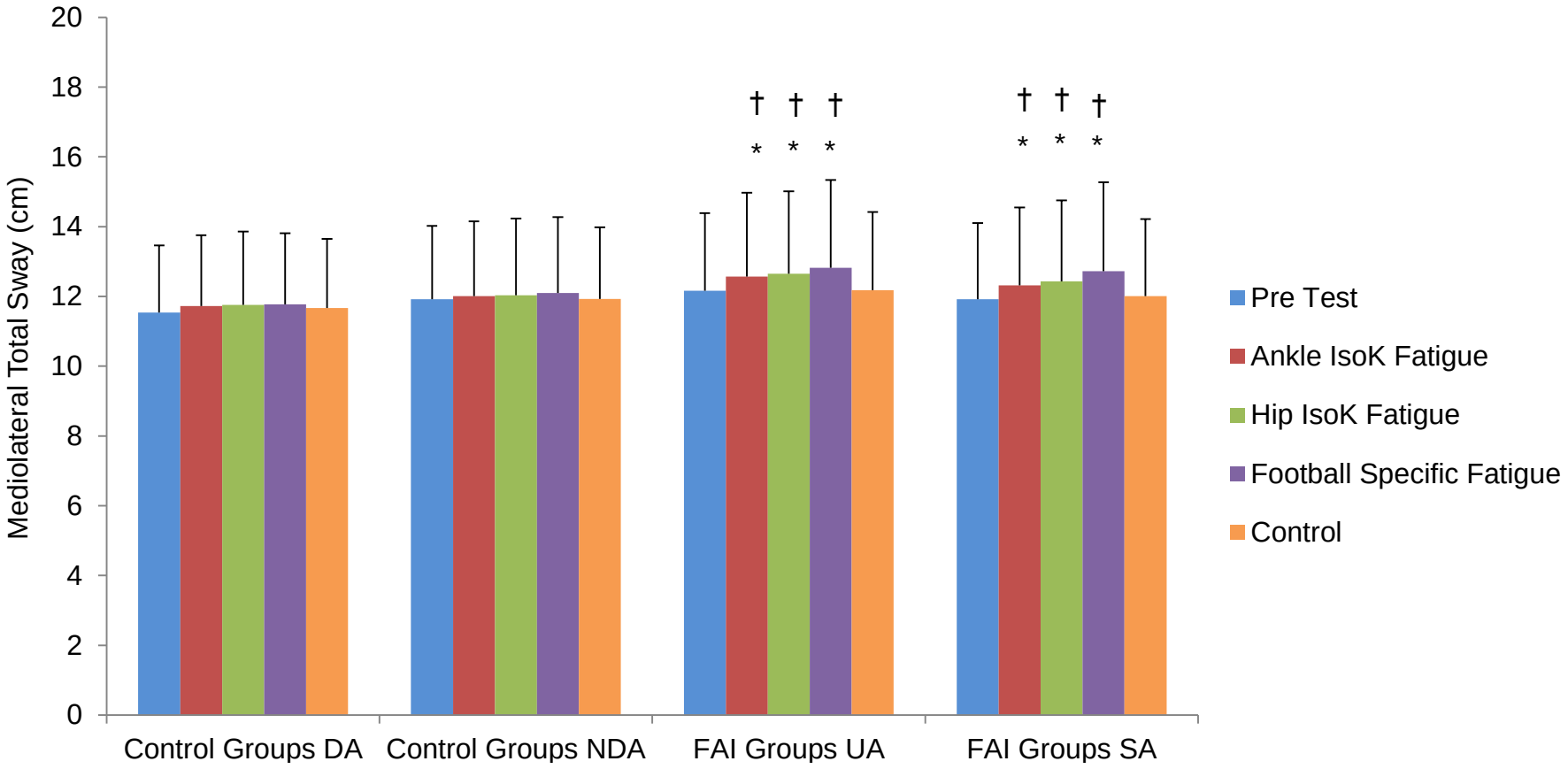


Figure 9.2. Mediolateral Sway Distance During the 3 Second Analysis for the Control Group's Dominant and Non-Dominant Ankle and the Functional Ankle Instability Group's Unstable and Stable ankle, Across all Five Conditions (Mean ± SD). * Significantly (*P*<0.01) higher than corresponding pre-test and control condition. † Significantly higher (*P*<0.125) than control groups DA and NDA.

9.1.4.2 200 ms Postural Sway Data

Results from the MANOVA found a significant increase (P<0.0125) in lateral (Table 9.2) and mediolateral (Figure 9.4) sway between the UA in the FAI subjects and both the DA and NDA of the control group, across all five conditions. The results also found a significant increase (P<0.0125) in lateral (Table 9.2) and mediolateral (Figure 9.4) sway between the SA in the FAI subjects and both the DA and NDA of the control group, across all five conditions. In addition, the results showed no significant differences when comparing the sway distances for the anterior, posterior, medial (Table 9.2) and anteroposterior (Figure 9.3) directions when comparing the DA, NDA, UA and SA, across all five conditions.

The MANOVA results also found a significant increase (P<0.01) in lateral (Table 9.2) and mediolateral (Figure 9.4) sway when comparing the ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue protocols to the pre test and control conditions in the UA and SA of the FAI subjects. The UA and SA of the FAI group also had a significant increase (P<0.01) in medial sway when comparing the football specific fatigue protocol to the pre test and control conditions. In addition, there was a significant increase (P<0.01) in medial, lateral (Table 9.2) and mediolateral (Figure 9.4) sway when comparing the football specific protocol to the pre test and control conditions in the DA and NDA of the control group. The results found no other significant differences in any other sway direction (Table 9.2, Figure 9.3 and 9.4).

Table 9.2. Sway Distance (cm) During the 200 ms Analysis for the Control and FAI group, across all five conditions.

Condition	**Control Group**		**FAI Group**	
	DA	**NDA**	**UA**	**SA**
***ANTERIOR SWAY* (cm)**				
Pre Test	3.31 (0.22)	3.23 (0.18)	3.38 (0.24)	3.34 (0.21)
Ankle IsoK Fatigue	3.38 (0.26)	3.36 (0.22)	3.43 (0.26)	3.41 (0.19)
Hip IsoK Fatigue	3.40 (0.25)	3.38 (0.23)	3.44 (0.21)	3.39 (0.23)
Football Specific Fatigue	3.43 (0.28)	3.40 (0.27)	3.47 (0.28)	3.43 (0.24)
Control	3.35 (0.21)	3.29 (0.20)	3.40 (0.26)	3.37 (0.24)
***POSTERIOR SWAY* (cm)**				
Pre Test	5.13 (0.38)	4.93 (0.32)	5.20 (0.42)	5.22 (0.40)
Ankle IsoK Fatigue	5.21 (0.42)	5.03 (0.37)	5.28 (0.44)	5.24 (0.43)
Hip IsoK Fatigue	5.24 (0.41)	5.10 (0.39)	5.30 (0.47)	5.28 (0.45)
Football Specific Fatigue	5.26 (0.46)	5.13 (0.44)	5.33 (0.51)	5.29 (0.42)
Control	5.21 (0.40)	5.03 (0.35)	5.19 (0.42)	5.15 (0.38)
***MEDIAL SWAY* (cm)**				
Pre Test	1.63 (0.09)	1.64 (0.12)	1.72 (0.16)	1.70 (0.14)
Ankle IsoK Fatigue	1.67 (0.11)	1.69 (0.13)	1.78 (0.17)	1.76 (0.16)
Hip IsoK Fatigue	1.70 (0.14)	1.72 (0.13)	1.80 (0.21)	1.78 (0.18)
Football Specific Fatigue	1.87 (0.19)*	1.85 (0.20)*	2.10 (0.28)*	2.04 (0.25)*
Control	1.64 (0.11)	1.66 (0.14)	1.73 (0.17)	1.67 (0.12)
***LATERAL SWAY* (cm)**				
Pre Test	2.63 (0.21)	2.72 (0.19)	4.43 (0.23)†	4.38 (0.21)†
Ankle IsoK Fatigue	2.71 (0.25)	2.76 (0.22)	4.78 (0.34)*†	4.75 (0.31)*†
Hip IsoK Fatigue	2.75 (0.28)	2.79 (0.26)	4.83 (0.37)*†	4.81 (0.35)*†
Football Specific Fatigue	2.85 (0.34)*	2.88 (0.37)*	4.93 (0.45)*†	4.90 (0.42)*†
Control	2.68 (0.20)	2.79 (0.22)	4.40 (0.25)†	4.44 (0.27)†

Results presented as Mean (SD). * Significantly (P<0.01) higher than pre-test and control conditions. † Significantly (P<0.0125) higher than the control groups DA and NDA.

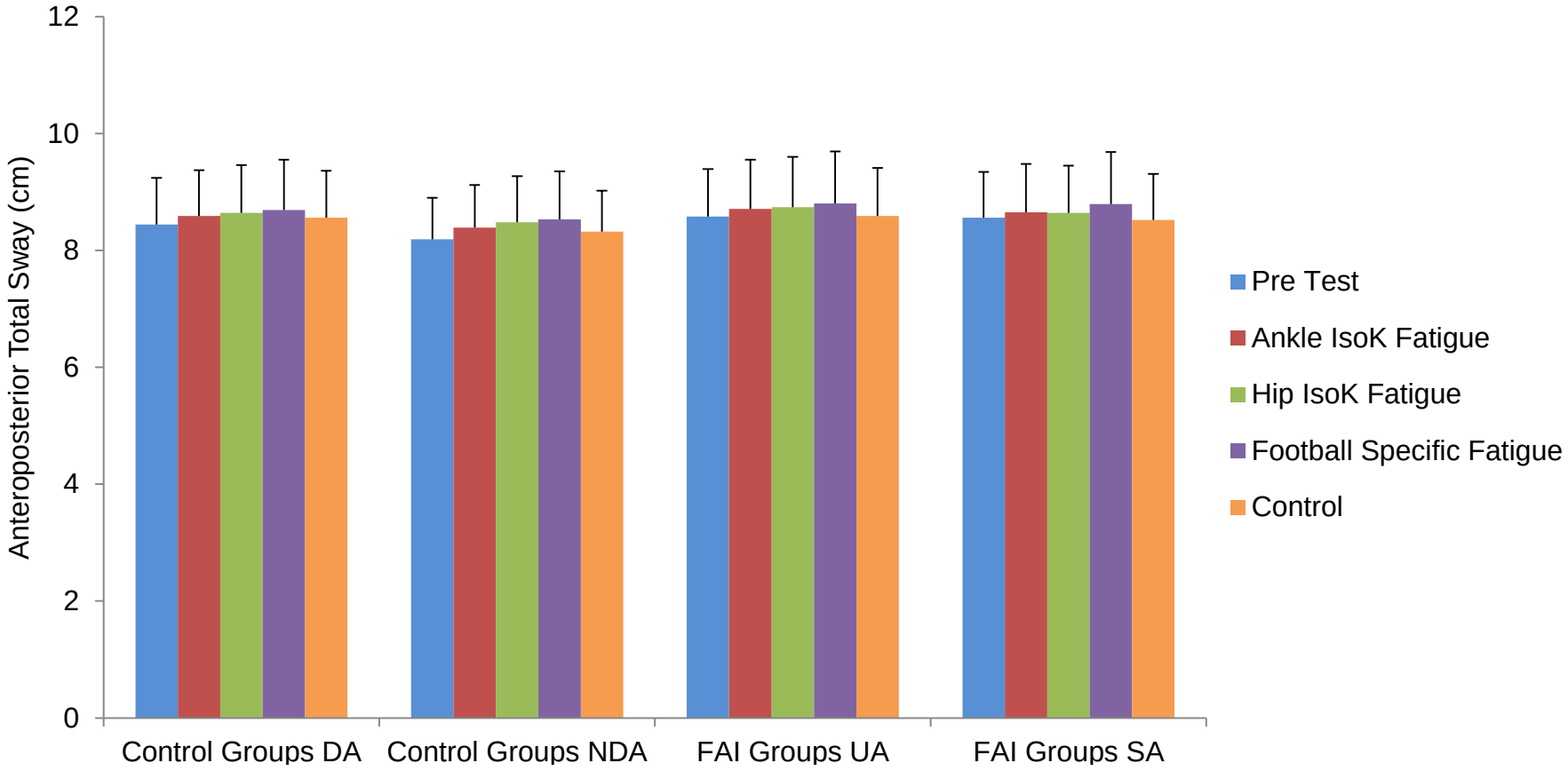


Figure 9.3. Anteroposterior Sway Distance During the 200 ms Analysis for the Control Group's Dominant and Non-Dominant Ankle and the Functional Ankle Instability Group's Unstable and Stable Ankle, Across all Five Conditions (Mean ± SD).

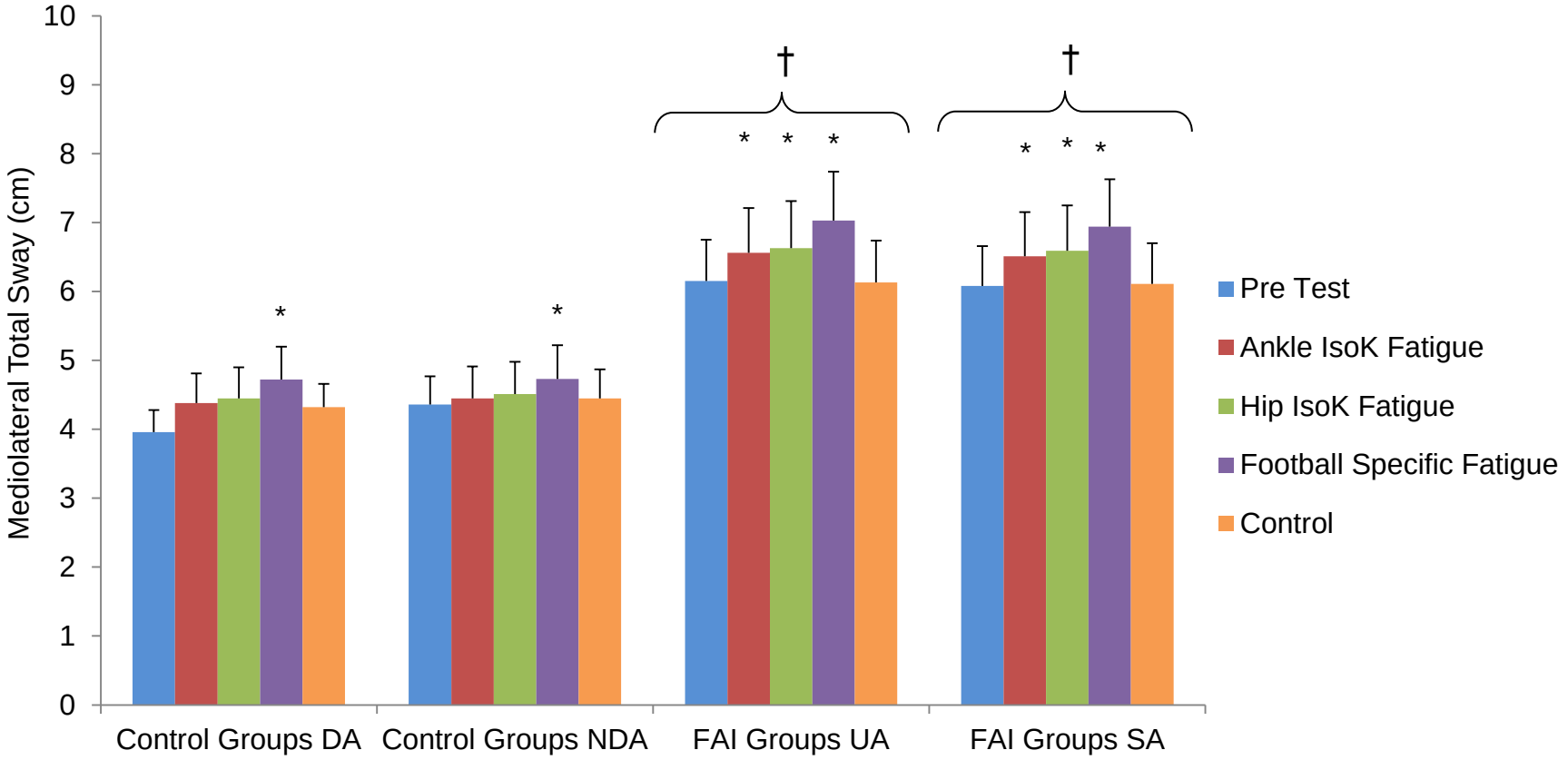


Figure 9.4. Mediolateral Sway Distance During the 200 ms Analysis for the Control Group's Dominant and Non-Dominant Ankle and the Functional Ankle Instability Group's Unstable and Stable Ankle, Across all Five Conditions (Mean ± SD). * Significantly (*P*<0.01) higher than corresponding pre-test and control condition. † Significantly higher (*P*<0.0125) than control groups DA and NDA for each corresponding condition.

9.1.5 Discussion

The aim of this study was to research postural sway following a single leg drop jump over a) 3 seconds, and b) 200 ms, in the unilateral FAI group's UA and SA compared to a healthy control group's DA and NDA, both before and immediately after 1) ankle inversion-eversion isokinetic fatigue, 2) hip abduction-adduction isokinetic fatigue, 3) football-specific fatigue, and 4) a control. In regards to the 3 second analysis results a significant increase in lateral and mediolateral sway was found between the UA and SA of the FAI subjects and both the DA and NDA of the control group, following the ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue protocols. Therefore, the hypothesis that the FAI subjects would have increased levels of postural sway in comparison to the healthy control group, across all conditions was partially accepted, as the increased postural sway was only found following the fatigue protocols, and not during the pre test or control conditions.

Results from the 3 second analysis also found a significant increase in lateral and mediolateral sway when comparing the ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue protocols to the pre test and control condition in the UA and SA of the FAI subjects. Therefore, the hypothesis that the fatigue protocols will further increase the effect of elevated postural sway levels in the FAI group in comparison to the pre test and control conditions was formally accepted.

In regards to the 200 ms analysis results a significant increase in lateral and mediolateral sway was found between the UA and SA of the FAI subjects and both the DA and NDA of the control group, across all five conditions. Therefore, the hypothesis

that the FAI subjects would have increased levels of postural sway in comparison to the healthy control group, across all five conditions was formally accepted.

Results from the 200 ms analysis also found a significant increase in lateral and mediolateral sway when comparing the ankle isokinetic fatigue, hip isokinetic fatigue and football specific fatigue protocols to the pre test and control conditions in the UA and SA of the FAI subjects. The UA and SA of the FAI group also had a significant increase in medial sway when comparing the football specific fatigue protocol to the pre test and control conditions. Therefore, the hypothesis that the fatigue protocols will further increase the effect of elevated postural sway in the FAI group in comparison to the pre test and control conditions was formally accepted. In addition, there was a significant increase in medial, lateral and mediolateral sway when comparing the football specific protocol to the pre test and control conditions in the DA and NDA of the control group.

9.1.5.1 Isokinetic Fatigue

The results from the present study suggest that there is an effect of localised fatigue of the frontal plane movers of the lower extremity on COP in the lateral and mediolateral direction in the FAI subjects. Whilst isokinetic fatigue to both muscle groups led to significant increases in postural control in the lateral and mediolateral directions, fatigue to the hip abductors and adductors created higher COP excursions compared to fatigue of the ankle invertors and evertors. The results of Gribble and Hertel (2004a) are comparable with the results of the present study as they found that COP excursion velocity was significantly increased following both hip and ankle fatigue. Gribble and

Hertel (2004a) also found that the hip fatigue protocol produced higher COP excursion velocities compared to the ankle fatigue protocol. Similarly, Miller and Bird (1976) found that fatigue to the proximal musculature of the hip and knee produced greater deficits in postural control compared to fatigue of the ankle musculature. The results from these studies, as well as the present study show that maintenance of upright stance in a fatigued state may rely more on proximal neuromuscular control than on the previously accepted ankle strategy of distal muscle recruitment in maintaining postural control in young populations.

Gribble and Hertel (2004a) explained that the muscles controlling the hip have larger cross sectional areas compared to muscles surrounding the ankle. It is inherent that the larger, more proximal musculature has the ability to create stronger contractions but with potential of less efficiency of corrective contractions during single-leg stance compared to the ankle (Gribble & Hertel, 2004a). During a fatigued state, it is possible that efficiency of compensatory muscle firing about the hip during a single-leg stance is reduced such that maintenance of single-stance is substantially impaired.

This phenomenon may also be occurring at the ankle. Under normal conditions as described in the ankle strategy of maintaining postural control (Nashner, Woollacott & Tuma, 1979), conduction of afferent and efferent systems about the ankle complex creates rapid compensatory contractions for maintaining the body's upright position over the fixed foot position in bilateral stance. As with the hip, the efficiency of this maintenance system may be affected negatively by fatigue. However, because the fatigue was taking place at a more distal joint, the slowed conduction of feedback systems and reduced muscle contraction rates and amplitudes may have resulted in

fewer or smaller compensatory contractions and was displayed as a smaller lateral and mediolateral displacement of COP.

Few researchers have investigated the effect that FAI and fatigue have on postural control collectively, especially dynamic postural control tasks. Gribble et al. (2004) used the SEBT as a measure of dynamic postural control and found that FAI subjects displayed smaller reach distance values and knee flexion angles for all reach directions compared with the uninjured side and the healthy group. The effect of fatigue also increased this trend. Even though the methods used by Gribble et al. (2004) differ to those used in the present study, our results agree with the findings of Gribble et al. (2004) as the FAI subjects showed increased levels of lateral and mediolateral sway in the pre test and control conditions during the 200 ms analysis, and these deficits were increased under the influence of fatigue. During the 3 second analysis there were no differences in postural sway between the FAI and healthy group during the pre test and control condition, however, following the isokinetic hip and ankle fatigue protocols deficits became present in the lateral and mediolateral directions. From the 3 second results it seems that during the pre test and control condition the FAI subjects are able to control their postural sway, however, following fatigue this ability is compromised, possibly due to pathological changes associated with FAI.

In the present study it is interesting to observe that the postural sway deficits in the FAI subjects only occur in the lateral, medial and mediolateral planes following the isokinetic fatigue protocols. The anterior, posterior and anteroposterior directions are on the other hand unchanged. This might be explained by the fact that during single-limb stance, the ankle strategy more efficiently controls anteroposterior than mediolateral sway, simply

because the foot is longer than it is wide (Baier & Hopf, 1998). The foot's narrow base of support makes it necessary to use the hip strategy to control substantial mediolateral disturbances of balance, whereas ankle movements can only achieve fine tuning of mediolateral sway (Baier & Hopf, 1998). Individuals with FAI have been shown to use more of a hip strategy to maintain unilateral stance (Hertel, 2002). This alteration in postural control strategy is possibly due to changes in central neural control that occur in the presence of ankle joint dysfunction (Hertel, 2002). It may be that the healthy subjects in the present study had the ability to compensate for the induction of fatigue; however, the FAI subjects lacked this ability due to deficits associated with their pathology.

The results of the present study found bilateral deficits in neuromuscular control in the FAI subjects. With the 200 ms analysis these bilateral deficits were present in the pre test and control conditions, with these deficits being increased under fatigued conditions. With the 3 second analysis there was no significant differences in sway distance between the healthy and FAI subjects in the pre test and control conditions, however, following the fatigue protocols these bilateral deficits were present in the FAI group. Evans et al. (2004), Hiller et al. (2007) and Tropp et al. (1984) found that subjects with FAI did not differ in unilateral stance abilities on the injured versus the uninjured ankles. However, a comparison of both limbs in the subjects with FAI with a healthy control group revealed significantly higher centre of pressure excursions in the lateral direction. Tropp et al. (1984) explained that this may indicate FAI affects the postural control system at a level that is high enough to influence stability on either extremity, or possibly a genetic predisposition to FAI in some individuals. Therefore, the results of the pre test 200 ms data may show that on a subconscious level the subjects

with functionally unstable ankles may have a predisposition to FAI, as evidenced by the decreased performance on the contra-lateral healthy limb, or that FAI affects the postural control system at a central level which may influence stability during stance on either extremity.

In addition to the FAI results, the present study found that when inducing hip or ankle isokinetic fatigue there was no change in postural sway measures in any direction in the healthy subjects DA and NDA. These results agree with the findings of Corbeil, Blouin, Begin, Nougier and Teasdale (2003) who induced fatigue of the plantarflexors and dorsiflexors in healthy males. Corbeil et al. (2003) theorised that even following fatigue, the postural control system was able to maintain the amplitude of the COP oscillations within the same physical limits of the base of support than that observed without fatigue. In addition to this, Wikstrom et al. (2004) failed to observe changes in time to stabilisation under fatigued conditions.

9.1.5.2 Football Specific Fatigue

The present study found that the football specific fatigue protocol created significant increases in lateral, medial and mediolateral sway in the FAI subjects UA and SA as well as the healthy subjects DA and NDA during the 200 ms analysis. This protocol also caused deficits in postural sway in the lateral and mediolateral directions in the FAI subjects UA and SA during the 3 second analysis. The football specific fatigue caused the highest sway excursions in the mentioned directions, which were higher than those caused by the isokinetic hip and ankle fatigue protocols, and significantly higher that the pre test and control conditions. In comparison to the isokinetic fatigue protocols, the

football specific protocol involved multiple joints, much larger muscle mass and a much longer fatigue protocol, which may explain the increased sway excursions, and the deficits present in the healthy subjects as well as the FAI subjects during the 200 ms analysis.

Similar to the results of the present study many generalised methods of inducing fatigue have been found to negatively effect postural stability. Protocols such as the Wingate exercise test (Yaggie & Armstrong, 2004), aerobic yo-yo test (Fox et al. 2008), a 25 minute treadmill run (Nardone et al. 1998) and maximal treadmill exercise (Bove et al. 2007), have all resulted in increased levels of postural sway. However, comparisons with past literature have to be made with caution as the methods of inducing fatigue in the above studies differ greatly to those used in the present study.

Unfortunately it is not possible to determine the exact mechanism behind the increase in postural sway following the fatigue protocols. Many mechanisms of fatigue have been proposed over the years (Hunter & Enoka, 2003, Kanehisa et al., 1995; Sahlin et al., 1998; Singh et al., 2005; Taylor et al., 2000; Westerblad & Allen, 2002). Central mechanisms of fatigue include factors such as reduced central drive (Taylor et al., 2000), decreased muscle spindle excitability (Singh et al., 2005), desensitisation of the motor neurons (Kernell, 1969) and changes in excitation-contraction coupling (Edwards, Hill, Jones & Merton, 1977). Peripheral factors include the accumulation of metabolites (Astrand, 1960; Spagenburg et al., 1998) and changes in muscle fibre recruitment. However, there may not be one distinct mechanism that was responsible for the increases in postural sway in the present study, but a combination of them all, however these factors were beyond the scope of this research.

9.1.5.4 Clinical Implications

Clinically, these results show that exercise fatiguing the proximal hip joint has a greater effect on postural sway, than fatigue of the more local ankle muscular. Clinicians and sports coaches should be aware of this factor in case training sessions involve training of the more proximal joints. They may assume that individuals with ankle instability may be unaffected by this, but these results show that hip fatigue has a greater affect on postural sway, and therefore, a higher probability of causing repeated injury. In addition to this, if pre-season screening identifies these more proximal deficits, prehabilitation involving the gluteus medius muscle in individuals with FAI, may prove to be beneficial and reduce the probablility of repeated sprains throughout the season.

The results also found that the football-specific fatigue caused the greatest deficits in postural sway. This highlights that the fatigued individual may be at a greater risk of musculoskeletal injury, especially during prolonged exercise that involves multiple joints, such as a football match. Therefore, during the early stages of rehabilitation it is important that steps are taken to help prevent muscle fatigue. As the rehabilitation progresses, players suffering from FAI need to be gradually advanced through this prolonged multi-joint exercise, to ensure that they are ready for return to play.

9.1.5.5 Limitations and Recommendations for Future Research

There is an argument to whether or not isokinetic fatigue can simulate the “real life” functional fatigue that occurs during sports participation. However, the benefits of isokinetic protocols are that they are standardised and easily repeatable. The present

study also investigated the effect of a football-specific protocol on muscle latencies, but due to technical impracticalities backwards movements, sideways movements and actions with the ball were not included. Future studies may wish to investigate postural sway directly following a sports game, to see if 'real life' sporting situations have an even greater effect on postural sway than found in the present study.

9.1.6 Conclusion

Fatigue of both the hip abductors and adductors and the ankle invertors and evertors produced a significant increase in lateral and mediolateral sway, in both limbs of the FAI subjects. The deficits were greater following the hip fatigue protocol, possibly due to the proximal muscles having a larger mass, and therefore, reducing their ability to perform as effectively, which had a larger impact on postural sway. The results of this study demonstrate ecological validity as the football-specific fatigue protocol had a greater effect on postural sway than either of the localised hip or ankle isokinetic fatigue protocols. The globalised football specific fatigue protocol caused significantly increased lateral, medial and mediolateral sway in the FAI subjects UA and SA as well as the healthy subjects DA and NDA during the 200 ms analysis. The football specific protocol also caused deficits in postural sway in the lateral and mediolateral directions in the FAI subjects UA and SA during the 3 second analysis. Possible reasons for sway deficits being greatest following the global fatigue protocol may be due to the protocol involving multiple joints, much larger muscle mass and a much longer fatigue protocol. Clinically, these results show that the fatigued individual may be at a greater risk of musculoskeletal injury, especially during prolonged exercise that involves multiple joints.

Chapter Ten

Discussion

This thesis had four main aims which were investigated during the four main studies. The four main aims of this thesis were 1) to evaluate muscle latency in FAI subject's compared to healthy controls, 2) to evaluate single limb postural sway in FAI subject's compared to healthy controls, 3) to research muscle latency in FAI subject's compared to healthy controls, both before and immediately after localised and globalised fatigue protocols, and 4) to research single limb postural sway in FAI subject's compared to healthy controls, both before and immediately after localised and globalised fatigue protocols.

In order to explore these aims it was hypothesised that the FAI subjects would have significantly increased (delayed) muscle latencies in comparison to the healthy controls. It was also hypothesised that the FAI subjects would have significantly increased levels of postural sway in comparison to the healthy controls. In terms of the fatigue interventions it was hypothesised that the fatigue protocols would further increase the effect of delayed muscle latencies in the FAI subjects, in comparison to the healthy controls. It was also hypothesised that the fatigue protocols would further increase the effect of greater postural sway in the FAI subjects, in comparison to the healthy controls.

Within this discussion, the aims and findings of the four main studies will be reviewed in reflection of the research undertaken. The clinical implications, contributions to the literature, limitations and recommendations for future research are also discussed.

10.1 Study One: Muscle Latencies in Healthy and Functionally Unstable Subjects During an Unexpected Plantarflexion and Inversion Tilt Perturbation

The aim to evaluate muscle latencies in FAI subjects versus healthy controls was addressed in Study One. The muscle latencies of the peroneus longus, tibialis anterior and gluteus medius muscles were determined using a tilt platform perturbation. Before conducting Study One a thorough review of the literature indicated FAI subjects had often been found to have increased (delayed) peroneal muscle latencies in comparison to healthy controls (Beckman & Buchanan, 1995; Konradsen & Ravn, 1990; Mitchell et al., 2008a). It has been stated that weakness in a proximal stabilising muscle, such as the gluteus medius, may produce deviations in joint motion, a subsequent loss of stability and may contribute towards a repeated injury at the ankle (Friel et al., 2006; Riemann, 2002). However, there were limited studies investigating the role of the gluteus medius muscle in FAI subjects (Beckman & Buchanan, 1995). Further to this there were limited studies investigating the role of the support limb during a tilt perturbation (Mitchell et al., 2008a). No previous studies were found that measured muscle latency of both the tilt limb and support limb in the peroneus longus, tibialis anterior and gluteus medius muscles of healthy versus FAI subjects.

A common concern with studies using EMG protocols is the reliability of the analysis procedure. Therefore, the aim of Pilot Study One and Two was to determine the relative and absolute reliability of the EMG analysis technique to be used in Study One and Study Three. The onset of EMG is one of the most common parameters evaluated; however, no standard method of determination of this parameter is used in the literature. Pilot Study One addressed the issue of sampling rates, analysis methods

and smoothing levels. It was found that the most reliable combination that provided ICC's above 0.80 and low levels of SEM variance across all conditions was 1000 Hz sampling rate, RMS analysis method and 2 ms smoothing level. Therefore this combination was deemed suitable for use in Study One and Study Three of the thesis. Pilot Study Two addressed the issue of baseline times, deviation levels and the number of samples exceeding the threshold. The results of Pilot Study Two highlighted the combination that provided ICC's above 0.80 and low levels of SEM variance across all conditions was 50 ms baseline, 3 SD level, 50 ms exceeding the threshold. Therefore, this combination was deemed suitable for use in Study One and Study Three of this thesis.

Study One researched muscle latency of the peroneus longus, tibialis anterior and gluteus medius muscles in unilateral FAI subject's UA and SA, compared to a healthy control group's DA and NDA, when acting as i) a tilt limb, and ii) a support limb. The results of Study One showed that when analysing the tilt limb there was a significant increase (delay) in muscle latencies of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to the DA and NDA of the control group. These results are in line with previous studies that have identified increased (delayed) muscle latencies in FAI subjects compared to healthy controls when studying the tilt limb (Beckman & Buchanan, 1995; Konradsen & Ravn, 1990; Mitchell et al., 2008a). When analysing the support limb there was no significant difference in muscle latencies of the peroneus longus, tibialis anterior and gluteus medius when comparing the UA and SA of the FAI group to the DA and NDA of the control group. Again, these results are in agreement with previous studies that have found no significant difference in muscle latency in FAI subjects compared to healthy

controls when studying the support limb (Beckman & Buchanan, 1995). As muscle latencies were increased (delayed) in both the UA and SA of the FAI subjects, when compared to the DA and NDA of the healthy control group this result offers two interpretations: 1) the patients with unilateral FAI may have a predisposition to FAI, as evidenced by the increased (delayed) muscles latencies on the contra-lateral stable limb; and 2) FAI affects muscle latencies at a central level that is high enough to influence stability during stance on either extremity.

10.1.1 Clinical Implications

The main clinical implications that have arisen from the findings of Study One are that any rehabilitation prescribed by sports injury professionals to subjects with unilateral FAI should ensure the exercises focus on both the UA and the SA, as deficits were present in both limbs of the FAI subjects. Study One also found that the deficits in the FAI subjects did not only exist in the muscles surrounding the ankle joint, but were also present in the more proximal gluteus medius muscle. This finding indicates that sports injury professionals should include rehabilitation exercises for the gluteus medius muscle, as well as the ankle musculature in FAI sufferers.

There is also relevance of the reliability findings of Pilot Studies One and Two, which lie predominantly in the research domain. It may be argued that the increase in reliability was marginal between some of the analysis combinations used. However, in the field of research where reliable protocols are a necessity, the 1000 Hz sampling rate, RMS analysis method and 2 ms smoothing level combination, and the 50 ms baseline, 3 SD level, 50 ms exceeding the threshold combination improved the reliability of the

protocol, and should be considered over the other combinations in future research that utilise the same methods.

10.1.2 Contributions to the Literature

The findings of Study One have contributed to the literature in a number of ways. This study is one of the first to investigate muscle latencies in FAI subjects versus healthy controls, in both the tilt limb and the support limb, but also examine more proximal muscles such as the gluteus medius, as well as the more commonly investigated peroneus longus and tibialis anterior muscles. By examining all these factors in one study a greater understanding of the possible deficits contributing towards FAI can be formed. In summary, the results of Study One found that in unilateral FAI subjects bilateral deficits were present, which indicates a more central processing problem, and in addition to this more proximal muscles such as the gluteus medius are affected in FAI sufferers. These results may indicate that FAI subjects are either genetically or biomechanically predisposed to ankle sprains, as it appears that these deficits may not be as a result of suffering a sprain, but may exist prior to the injury being sustained, as indicated by the finding of delayed muscle latencies on the SA as well as the UA in the FAI sufferers.

10.1.3 Limitations and Recommendations for Future Research

Injuries rarely occur with a person standing at rest. However, in the literature it has been stated that to make comparisons there has to be standardisation (Lynch et al., 1996). The use of a tilt platform is a fairly static task. Future research should investigate

muscle activity during activities such as walking, running or jumping to see if greater deficits occur in more dynamic situations.

It has often been stated in the literature that any deficits are exacerbated under the influence of fatigue (Gribble, Hertel et al., 2007). Some would suggest that fatigue may play a role in contributing to the occurrence of lateral ankle sprains (Gutierrez et al., 2007). Research on elite soccer players has shown that injury risk is highest in the last 15 minutes of the contest (Rahnama et al., 2002), when fatigue has set in. Further research should investigate the effect of lower extremity fatigue on muscle latencies in subjects with FAI to see if any further deficits are identified. This final aim was investigated during Study Three, and will be evaluated later in this discussion section.

10.2 Study Two: Postural Sway in Healthy and Functionally Unstable Subjects Following a Single Leg Drop Jump Landing

The aim to evaluate postural sway in FAI subjects versus healthy controls was addressed in Study Two. Postural sway in the anterior, posterior, anteroposterior, medial, lateral and mediolateral directions were determined using an AMTI force platform. Before conducting Study Two a thorough review of the literature indicated that FAI subjects often showed increased postural sway measures in comparison to healthy controls (Evans et al., 2004; Hiller et al., 2007; Tropp et al., 1984). However, there were limited studies that had investigated postural sway following more dynamic tasks (Docherty, Valovich-McLeod et al., 2006; Ross & Guskiewicz, 2004), such as a single leg drop jump. It was also identified that many authors do not provide a rationale for the balance time used in their studies (Fu & Hui-Chan, 2005; McGuine et al., 2000; Trojian

& McKeag, 2006), and often the long duration of balancing time is not specific to a 'real' sporting situation. No study to date has analysed postural control in a subconscious time period (initial 200 ms). The 200 ms time period was identified by Wilkinson and Allison (1989) to be the average fastest reaction time in 20-29 year olds, therefore, anything prior to 200 ms would be beyond human conscious control. Analysis of this subconscious time period may identify postural sway deficits that are sometimes not present in FAI subjects when analysing a conscious time period.

A concern with studies that use force platforms is the reliability of the data produced. Therefore, the aim of Pilot Study Three was to determine the relative and absolute reliability of the postural sway data that was to be used in Study Two and Four. It is essential that the methods used to assess postural stability are determined as reliable in order to evaluate the extent of balance impairment. Pilot Study Three addressed the issue of sampling rates and balance duration on the force platform following a single leg drop jump. The results from Pilot Study Three highlighted two combinations that provided ICC's above 0.80 and low levels of SEM variance across all conditions, these were 200 Hz sampling rate with 200 ms balance duration, and 200 Hz sampling rate with 3 seconds balance duration. Therefore, these combinations were deemed suitable for use in Study Two and Study Four of the thesis.

Study Two researched postural sway (anterior, posterior, anteroposterior, medial, lateral and mediolateral directions) following a single leg drop jump landing over i) 3 seconds, and ii) 200 ms, in FAI subjects UA and SA compared to a healthy control group's DA and NDA. The results of Study Two indicated that when analysing the 3 second data there were no significant differences in postural sway for any of the sway directions

between the UA, SA, DA and NDA. This was in contrast to the majority of the literature (Evans et al., 2004; Hiller et al., 2007; McGuine et al., 2000; Mitchell et al., 2008b; Tropp et al., 1984), however, there were some studies that failed to find differences in postural sway between FAI and healthy subjects (Gribble et al., 2006). When analysing the 200ms data there was a significant increase in postural sway in the lateral and mediolateral directions in both the UA and SA of the FAI group when compared to the DA and NDA of the control group. No study has previously analysed this time period, therefore comparisons with the literature were difficult. However, as previously mentioned the majority of literature has identified increased levels of postural sway in FAI subjects compared to healthy controls (Evans et al., 2004; Hiller et al., 2007; McGuine et al., 2000; Mitchell et al., 2008b; Tropp et al., 1984). The results indicate that the FAI subject's postural control may be decreased, but only on a subconscious level as seen by an increase in lateral and mediolateral sway under the 200 ms analysis. It may be possible that after this initial 200 ms the FAI subject is able to regain control of their postural stability on a conscious level, and this is supported by there being no difference in postural sway when looking at the 3 second data. As postural sway was increased on both the UA and SA in the FAI subjects, this may also indicate that FAI affects the postural control system at a level that is high enough to influence stability on either extremity.

10.2.1 Clinical Implications

The main clinical implications that have arisen from the finding of Study Two are that rehabilitation exercises prescribed by sports injury professionals to subjects with unilateral FAI should ensure that the exercises focus on both the UA and SA, as deficits

in postural sway were present in both limbs of the FAI subjects. In addition, if clinicians have access to the use of force platforms they should consider analysing a subconscious time period, as well as the more common conscious time scales, as the present study only found deficits in FAI subjects under the 200 ms analysis.

There is also relevance of the reliability findings of Pilot Study Three, which lie predominantly in the research field. It may be argued that the increase in reliability was marginal between some of the analysis combinations used. However, in the field of research where reliable protocols are a necessity, the two combinations that provided ICC's above 0.80 and low levels of SEM variance across all conditions, were 200 Hz sampling rate with 200 ms balance duration, and 200 Hz sampling rate with 3 seconds balance duration. These combinations improved the reliability of the protocol, and should be considered over the other combinations in future research that utilise the same methods.

10.2.2 Contributions to the Literature

The findings of Study Two have contributed greatly to the literature as this is the first study to investigate postural sway during the subconscious time period (200 ms), as well as the more commonly assessed conscious time frame (3 seconds) in FAI subjects versus healthy controls. Analysis of the subconscious time period was able to identify postural sway deficits that were not present in FAI subjects when analysing a conscious time period. As this study is the first of its kind, these findings warrant further investigation.

10.2.3 Limitations and Recommendations for Future Research

It has previously been reported that postural equilibrium is controlled by the afferent information from the vestibular, visual and somato-sensory systems (Fukuoka et al., 2001; Maurer et al., 2006). The present study did not control for visual or vestibular cues. Further research should look at the effect of blindfolding a subject, minimising vestibular signs, and the effect of a combination of both. It would be interesting to see if the subconscious postural sway deficits were increased when visual and vestibular cues were removed, but it would also be intriguing to see if the conscious postural sway results showed any significant differences between FAI subjects and healthy controls.

It has previously been stated that any deficits are exacerbated under the influence of fatigue (Gribble, Hertel et al., 2007). Fatigue has been suggested as a possible contributing factor to the occurrence of lateral ankle sprains (Gutierrez et al., 2007). It has previously been reported that most injuries occur at the end of an activity when the participant is fatigued, particularly in the last quarter (Hawkins et al., 2001). Further research should investigate the effect of lower extremity fatigue on postural sway in subjects with FAI to see if any further deficits are identified. This final aim was investigated during Study Four, and will be evaluated later in this discussion section.

10.3 Study Three: The Effect of Localised and Globalised Fatigue on Muscle Latency in Healthy and Functionally Unstable Subjects Following a Simulated Ankle Sprain

The aim to evaluate muscle latency in FAI subjects versus healthy controls, both before and immediately after localised and globalised fatigue protocols was addressed in Study Three. Muscle latencies were again determined using a tilt platform perturbation (as discussed in section 10.1, paragraph 1). The localised fatigue protocols were performed on an isokinetic dynamometer and the globalised football-specific fatigue protocol was performed on a treadmill. Before conducting Study Three an extensive review of the literature indicated that localised isokinetic fatigue protocols had previously been found to increase (delay) muscle latency (Cools et al., 2002). However, it has often been argued that these isokinetic protocols are not 'sports specific' and new methods of fatigue, such as the football-specific protocol employed by Drust et al. (2000) should be investigated. No studies could be found that investigated the effect of a globilised fatigue protocol on muscle latency. If fatigue has a detrimental effect on muscle latency, this could lead to an increased risk of injury. It has been previously suggested that muscle fatigue can lead to injury (Davis & Bailey, 1997), as reflected in the increased risk of injury in the second half of a football match, especially during the last quarter of the match (Hawkins et al., 2001). While several studies have evaluated muscle latencies in healthy versus FAI subjects (Beckman & Buchanan, 1995; Ebig et al., 1997; Johnson and Johnson, 1993; Konradsen et al., 1998; Konradsen & Ravn, 1991), a better understanding of the musculature's responses to an inversion-plantarflexion stress in a fatigued state may help to clear up discrepancies in the literature. It may also

help to identify if fatigue is a risk factor that may lead to an ankle sprain in healthy subjects, or lead to repeated sprains in FAI subjects.

A common concern with studies using EMG protocols is the reliability of the analysis procedure. This has been previously discussed in section 10.1, paragraph 2. Another concern for Study Three was the reliability of the isokinetic fatigue protocols used. Therefore, the aims of Pilot Studies Four, Five and Six were to determine the reliability of the isokinetic ankle inversion-eversion fatigue protocol to be used in Study Three and Four. Pilot Study Four addressed the issue of identifying the most reliable isokinetic speed to be used for Study Three and Four. The results identified $120° \cdot s^{-1}$ as the most reliable testing speed during isokinetic ankle inversion-eversion. Therefore, this speed was deemed suitable for use in Study Three and Four. Pilot Study Five addressed the issue of identifying the most reliable ankle setup position to be used in Study Three and Four. The results indicated that the 10° plantarflexion position was the most reliable setup, and also enabled the subject to produce significantly higher peak torque values. Therefore, this setup position was deemed appropriate for Study Three and Four. Pilot Study Six addressed the effect of load range on peak torque values during isokinetic ankle inversion-eversion. The results found that the peak torque values produced by the isokinetic dynamometer should be adjusted to account for load range. Therefore, it was very important that this method was adopted during Study Three and Four.

The aims of Pilot Studies Seven, Eight and Nine were to determine the reliability of the isokinetic hip abduction-adduction fatigue protocol to be used in Study Three and Four. Pilot Study Seven addressed the issue of identifying the most reliable isokinetic speed to be used for Study Three and Four. The results identified $120° \cdot s^{-1}$ as the most

reliable testing speed during the isokinetic hip abduction-adduction fatigue protocol. Therefore, this speed was deemed suitable for use in Study Three and Four. Pilot Study Eight addressed the issue of identifying the most reliable hip setup position to be used in Study Three and Four. The results indicated that the addition of three extra stabilisation straps increased the reliability of peak torque measures, and also enabled the subject to produce significantly higher peak torque values. Therefore, this setup position was deemed appropriate for Study Three and Four. Pilot Study Nine addressed the effect of load range on peak torque values during isokinetic hip abduction-adduction. The results found that the peak torque values given by the isokinetic dynamometer had to be adjusted to account for load range. Therefore, it was very important that this method was adopted during Study Three and Four.

The football-specific globalised fatigue protocol that was used in Study Three and Four was identical to the protocol previously used by Rahnama et al. (2006) and Rahnama et al. (2003). The procedure for this protocol on the treadmill was determined by Drust (1997) to be reliable and repeatable, with a reported coefficient of variation of 4.8% and 95% ratio limits of agreement of 9.4%. Therefore, no pilot work was undertaken for this football-specific fatigue protocol as an identical procedure was being undertaken.

Study Three researched muscle latency, in response to a tilt perturbation, in unilateral FAI subject's UA and SA, compared to a healthy control group's DA and NDA, both before and immediately after a) ankle inversion-eversion isokinetic fatigue, b) hip abduction-adduction isokinetic fatigue, c) treadmill exercise simulating football match play, and d) a control. The results of Study Three indicated that the fatigue conditions when compared to the pre-test and control conditions showed no significant difference

in muscle latency for all muscles and ankles tested. It has previously been suggested that muscle fatigue may predispose an individual to injury. However, the results from the present study suggest that fatigue does not lead to increased (delayed) muscle latencies, and therefore, factors other than fatigue must be present that lead to this increase in injury rate.

10.3.1 Clinical Implications

The main clinical implications that have arisen from the findings of Study Three are that fatigue does not lead to increased (delayed) muscle latencies. Therefore, in terms of muscle latency individuals that participate in sports, as well as sports clinicians and coaches, should not be concerned about the theorised relationship between the onset of fatigue and an increased injury risk as a result of biomechanical deficits. Rather it seems that the increased incidence of injury late in matches may be due to factors such as poor decision making with increasing urgency or tension.

There is also huge clinical relevance in regards to the findings of Pilot Studies Four to Nine. Pilot Studies Four and Seven found that the most reliable testing speed for both the ankle and hip isokinetic protocols was $120° \cdot s^{-1}$. The implication of this finding is that clinicians can now perform an isokinetic fatigue protocol on the ankle invertors and evertors and the hip abductors and adductors with the reassurance that the procedure is reliable in both healthy individuals, and individuals with a history of functional ankle instability at $120° \cdot s^{-1}$. Pilot Studies Five and Eight investigated the reliability of the ankle and hip set up positions, respectively. The results from Pilot Study Five showed that future clinicians using the same protocol should adopt the 10° plantarflexion

position, as this produced the most reliable peak torque and total work results. Pilot Study Eight found that three additional straps during the hip abduction-adduction protocol also increase the reliability of peak torque and total work measures. Therefore, clinicians and researchers using the same hip protocol should adopt this new setup position in future. Pilot Studies Six and Nine found that the peak torque values produced by the isokinetic dynamometer during the ankle and hip isokinetic protocol should be manually adjusted to account for load range. These results imply that if this procedure is not performed by the clinician the peak torque results can often contain errors, with increased inaccuracy at higher velocities.

10.3.2 Contributions to the Literature

The findings of Study Three have contributed to the literature in a number of ways. This was the first study to investigate the effects of both a localised and globalised fatigue protocol on muscle latencies in FAI subjects compared to healthy controls. Even though the results found no significant differences following any of the fatigue protocols, the findings are still relevant and show that there is no relationship between the onset of fatigue and an increased risk of injury. Therefore, other factors must be present that lead to this increase in injury rate following fatigue.

Pilot Studies Seven and Nine have significantly contributed to the literature as they have both been published in the International Journal of Sports Medicine. The first paper is entitled: The effect of isokinetic testing speed on the reliability of muscle fatigue indicators during a hip fatigue protocol. The second paper is entitled: The effect of velocity on load range during isokinetic hip abduction and adduction exercise.

10.3.3 Limitations and Recommendations for Future Research

There is an argument to whether or not isokinetic fatigue can simulate the “real life” functional fatigue that occurs during sports participation. However, the benefits of isokinetic protocols are that they are standardised and easily repeatable. Study Three also investigated the effect of a football specific protocol on muscle latencies, but due to technical impracticalities and safety reasons backwards movements, sideways movements and actions with the ball were not included. Future studies may wish to investigate muscle latencies directly following a sports game, to see if ‘real life’ sporting situations have a greater effect on muscle latency than found in the present study. However, it must be remembered that the functional activities occurring during a match are almost impossible to control, and therefore, would lead to a loss of standardization.

10.4 Study Four: The Effect of Localised and Globalised Fatigue on Postural Sway in Healthy and Functionally Unstable Subjects Following a Single Leg Drop Jump Landing

The aim to evaluate postural sway in FAI subjects versus healthy controls, both before and immediately after localised and globalised fatigue protocols was addressed in Study Four. Postural sway in the anterior, posterior, anteroposterior, medial, lateral and mediolateral directions were determined using an AMTI force platform (as discussed in section 10.2, paragraph 1). The localised fatigue protocols were performed on an isokinetic dynamometer and the globalised football-specific fatigue protocol was performed on a treadmill. Before conducting Study Four an extensive review of the literature indicated that localised isokinetic fatigue protocols had previously been found

to increase postural sway (Gribble & Hertel, 2004a; Miller & Bird, 1976). However, as previously mentioned, it has been argued that isokinetic protocols are not 'sports specific' and new methods of fatigue, such as the football-specific protocol employed by Drust et al. (2000) should be investigated. No studies could be found that directly investigated the effects of a football-specific fatigue protocol on postural sway. However, other globalised fatigue protocols had been performed, such as the Wingate exercise test, the aerobic yo-yo test, and maximal treadmill exercise. These globalised fatigue protocols found an increase in postural sway following completion (Bove et al., 2007; Fox et al., 2008; Yaggie & Armstrong, 2004). If fatigue has a detrimental effect on postural sway, this could lead to an increased risk of injury. Fatigue and deficits in postural control may be predispositions to musculoskeletal injury, such as FAI. While several studies have evaluated postural sway in healthy versus FAI subjects (Evans et al., 2004; Hiller et al., 2007; Tropp et al., 1984), a better understanding of the effects of fatigue on postural sway may help to identify if fatigue is a risk factor for ankle sprains.

A concern with studies that use force platforms is the reliability of the data produced. This has been previously discussed in section 10.2, paragraph 2. Another concern for Study Four was the reliability of the isokinetic fatigue protocols used. This has also been previously discussed in section 10.3, paragraphs 2-4.

Study Four researched postural sway following a single leg drop jump over i) 3 seconds, and ii) 200 ms, in a FAI group's UA and SA compared to a healthy control group's DA and NDA, both before and immediately after a) ankle inversion-eversion isokinetic fatigue, b) hip abduction-adduction isokinetic fatigue, c) treadmill exercise simulating football match play, and d) a control. The results of Study Four indicated that fatigue of

both the hip abductors and adductors and the ankle invertors and evertors produced a significant increase in lateral and mediolateral sway, in both limbs of the FAI subjects during the 3 second and 200 ms analysis. The deficits were greater following the hip fatigue protocol. Even though both deficits were statistically significant there appears to be a relationship between muscle fatigue and impaired postural control in which fatigue to the more proximal musculature impairs postural control to a greater extent than distal musculature fatigue. These results are in agreement with previous studies that have found greater deficits following hip fatigue in comparison to ankle fatigue (Gribble & Hertel, 2004a; Miller & Bird, 1976). The results of Study Four also identified that a globalised football-specific fatigue protocol had an even greater effect on postural sway than either of the localised hip or ankle isokinetic fatigue protocols. The globalised football-specific fatigue protocol caused significantly increased lateral, medial and mediolateral sway in the FAI subjects UA and SA as well as the healthy subjects DA and NDA during the 200 ms analysis. The football specific protocol also caused deficits in postural sway in the lateral and mediolateral directions in the FAI subjects UA and SA during the 3 second analysis. Possible reasons for sway deficits being greatest following the global fatigue protocol may be due to the protocol involving multiple joints, much larger muscle mass and a much longer fatigue protocol. The results of Study Four highlight that both localised and globalised fatigue protocols may result in an increased risk of injury in the FAI subjects, but also that globalised fatigue may lead to injury in previously healthy subjects, as indicated by the increased levels of postural sway.

10.4.1 Clinical Implications

Clinically, these results show that exercise fatiguing the proximal hip joint has a greater effect on postural sway, than fatigue of the more local ankle muscular. Clinicians and sports coaches should be aware of this factor in case training sessions involve training of the more proximal joints. They may assume that individuals with ankle instability may be unaffected by this, but these results show that hip fatigue has a greater affect on postural sway, and therefore, a higher probability of causing repeated injury. In addition to this, if pre-season screening identifies these more proximal deficits, prehabilitation involving the gluteus medius muscle in individuals with FAI, may prove to be beneficial and reduce the probablility of repeated sprains throughout the season.

The results also found that the football-specific fatigue caused the greatest deficits in postural sway. This highlights that the fatigued individual may be at a greater risk of musculoskeletal injury, especially during prolonged exercise that involves multiple joints, such as a football match. Therefore, during the early stages of rehabilitation it is important that steps are taken to help prevent muscle fatigue. As the rehabilitation progresses, players suffering from FAI need to be gradually advanced through this prolonged multi-joint exercise, to ensure that they are ready for return to play.

There is also a large clinical relevance in regards to the findings of Pilot Studies Four to Nine that investigated the reliability of the isokinetic fatigue protocols to be used in Study Three and Four. This has been previously discussed in section 10.3.1, paragraph 2.

10.4.2 Contributions to the Literature

The findings of Study Four have contributed to the literature in a number of ways. This was the first study to investigate the effects of both a localised and globalised fatigue protocol on postural sway in FAI subjects compared to healthy controls. The study found that hip fatigue had a greater negative effect on postural sway than ankle fatigue, but that a globalised football-specific protocol had an even larger negative effect on postural sway. Globalised fatigue may lead to repeated sprains in FAI sufferers, but may also cause a 'first time' sprain in healthy individuals, as evidenced by the increased postural sway results.

As previously mentioned Pilot Studies Seven and Nine have been published in the International Journal of Sports Medicine. See section 10.3.2, paragraph 2.

10.4.3 Limitations and Recommendations for Future Research

As previously stated in section 10.3.3, paragraph 1, there is an argument to whether or not isokinetic fatigue can simulate the "real life" functional fatigue that occurs during sports participation. However, the benefits of isokinetic protocols are that they are standardised and easily repeatable. Study Four also investigated the effect of a football-specific protocol on muscle latencies, but due to technical impracticalities several football specific movement were not included. Future studies may wish to investigate postural sway directly following a sports game, to see if 'real life' sporting situations have an even greater effect on postural sway than found in the present study.

10.6 Review of Hypotheses

With reference to the original hypotheses highlighted in the introduction of this thesis (Section 1.3), it can be concluded that the first two hypotheses can be formally accepted as subjects with FAI were shown to have increased (delayed) muscle latencies, and increased levels of postural sway, in comparison to the healthy controls. In addition to this the third hypothesis can be officially rejected, as no increased (delayed) muscle latencies were found in the FAI subjects in comparison to the healthy controls, following the fatigue protocols. The final hypothesis can be formally accepted as increased levels of postural sway were found in the FAI subjects in comparison to the healthy controls, following the fatigue protocols. A review of more specific hypotheses can be found within the individual studies.

10.7 Conclusion

The findings of this thesis have provided valuable insight into the deficits associated with FAI. Functionally unstable subject's exhibited increased (delayed) muscle latencies when analysing the tilt limb, and increased levels of postural sway during the 200 ms analysis, in comparison to healthy subjects. It had previously been suggested that muscle fatigue could lead to injury, as reflected in the increased risk of injury in the second half of a football match.This thesis therefore intended to provide insight into the effects of fatigue in both FAI subjects and healthy controls. The induction of both localised and globalised fatigue had no effect on muscle latencies in the FAI or healthy subjects, indicating that other factors must be present that lead to this increased injury rate. With this in mind, it was identified that both the localised and globalised fatigue

protocols created significantly increased levels of postural sway in the FAI subjects, with the globalised fatigue protocol also significantly increasing postural sway in the healthy subjects. The globalised football-specific fatigue protocol caused the greatest deficits in the FAI subjects, but also the healthy controls, indicating that the fatigued individual may be at greater risk of musculoskeletal injury during prolonged exercise that involves multiple joints. The results from this thesis conclude that muscular latency is affected by FAI but not fatigue. However, postural sway is affected on a subconscious level by FAI, and with the addition of fatigue a conscious deficit in postural stability is also created.

References

Ageberg, E., Zatterstrom, R. & Moritz, U. (1998). Stabilometry and one-leg hop test have high test-retest reliability. *Scandinavian Journal of Medicine and Science in Sport, 8*, 198-202.

Akhbari, B., Takamjani, I. E., Salavati, M., & Sanjari, M. A. (2007). A 4-week biodex stability exercise programme improved ankle musculature onset, peak latency and balance measures in functionally unstable ankles. *Journal of Physical Therapy in Sport, 8*, 117-129.

Alderton, A. K. & Moritz, U. (1996). Does calf muscle fatigue affect standing balance? *Scandinavian Journal of Medicine and Science in Sport, 6*, 211-215.

Alexander, R. M. & Bennet-Clarke, H. C. (1977). Storage of elastic strain energy in muscle and other tissues. *Nature, 265*(5590), 114-117.

Amaral De Noronha, M. & Borges Junior, N. G. (2004). Lateral ankle sprain: isokinetic test reliability and comparison between invertors and evertors. *Clinical Biomechanics, 19*, 868-871.

Arnold, B. L. & Perrin, D. H. (1993). The reliability of four different methods of calculating quadriceps peak torque and angle-specific torques at 30°, 60° and 75°. *Journal of Sport Rehabilitation*, 2, 243-250.

Ashton-Miller, J. A., Ottaviani, R. A., Hutchinson, C. & Wojtys, E. M. (1996). What best protects the inverted weight bearing ankle against further inversion? Evertor muscle strength compares favourably with shoe height, athletic tape and three orthoses. *American Journal of Sports Medicine, 24*, 800-809.

Astrand, I. (1960). Aerobic work capacity in men and women with special reference to age. Acta Physiologica Scandinavica, 49, 1-92.

Aydog, E., Aydog, S. T., Cakci, A. & Doral, M. N. (2004). Reliability of isokinetic ankle inversion and eversion strength measurement in neutral foot position, using the

Biodex dynamometer. *Knee Surgery, Sports Traumatology, Arthroscopy, 12*, 478-481.

Bahr, R., Pena, F., Shine, J., Lew, W. D., Lindquist, C., Tyrdal, S., et al. (1997). Mechanics of the anterior drawer and talar tilt tests. *Acta Orthopaedica Scandinavica, 68*, 435-441.

Baier, M. & Hopf, T. (1998). Ankle orthoses effect on single-limb standing balance in athletes with functional ankle instability. *Journal of Physical Medicine and Rehabilitation, 79*, 939-944.

Bangsbo, J. (1994). *Fitness training in football: a scientific approach*. Bagsvaerd: HO & Storm.

Bangsbo, J., Norregaard, L. & Thorso, F. (1991). Activity profile of competitive soccer. *Canadian Journal of Sport Science, 16*, 110-116.

Barker, H. B., Beynnon, B. D. & Renstrom, P. A. (1997). Ankle injury risk factors in sports. *Journal of Sports Medicine, 23*, 69-74.

Barrett, J. & Bilisko, T. (1995). The role of shoes in the prevention of ankle sprains. *Journal of Sports Medicine, 20*, 277-280.

Beckman, S. M. & Buchanan, T. S. (1995). Ankle inversion injury and hypermobility: effect on hip and ankle muscle electromyography onset latency. *Archive of Physical Medicine and Rehabilitation, 76*, 1138-1143.

Belenkii, V., Gurfinkel, V. S. & Paltsey, Y. (1967). Elements of control of voluntary movement. *Biofizika, 12*, 134-141.

Bellew, J. W. & Fenter, P. C. (2006). Control of balance differs after knee or ankle fatigue in older women. *Archive of Physical Medicine in Rehabilitation, 87*, 1486-1489.

Benesch, S., Putz, W., Rosenbaum, D. & Becker, H. (2000). Reliability of peroneal reaction time measurements *Clinical Biomechanics, 15*, 21-28.

Bernier, J. N., Perrin, D. H. & Rijke, A. (1997). Effect of unilateral functional instability of the ankle on postural sway and inversion and eversion strength. *Journal of Athletic Training, 32*, 226-231.

Bigland-Richie, B. & Woods, J. J. (1984). Changes in muscular contractile properties and neural control during human muscular fatigue. *Muscle Nerve, 7,* 691-699.

Bohannon, R. W. & Smith, M. B. (1989). Intrasession reliability of angle specific knee extension torque measurements with gravity corrections. *Journal of Orthopaedic and Sports Physical Medicine, 11*, 155-157.

Bosien, W. R., Staples, S. & Russell, S. W. (1955). Residual disability following acute ankle sprains. *Journal of Bone and Joint Surgery, 37*, 1237-1243.

Bosquet, L., Maquet, D., Forthomme, B., Nowak, N., Lehance, C. & Crosier, J. (2010). Effect of the lengthening of the protocol on the reliability of muscle fatigue indicators. *International Journal of Sports Medicine*, *31*, 82-88.

Bove, M., Faelli, E., Tacchino, A., Lofrano, F., Cogo, C. E. & Ruggeri, P. (2007). Postural control after a strenuous treadmill exercise. *Neuroscience Letters, 418*, 276-281.

Boyle, J. & Negus, V. (1998). Joint position sense in the recurrently sprained ankle. *Australian Journal of Physiotherapy, 44*, 159-163.

Bressel, E., Yonker, J. C., Kras, J. & Heath, E. M. (2007). Comparison of static and dynamic balance in female collegiate soccer, basketball and gymnastic athletes. *Journal of Athletic training, 42*, 42-46.

Bridgman, S. A., Clement, D., Downing, A., Walley, G., Phair, I. & Maffulli, N. (2003). Population based epidemiology of ankle sprains attending accident and

emergency units in the West Midlands of England, and a survey of UK practice for severe ankle sprains. *Emergency Medicine Journal, 20*, 508-510.

Brooks, S. C., Potter, B. T. & Rainey, J. B. (1981). Treatment for partial tears of the lateral ligaments of the ankle: a prospective trial. *British Medical Journal, 282*, 606-607.

Brown, L. E., Sjostrom, T., Comeau, M. J., Whitehurst, M., Greenwood, M. & Findley, B. W. (2005). Kinematics of biophysically asymmetric limbs within rate of velocity development. *Journal of Strength and Conditioning Research, 19*, 298-301.

Brown, L. E. & Whitehurst, M. (2000). Load range. In L. Brown (Ed.), *Isokinetics in Human Performance*. (pp. 97-121). Champaign, IL: Human Kinetics Publishers.

Brown, L. E. & Whitehurst, M. (2003). The effect of short term isokinetic training on force and rate of velocity development. *Journal of Strength and Conditioning Research, 17*, 88-94.

Brown, L. E., Whitehurst, M., Bryant, J. R. & Buchalter, D. N. (1993).Reliability of the Biodex System 2 isokinetic dynamometer concentric mode. *Isokinetics in Exercise Science, 3,* 160-163.

Brown, L. E., Whitehurst, M. & Findley, B. W. (2005). Reliability of rate of velocity development and phase measures on an isokinetic device. *Journal of Strength and Conditioning Research, 19,* 189-192.

Brown, L. E., Whitehurst, M., Findley, B. W., Gilbert, R. & Buchalter, D. N. (1995). Isokinetic load range during shoulder rotation exercise in elite male junior tennis players. *Journal of Strength and Conditioning Research, 9*, 160-164.

Brown, L. E., Whitehurst, M., Gilbert, R. & Buchalter, D. N. (1995). The effect of velocity and gender on load range during knee extension and flexion exercise

on an isokinetic device. *Journal of Orthopaedic and Sports Physical Therapy, 21*, 107-112.

Brunt, D., Andersen, J. C., Huntsman, B., Reinhart, L. B., Thorell, A. C. & Sterling, J. C. (1992). Postural responses to lateral perturbation in healthy subjects and ankle sprain patients. *Medicine and Science in Sports and Exercise, 24*, 171-176.

Bullock-Saxton, J. E. (1994). Local sensation changes and altered hip muscle function following severe ankle sprain. *Physical Therapy, 74*, 17-28.

Bullock-Saxton, J. E., Janda, V. & Bullock, M. I. (1994). The influence of ankle sprain injury on muscle activation during hip extension. *International Journal of Sports Medicine, 15*, 330-334.

Bullock-Saxton, J. E., Janda, V. & Bullock, M. I. (1993). Reflex activation of the gluteal muscles in walking with balance shoes: an approach to the restoration of function for chronic lower back pain patients. *Spine, 18*, 704-708.

Burdett, R. G. & Van Swearingen, J. (1987). Reliability of isokinetic muscle endurance tests. *Journal of Orthopaedic and Sports Physical Therapy, 8*, 484-488.

Burnett, C. N., Betts, E. F. & King, W. M. (1990). Reliability of isokinetic measurements of hip muscle torque in young boys. *Physical Therapy, 70*, 244-249.

Callaghan, M. J. (1997). The role of ankle taping and bracing in the athlete. *British Journal of Sports Medicine, 31*, 102-108.

Carcia, C. R., Martin, R. L. & Drouin, J. M. (2008). Validity of the foot and ankle ability measure in athletes with chronic ankle instability. *Journal of Athletic Training, 43*, 179-183.

Caulfield, B. M. (2000). Functional instability of the ankle. Features and underlying causes. *Physiotherapy, 86*, 401-411.

Caulfield, B. M. & Garrett, M. (2002). Functional instability of the ankle: differences in patterns of ankle and knee movement prior to and post landing in a single leg jump. *International Journal of Sports Medicine, 23*, 64-68.

Cavanagh, P. R. & Komi, P. (1979). Electromechanical delay in human skeletal muscle under concentric and eccentric conditions. *European Journal of Applied Physiology and Occupational Physiology, 42*(3), 159-163.

Cawthorn, M., Cummings, G., Walker, J. R. & Donatelli, R. (1991). Isokinetic measurement of foot invertor and evertor force in three positions of plantarflexion and dorsiflexion. *Journal of Orthopaedic and Sports Physical Therapy, 14*, 75-81.

Chanaud, C. M. & Macpherson, J. M. (1991). Functionally complex muscles of the cat hindlimb, III: differential activation with biceps femoris during postural perturbation. *Experimental Brain Research, 85*, 271-280.

Cingel, R. E. H., Kleinrensink, G., Uitterlinden, E. J., Rooijens, P. P. G., Mulder, P. G. H., Aufdemkampe, G., et al. (2006). Repeated ankle sprains and delayed neuromuscular response: acceleration time parameters. *Journal of Orthopaedic and Sports Physical Therapy, 36*, 72-79.

Claiborne, T. L., Timmons, M. K. & Pincivero, D. M. (2009). Test-retest reliability of cardinal plane isokinetic hip torque and EMG. *Journal of Electromyography and Kinesiology, 19*, 345-352.

Cools, A. M., Witvrouw, E. E., Declercq, G. A., Danneels, L. A. & Cambier, D. C. (2002). Scapular muscle recruitment patterns: trapezius muscle latency with

and without impingement symptoms. *The American Journal of Sports Medicine, 31*, 542-549.

Corbeil, P., Blouin, J., Begin, F., Nougier, V. & Teasdale, N. (2003). Perturbation of the postural control system induced by muscular fatigue. *Gait and Posture, 18,* 92-100.

Crenna, P. & Frigo, C. (1987). Excitability of the soleus H-reflex arc during walking and stepping in man. *Experimental Brain Research, 66*, 49-60.

Croisier, J. L., Ganteaume, S., Binet, J., Genty, M. & Ferret, J. (2008). Strength imbalances and prevention of hamstring injury in professional soccer players. *American Journal of Sports Medicine, 36,* 1469-1475.

Currier, D. (1990). *Elements of Research in Physical Therapy.* Baltimore: Williams and Wilkins.

Davies, G. J. (1992). *A Compendium of Isokinetics in Clinical Usage*. Onalaska, WI: S & S Publishers.

Davis, J. M. & Bailey, S. P. (1997). Possible mechanisms of central nervous system fatigue during exercise. *Journal of Medicine and Science in Sport and Exercise, 29*, 45-57.

De Carlo, M. S. & Talbot, R. W. (1986). Evaluation of ankle joint proprioception following injection of the anterior talofibular ligament. *Journal of Orthopaedics and Sports Physical Therapy, 8*, 70-76.

Delahunt, E. (2007a). Neuromuscular contributions to functional instability of the ankle joint. *Journal of Bodywork and Movement Therapies, 11*, 203-213.

Delahunt, E. (2007b). Peroneal reflex contribution to the development of functional instability of the ankle joint. *Journal of Physical Therapy in Sport, 8*, 98-104.

Delahunt, E., Monaghan, K. & Caulfield, B. (2006a). Altered neuromuscular control and ankle joint kinematics during walking in subjects with functional instability of the ankle joint. *American Journal of Sports Medicine, 34*, 1970-1976.

Delahunt, E., Monaghan, K. & Caulfield, B. (2006b). Changes in lower limb kinematics, kinetics and muscle activity in subjects with functional instability of the ankle joint during a single leg drop jump. *Journal of Orthopaedic Research, 24*, 1991-2000.

DeLoes, M. (1990). Medical treatment and costs of sports related injuries in a total population. *International Journal of Sports Medicine, 11*, 66-72.

DeLoes, M. & Goldie, I. (1988). Incidence rate of injuries during sport activity and physical exercise in a rural Swedish municipality: incidence rates in 17 sports. *International Journal of Sports Medicine, 9*, 461-467.

DeLuca, C. (1997). The use of electromyography in biomechanics. *Journal of Applied Biomechanics, 13*, 135-163.

DeLuca, C. & Merletti, R. (1988). Surface myoelectric signal cross-talk among muscles of the leg. *Electroenceph Clinical Neurophysiology, 69*, 568-578.

Demeritt, K. M., Shultz, S. J., Docherty, C. L., Gansneder, B. M. & Perrin, D. H. (2002). Chronic ankle instability does not affect lower extremity functional performance. *Journal of Athletic Training, 37*, 507-511.

DiFabio, R. P. (1987). Reliability of computerized surface electromyography for determining the onset of muscle activity. *Physical Therapy, 67*(1), 43-48.

Docherty, C. L., Arnold, B. L. & Hurwitz, S. (2006a). Contralateral force sense deficits are related to the presence of functional ankle instability. *Journal of Orthopaedic Research, 24*, 1412-1419.

Docherty, C. L., Gansneder, B. M., Arnold, B. L. & Hurwitz, S. R. (2006b). Development and reliability of the ankle instability instrument. *Journal of Athletic Training, 41*, 154-158.

Docherty, C. L. & Rybak-Webb, K. (2009). Reliability of the anterior drawer and talar tilt tests using the LigMaster joint arthrometer. *Journal of Sport Rehabilitation, 18*, 389-397.

Docherty, C. L., Valovich-McLeod, T. C. & Shultz, S. J. (2006c). Postural control deficits in participants with functional ankle stability as measured by the balance error scoring system. *Clinical Journal of Sports Medicine, 16*, 203-208.

Drouin, J. M., Valovich-McLeod, T.C., Shultz, S. J., Gansneder, B. M. & Perrin, D. H. (2004). Reliability and validity of the Biodex system 3 pro isokinetic dynamometer velocity, torque and position measurements. *European Journal of Applied Physiology, 91*, 22-29.

Drust, B. (1997). *Metabolic responses to soccer-specific intermittent exercise.* Unpublished PhD thesis, Liverpool John Moores University, Liverpool.

Drust, B., Cable, N. T. & Reilly, T. (2000). Investigation of the effect of pre-cooling on the physiological responses to soccer-specific intermittent exercise. *European Journal of Applied Physiology, 81*, 11-17.

Ebig, M., Lephart, S. M., Burdett, R. G., Miller, M. C. & Pincivero, D. M. (1997). The effect of sudden inversion stress on EMG activity of the peroneal and tibialis anterior muscles in the chronically unstable ankle. *Journal of Orthopaedic and Sports Physical Therapy, 26*, 73-77.

Edwards, R. H. T., Hill, D. K., Jones, D. A. & Merton, P. A. (1977). Fatigue of long duration in human skeletal muscle after exercise. *Journal of Physiology, 272*, 769-778.

Eechaute, C., Vaes, P., Duquet, W. & Gheluwe, B. V. (2007). Test-retest reliability of sudden ankle inversion measurements in subjects with healthy ankle joints. *Journal of Athletic Training, 42*, 60-65.

Ekdahl, C., Jarnlo, G. B. & Andersson, S. I. (1989). Standing balance in healthy subjects: evaluation of a quantitative test battery on a force platform. *Scandinavian Journal of Rehabilitation Medicine, 21*, 187-195.

Ekstrand, J. & Gillquist, J. (1983). Soccer injuries and their mechanisms: a prospective study. *Journal of Medical Science and Sports and Exercise, 15*, 367-370.

Ekstrand, J. & Hilding, J. (1999). The incidence and differential diagnosis of acute groin injuries in male soccer players. *Scandinavian Journal of Medicine and Science in Sports, 9*, 98-103.

Emery, C. A., Maitland, M. E. & Meeuwisse, W. H. (1999). Test-retest reliability of isokinetic hip adductor and flexor muscle strength. *Clinical Journal of Sport Medicine, 9,* 79-85.

Evans, G. A, Hardcastle, P. & Frenyo, A. D. (1984). Acute rupture of the lateral ligament of the ankle: to suture or not to suture*? Journal of Bone and Joint Surgery, 66*, 209-212.

Evans, T., Hertel, J. & Sebastianelli, W. (2004). Bilateral deficits in postural control following lateral ankle sprain. *International Journal of the Foot and Ankle, 25*, 833-839.

Faul, F., Erdfelder, E., Lang, A. G. & Buchner, A. (2007). G*Power 3: A flexible statistical power analysis program for the social, behavioural and biomedical sciences. *Behaviour Research Methods, 39*, 175-191.

Feiring, D. C., Ellenbecker, T. S. & Dershield, G. L. (1990). Test-retest reliability of the Biodex isokinetic dynamometer. *Journal of Orthopaedics in Sport Physical Therapy, 11*, 298-300.

Ferber, R., McClay-Davis, I. & Williams, D. S. (2003). Gender differences in lower extremity mechanics during running*. Clinical Biomechanics, 18*, 350-357.

Fernandes, N., Allison, G. T. & Hopper, D. (2000). Peroneal latency in normal and injured ankles at varying angles of perturbation. *Clinical Orthopaedics and Related Research, 375,* 193-201.

Findley, B. W., Brown, L. E., Whitehurst, M., Keating, T., Murray, D. P. & Gardner, L. M. (2006). The influence of body position of load range during isokinetic knee extension/flexion. *Journal of Sports Science Medicine, 5*, 400-406.

Fitch, L. K. & McComas, A. J. (1985). Depression of muscle excitability in humans after fatiguing stimulation. *Journal of Physiology, 496*, 851-855.

Forestier, N. & Toschi, P. (2005). The effects of an ankle destabilization device on muscular activity while walking. *International Journal of Sports Medicine, 26*, 464-470.

Fox, J., Docherty, C. L., Schrader, J. & Applegate, T. (2008). Eccentric plantar-flexor torque deficits in participants with functional ankle instability. *Journal of Athletic Training, 43*, 51-54.

Fredericson, M., Cookingham, C. L., Chaudhari, A. M., Dowdell, B. C., Oestreicher, N. & Sahrmann, S. A. (2000). Hip abductor weakness in distance runners with iliotibial band syndrome. *Clinical Journal of Sports Medicine, 10,* 169-175.

Freeman, M. A. (1965a). Treatment of ruptures of the lateral ligament of the ankle. *Journal of Bone and Joint Surgery, 47*, 661-668.

Freeman, M. A. (1965b). Instability of the foot after injuries to the lateral ligament of the ankle. *Journal of Bone and Joint Surgery, 47*, 669-677.

Freeman, M. A., Dean, M. R. & Hanham, I. W. (1965). The etiology and prevention of functional ankle instability of the foot. *Journal of Bone and Joint Surgery, 47*, 678-685.

Friden, T., Zatterstrom, R., Lindstrand, A. & Moritz, U. (1989). A stabliometric technique for evaluation of lower limb instabilities. *American Journal of Sports Medicine, 17*, 118-122.

Friel, K., McLean, N., Myers, C., & Caceres, M. (2006). Ipsilateral hip abductor weakness after inversion ankle sprain. *Journal of Athletic Training, 41*, 74-78.

Fritschy, D., de Reynier, J. C. & Blanc, Y. (1988). Plastic surgery of the ligament for chronic lateral instability of the ankle. *International Orthopaedics, 12*, 239-247.

Fu, A. S. & Hui-Chan, C. W. (2005). Ankle joint proprioception and postural control in basketball players with bilateral ankle sprains. *American Journal of Sports Medicine, 33*, 1174-1182.

Fukuoka, Y., Nagata, T., Ishida, A. & Minamitani, H. (2001). Characteristics of somatosensory feedback in postural control during standing. *Journal of Transactions on Neural Systems and Rehabilitation Engineering, 9*, 145-151.

Gabriel, D. A. & Basford, J. R. (2001). Neural adaptions to fatigue: implications for muscle strength and training. *Medicine and Science in Sport and Exercise, 33*, 1354-1360.

Garn, S. N. & Newton, R. A. (1988). Kinaesthetic awareness in subjects with multiple ankle sprains. *Journal of Physical Therapy in Sport, 68*, 1667-1671.

Garrick, J. G. (1977). The frequency of injury, mechanism of injury, and epidemiology of ankle sprains. *American Journal of Sports Medicine, 5*, 241-242.

Garrick, J. G. (1987). Epidemiology of foot and ankle injuries. *Medicine and Science in Sports and Exercise, 23*, 1-7.

Garrick, J. G. & Requa, R. K. (1989). The epidemiology of foot and ankle injuries in sports. *Clinics in Podiatric and Medical Surgery, 6*, 629-637.

Gauffin, H., Tropp, H. & Odenrick, P. (1988). Effect of ankle disc training on postural control in patients with functional instability of the ankle joint. *International Journal of Sports Medicine, 9*, 141-144.

Gautrey, C. N., Watson, T. & Mitchell, A. (2013a). The effect of isokinetic testing speed on the reliability of muscle fatigue indicators during a hip abduction-adduction fatigue protocol*. International Journal of Sports Medicine, 34*, 646-653.

Gautrey, C. N., Watson, T. & Mitchell, A. (2013b). The effect of velocity on load range during isokinetic hip abduction-adduction exercise. *International Journal of Sports Medicine, 34,* 623-630.

Gear, W. S. (2011). Effect of different levels of localized muscle fatigue o knee position sense. *Journal of Sports Medicine, 10*, 725-730.

Gerdle, B. & Elert, J. (1994). The temporal occurrence of the mean power frequency shift of the electromyogram during maximal prolonged dynamic and static working cycles. *International Journal of Sports Medicine, 15*, S32-S37.

Glaister, M., Stone, M. H., Stewart, A. M., Hughes, M. & Moir, G. L. (2004). The reliability and validity of fatigue measures during short duration maximal intensity intermittent exercise. *Journal of Strength and Conditioning Research, 18*, 459-462.

Glencross, D. & Thornton, E. (1981). Position sense following joint injury. *Journal of Sports Medicine and Physical Fitness, 21*, 23-27.

Gleeson, N. P. & Mercer, T. H. (1992). Reproducibility of isokinetic leg strength and endurance characteristics of adult men and woman. *European Journal of Applied Physiology, 65*, 221-228.

Gleeson, N. P. & Mercer, T. H. (1996). The utility of isokinetic dynamometry in the assessment of human muscle function. *Journal of Sports Medicine, 21*, 18-34.

Glick, J. M., Gordon, R. B. & Nishimoto, D. (1976). The prevention and treatment of ankle injuries*. American Journal of Sports Medicine, 4*, 136-141.

Goldie, P. A., Evans, O. M. & Bach, T. M. (1994). Postural control following inversion injuries of the ankle. *Archive of Physical and Medical Rehabilitation, 75*, 969-975.

Goldie, P. A., Evans, O. M. & Bach, T. M. (1992). Steadiness in one-legged stance: development of a reliable force platform testing procedure. *Archives of Physical Medicine and Rehabilitation, 73,* 348-354.

Goldie, P. A., Bach, T. M., Evans, O. M. (1989). Force platform measures for evaluating postural control: reliability and validity. *Archives of Physical Medicine and Rehabilitation, 70*, 510-517.

Granata, K. P., Abel, M. F. & Damiano, D. L. (2000). Joint angular velocity in spastic gait and the influence of muscle tendon lengthening. *Journal of Bone and Joint Surgery, 82,* 174-186.

Green, H. J. (1995). Mechanisms of muscle fatigue in intense exercise. *Journal of Sport Sciences, 15*, 247-256.

Greenisen, M. C., Vroomen, L. J. & Vroomen, L. C. (1979). Computerised interpretation of initiation and relaxation of EMG recorded dynamic muscle contraction. *Electrophysiological Kinesiology, 1*, 172-173.

Greig, M. P. & Siegler, J. C. (2009). Soccer-specific fatigue and eccentric hamstring muscle strength. *Journal of Athletic Training, 44*, 180-184.

Greig, M. P., McNaughton, L. R. & Lovell, R. J. (2006). Physiological and mechanical response to soccer-specific intermittent activity and steady-state activity. *Research in Sports Medicine, 14*, 29-52.

Grey, E. R. & Basmajian, J. V. (1968). Electromyography and cinematography of leg and foot ("normal" and flat) during walking. *The Anatomical Record, 161*, 1-16.

Gribble, P. A. & Hertel, J. (2004a). Effect of hip and ankle muscle fatigue on unipedal postural control. *Journal of Electromyography and Kinesiology, 14*, 641-646.

Gribble, P. A. & Hertel, J. (2004b). Changes in postural control during a 48-hour sleep deprivation period. *Perception of Motor Skills, 99*, 1035-1045.

Gribble, P. A., Hertel, J. & Denegar, C. R. (2007). Chronic and instability and fatigue create proximal joint alterations during performance of the star excursion balance test. *International Journal of Sports Medicine, 28*, 236-242.

Gribble, P. A., Hertel, J., Denegar, C. R. & Buckley, W. E. (2004). The effects of fatigue and chronic ankle instability on dynamic postural control. *Journal of Athletic Training, 39*, 321-329.

Gribble, P. A., Radel, S. & Armstrong, C. W. (2006). The effects of ankle bracing on the activation of the peroneal muscles during a lateral shuffling movement. *Journal of Physical Therapy in Sport, 7*, 14-21.

Gribble, P. A., Tucker, S., & White, P. A. (2007). Time-of-day influences on static and dynamic postural control. *Journal of Athletic Training, 42*, 35-41.

Gross, M. T., Huffman, G. M., Phillips, C. N. & Wray, J. A. (1991). Intramachine and intermachine reliability of the Biodex and the Cybex II for knee flexion and

extension peak torque and angular work. *Journal of Orthopaedic and Sports Physical Therapy, 13*, 329-335.

Gruneberg, C., Nieuwenhuijzen, P. H. & Duysens, J. (2003). Reflex responses in the lower leg following landing impact on an inverting and non-inverting platform. *Journal of Physiology, 550*, 985-993.

Gutierrez, G. M., Jackson, N. D., Dorr, K. A., Margiotta, S. E. & Kaminski, T. W. (2007). Effect of fatigue on neuromuscular function at the ankle. *Journal of Sports Rehabilitation, 16*, 295-306.

Hale, S. A. & Hertel, J. (2005). Reliability and sensitivity of the Foot and Ankle Disability Index in subjects with chronic ankle instability. *Journal of Athletic Training, 40*, 35-40.

Hale, S. A., Hertel, J. & Olmsted-Kramer, L. C. (2007). The effect of a 4-week comprehensive rehabilitation program on postural control and lower extremity function in individuals with chronic ankle instability. *Journal of Orthopaedic and Sports Physical Therapy, 37*, 303-311.

Hamdoun-Kahlaoui, S., Lebib, S., Miri, I., Ghorbel, S., Koubaa, S., Rahali-Khachlouf, H., Ben Salah, F. Z. & Dziri, C. (2010). Isokinetic evaluation and rehabilitation of the knee in patients with patellofemoral pain syndrome. *Journal de Readaptation Medicale, 30,* 3-11.

Hammami, N., Coroian, F. O., Julia, M., Amri, M., Mottet, D., Herisson, C. & Laffont, I. (2012). Isokinetic muscular strengthening after acquired cerebral damage: a literature review. *Annals of Physical and Rehabilitation Medicine, 55*, 279-291.

Hanten, W. P. & Ramberg, C. L. (1988). Effect of stabilisation on maximal isokinetic torque of the quadriceps femoris muscle group during concentric and eccentric contractions. *Physical Therapy, 68*, 219-222.

Happee, R. (1992). Goal directed arm movements: analysis of EMG records in shoulder and elbow muscles. *Journal of Electromyography and Kinesiology, 2*, 165-178.

Harkins, K. M., Mattacola, C. G., Uhl, T. L., Malone, T. R. & McCrory, J. L. (2005). Effects of two ankle fatigue models on the duration of postural stability dysfunction. *Journal of Athletic Training, 40*, 191-196.

Hart, D. L., Stobbe, T. J., Till, W. W. & Plummer, R. W. (1984). Effect of trunk stabilisation on quadriceps femoris muscle torque. *Physical Therapy, 64*, 1375-1380.

Hartsell, H. D. & Spaulding, S. J. (1999). Eccentric/concentric ratios at selected velocities for the invertor and evertor muscles of the chronically unstable ankle. *British Journal of Sports Medicine, 33*, 255-258.

Hawkins, R. D., Hulse, M. A., Wilkinson, C., Hodson, A. & Gibson, M. (2001). The association football medical research programme: an audit of injuries in professional football. *British Journal of Sports Medicine, 35*, 43-47.

Hertel, J. (2000). Functional instability following lateral ankle sprain. *Journal of Sports Medicine, 29*, 361-371.

Hertel, J. (2002). Functional anatomy, pathomechanics, and pathophysiology of lateral ankle instability. *Journal of Athletic Training, 37*, 364-375.

Hertel, J., Buckley, W. E. & Denegar, C. R. (2001). Serial testing of postural control after acute lateral ankle sprain. *Journal of Athletic Training, 36*, 363-368.

Heyworth, J. (2003). Ottowa ankle rules for the injured ankle. *British Journal of Sports Medicine, 37*, 194.

Hiller, C. E., Refshauge, K.M., Bundy, A. C., Herbert, R. D. & Kilbreath, S. L. (2006). The Cumberland Ankle instability Tool: report of validity and reliability. *Archives of Physical Medicine and Rehabilitation, 87*, 1235-1241.

Hiller, C. E., Refshauge, K. M., Herbert, R. D. & Kilbreath, S. L. (2007). Balance recovery from a perturbation are impaired in people with functional ankle instability. *Clinical Journal of Sports Medicine, 17*, 269-275.

Hislop, H. J. & Perrine, J. J. (1967). The isokinetic concept of exercise. *Physical Therapy, 47*, 114-117.

Hodges, P. W. & Bui, B. H. (1996). A comparison of computer based methods of the determination of onset of muscle contraction using electromyography. *Electroenceph Clinical Neurophysiology, 101*, 511-519.

Hogervorst, T. & Brand, R. A. (1998). Mechanoreceptors in joint function. *Journal of Orthopaedics and Sports Physical Therapy, 80*, 1365-1378.

Holmich P. (1998). Adductor related groin pain in athletes. *Sports Medicine Arthroscopy Review, 5,* 285-291.

Hopkins, W. G. (2000). Measures of reliability in sports medicine and science. *Journal of Sports Medicine, 30*, 375-381.

Houweling, T. A., Head, A. & Hamzeh, M. A. (2009). Validity of isokinetic testing for previous hamstring injury detection in soccer players*. Isokinetics in Exercise Science, 17,* 213-220.

Hubbard, T. J. & Kaminski, T. W. (2002). Kinaesthesia is not affected by functional ankle instability status. *Journal of Athletic Training, 37*, 481-486.

Hume, P. A. & Gerrard, D. F. (1998). Effectiveness of external ankle support. *Journal of Sports Medicine, 25*, 285-312.

Hunter, S. K. & Enoka, R. M. (2003). Changes in muscle activation can prolong the endurance time of a submaximal isometric contraction in humans. *Journal of Applied Physiology, 94*, 108-118.

Huston, J. L., Sandrey, M. A., Lively, M. W. & Kotsko, K. (2005). The effects of calf muscle fatigue on sagittal-plane joint-position sense in the ankle. *Journal of Sports Rehabilitation, 14*, 168-184.

Inglis, J. T., Horak, F. B., Shupert, C. L. & Jones-Rycewicz, C. (1994). The importance of somatosensory information in triggering and scaling automatic postural responses in humans. *Experimental Brain Research, 101*, 159-164.

Inman, V. T. (1947). Functional aspects of the abductor muscles of the hip. *Journal of Bone and Joint Surgery, 29,* 607-619.

Isakov, E., Mizrahi, J., Solzi, P., Susak, Z. & Lotem, M. (1986). Response of the peroneal muscles to sudden inversion of the ankle during standing. *International Journal of Sport Biomechanics, 2*, 100-109.

Ives, J. C. & Wigglesworth, J. K. (2003). Sampling rate effects on surface EMG timing and amplitude measures. *Clinical Biomechanics, 18*, 543-552.

Jackson, D. W., Ashley, R. L. & Powell, J. W. (1974). Ankle sprains in young athletes: relation of severity and disability. *Clinical Orthopaedics, 101*, 201-215.

Jackson, N. D., Gutierrez, G. M. Kaminski, T. (2009). The effect of fatigue and habituation on the stretch reflex of the ankle musculature. *Journal of Electromyography and Kinesiology, 19*, 75-84.

Jacobs, C. A., Uhl, T. L., Mattacola, C. G., Shapiro, R. & Rayens, W. S. (2007). Hip abductor function and lower extremity landing kinematics: sex differences. *Journal of Athletic Training, 42*, 76-83.

Jacobs, C., Uhl, T. L., Seeley, M., Sterling, W. & Goodrich, L. (2005). Strength and fatigability of the dominant and nondominant hip abductors. *Journal of Athletic Training, 40*, 203-206.

Janssen, J. C. & Le-Ngoc, L. (2009). Intratester reliability and validity of concentric measurements using a new hand-held dynamometer. *Archives of Physical Medicine and Rehabilitation, 90*, 1541-1547.

Jerosch, J. & Bischof, M. (1996). Proprioceptive capabilities of the ankle in stable and unstable joints. *Sports and Exercise Injuries, 2*, 167-171.

Johansson, R. & Magnusson, M. (1991). Human postural dynamics. *Critical Reviews in Biomedical Engineering, 18*, 413-437.

Johansson, H., Sjolander, P. & Sojka, P. (1991). A sensory role for the cruciate ligaments. *Clinical Orthopaedics, 265*, 161-178.

Johnson, M. B. & Johnson, C. L. (1993). Electromyographic responses of peroneal muscles in surgical and non-surgical injured ankles during sudden inversion. *Journal of Orthopaedic and Sports Physical Therapy 18*, 497-501.

Johnson, M. E., Millie, M. L., Martinez, K. M., Crombie, G. & Rogers, M. W. (2004). Age-related changes in hip abductor and adductor joint torques. *Archives of Physical Medicine and Rehabilitation, 85,* 593-597.

Johnson, J. & Siegel, D. (1978). Reliability of an isokinetic movement of the knee extensors. *Research Quarterly, 49*, 88-90.

Johnston, R. B., Howard, M. E., Cawley, P. M. & Losse, G. M. (1998). Effect of lower limb muscular fatigue on motor control performance. *Medicine and Science in Sport and Exercise, 30*, 1703-1707.

Kaminski, T. W., Buckley, B. D., Powers, M. E., Hubbard, T. J. & Ortiz, C. (2003) Effect of strength and proprioception training on eversion to inversion strength

ratios in subjects with unilateral functional ankle instability. *British Journal of Sports Medicine, 37*, 410-415.

Kaminski, T. W. & Dover, G. C. (2001). Reliability of inversion and eversion peak- and average-torque measurements from the Biodex system 3 dynamometer. *Journal of Sports Rehabilitation, 10*, 205-220.

Kaminski, T. W., Perrin, D. G. & Gansneder, B. M. (1999). Eversion strength analysis of uninjured and functionally unstable ankles. *Journal of Athletic Training, 34*, 239-245.

Kaminski, T. W., Perrin, D. H., Mattacola, C. G., Szczerba, J. E. & Bernier, J. N. (1995). The reliability and validity of ankle inversion and eversion torque measurements from the Kin Com II isokinetic dynamometer. *Journal of Sports Rehabilitation, 4,* 210-218.

Kanehisa, H., Yata, H., Ikegawa, S., Fukunaga, T. (1995). A cross-sectional study of the size and strength of the lower leg muscles during growth. *European Journal of Applied Physiology and Occupational Physiology, 72,* 150–156.

Kannus, P. (1994). Isokinetic evaluation of muscular performance: implications for muscle testing and rehabilitation. *International Journal of Sports Medicine, 15,* S11-S18.

Karlsson, J. & Andreasson, G. O. (1992). The effect of external ankle support in chronic lateral ankle joint instability. *American Journal of Sports Medicine, 20*, 257-261.

Karnofel, H., Wilkinson, K. & Lentell, G. (1989). Reliability of isokinetic muscle testing at the ankle. *Journal of Orthopeadic and Sports Physical Therapy, 11*, 150-154.

Karst, G. M. & Willet, G. M. (1995). Onset timing of electromyographic activity in the vastus medialis oblique and vastus lateralis muscles in subjects with and without patellofemoral pain syndrome. *Physical Therapy, 75*, 813-823.

Kellis, E., Katis, A. & Vrabas, I. S. (2006). Effects of an intermittent exercise fatigue protocol on biomechanics of soccer kick performance. *Scandinavian Journal of Medicine and Science in Sports, 16,* 334-344.

Kent-Braun, J. A. (1999). Central and peripheral contributions to muscle fatigue in human during sustained maximal effort. *European Journal of Applied Physiology, 80*, 57-63.

Kernell, D. (1969). *The motorneurone and its muscle fibres*. (3rd ed.). Oxford: Oxford Scholarship.

Kernozek, T., Durall, C., Friske, A. & Mussallem, M. (2008). Ankle bracing, plantar flexion angle, and ankle muscle latencies during inversion stress in healthy participants. *Journal of Athletic Training, 43*, 37-43.

Khin, M. H., Ishii, T., Sakane, M. & Hayashi, K. (1999). Effect of anaesthesia of the sinus tarsi on the peroneal reaction time in patients with functional instability of the ankle. *Foot and Ankle International, 20*, 554-559.

Kinsella, S. & Harrison, D. (1998). A study to examine the balance of subjects with recurrent ankle sprains. *Physiotherapy Ireland, 19*, 9-13.

Kleinrensink, G. J., Stoeckart, R. & Meulstee, J. (1994). Lowered motor conduction velocity of the peroneal nerve after inversion trauma. *Journal of Medical Science and Sports and Exercise, 26*, 877-883.

Konradsen, L. (2002). Sensori-motor control of the uninjured and injured human ankle. *Journal of Electromyography and Kinesiology, 12*, 199-203.

Konradsen, L., Olesen, S. & Hansen, H. (1998). Ankle sensorimotor control and eversion strength after acute ankle inversion injuries. *American Journal of Sports Medicine, 26*, 72-77.

Konradsen, L. & Ravn, J. B. (1990). Ankle instability caused by prolonged peroneal reaction time. *Scandinavian Journal of Orthopaedic Surgery, 61*, 388-390.

Konradsen, L. & Ravn, J. B. (1991). Prolonged peroneal reaction time in ankle instability. *International Journal of Sports Medicine, 12*, 290-292.

Konradsen, L., Ravn, J. B. & Sorensen, A. I. (1993). Proprioception at the ankle: the effect of anaesthetic blockage of ligament receptors. *Journal of Bone and Joint Surgery, 75*, 433-436.

Konradsen, L., Voigt, M. & Hojsgaard, C. (1997). Ankle inversion injuries. The role of the dynamic defense mechanism. *American Journal of Sports Medicine, 25*, 54-58.

Kovaleski, J. E., Heitman, R. H., Trundle, T. L. & Gilley, W. F. (1995). Isokinetic preload versus isokinetic knee extension resistance training. *Medicine and Science in Sports and Exercise, 27*, 895-899.

Kurdak, S. S., Ozgunen, K., Adas, U., Zeren, C., Aslangiray, B., Yazici, Z. & Korkmaz, S. (2005). Analysis of isokinetic knee extension/flexion in male elite adolescent wrestlers. *Journal of Sports Science and Medicine, 4*, 489-498.

Kuwada, G. T. (1995). Current concepts in the diagnosis and treatment of ankle sprains. *Clinics in Podiatric Medicine and Surgery, 12*, 653-665.

Laheru, D., Kerr, J. C. & McGregor, A. H. (2007). Assessing hip abduction and adduction strength: Can greater segmental fixation enhance reproducibility? *Archives of Physical Medicine and Rehabilitation, 88*, 1147-1153.

Larsson, B., Karlsson, S., Eriksson, M. & Gerdle, B. (2003). Test-retest reliability of EMG and peak torque during repetitive maximum concentric knee extensions. *Journal of Electromyography and Kinesiology*, 13, 281-287.

Latash, M. L., Aruin, A. S. & Shapiro, M. B. (1995). The relation between posture and movement: a study of a simple synergy in a two-joint task. *Human Movement Science, 14*, 79-107.

Leanderson, J., Bergqvist, M., Rolf, C., Westblad, P., Wigelius-Roovers, S. & Wredmark, T. (1999). Early influence of an ankle sprain on objective measures of ankle joint function. *Journal of Knee Surgery, Sports Traumatology and Arthroscopy, 7*, 51-58.

Leanderson, J., Eriksson, E. & Nemeth, G. (1993). Ankle injuries in basketball players. *Journal of Knee Surgery, Sports Traumatology and Arthroscopy, 1*, 200-202.

Lee, W. A., Buchanan, T. S. & Rogers, M. W. (1987). Effects of arm acceleration and behavioural conditions on the organisation of postural adjustments during arm flexion. *Experimental Brain Research, 66*, 257-270.

Lee, H., Ho, P., Rastgaar, M. A., Krebs, H. & Hogan, N. (2011). Multivariate static ankle mechanical impedance with relaxed muscles. *Journal of Biomechanics, 44*, 1901-1908.

Lentell, G. L., Katzman, L. L. & Walters, M. R. (1990). The relationship between muscle function and ankle stability. *Journal of Sports Physical Therapy, 11*, 605-611.

Lepers, R., Bigard, A. X., Diard, J. P., Gouteyron, J. F. & Guezennec, C. Y. (1997). Posture control after prolonged exercise. *European Journal of Applied Physiology, 76*, 55-61.

Lepers, R., Maffiuletti, N., Rochette, L., Brugniaux, J. & Millet, F. (2002). Neuromuscular fatigue during a long-duration cycling exercise*. Journal of Applied Physiology, 92*, 1487-1493.

Lephart, S. M., Pincivero, D. M. & Rozzi, S. L. (1998). Proprioception of the ankle and knee. *Journal of Sports Medicine, 25*, 149-155.

Leslie, M., Zachazewski, J. & Browne, P. (1990). Reliability of isokinetic torque values for ankle invertors and evertors. *Journal of Orthopaedics and Sports Physical Therapy, 11,* 612-616.

Liu, S. H. & Jason, W. J. (1994). Lateral ankle sprains and instability problems. *Clinical Sports Medicine, 13*, 793-809.

Lofvenberg, R., Karrholm, J., Sundelin, G. & Ahlgren, O. (1995). Prolonged reaction time in patients with chronic lateral instability of the ankle. *American Journal of Sports Medicine, 23*, 414-417.

Lundin, T. M., Feuerbach, J. W. & Grabiner, M. D. (1993). The effect of plantar flexor and dorsi flexor fatigue on unilateral postural control. *Journal of Applied Biomechanics, 9*, 191-201.

Lynch, S. A., Eklund, U., Gottlieb, D., Renstrom, P. A. & Beynnon, B. D. (1996). Electromyographic latency changes in the ankle musculature during inversion moments. *The American Journal of Sports Medicine, 24*, 362-369.

Mack, R. P. (1982). Ankle injuries in athletes. *Clinics in Sports Medicine, 1*, 71-84.

MacKinnon, C. D. & Winter, D. A. (1993). Control of whole body balance in the frontal plane during human walking. *Journal of Biomechanics, 26*, 633-644.

Maffiuletti, N. A., Bizzini, M., Desbrosses, K., Babault, N. & Munzinger, U. (2007). Reliability of knee extension and flexion measurements using the Con-Trex

isokinetic dynamometer. *Clinical and Physiological Functional Imaging, 27*, 346-353.

Makofsky, H., Panicker, S., Abbruzzese, J., Aridas, C., Camp, M., Drakes, J., Franco, C. & Sileo, R. (2007). Immediate effect of grade IV inferior hip joint mobilization on hip abductor torque: a pilot study. *Journal of Manual Manipulative Therapy, 15,* 103-111.

Marieb, E. N., Wilhelm, P. B. & Mallatt, J. (2012). *Human Anatomy*. (7th ed.). Glenview, IL: Pearson.

Masharawi, Y., Carmeli, E., Masharawi, R. & Trott, P. (2003). The effect of braces on restricting weight-bearing ankle inversion in elite netballers. *Journal of Physical Therapy in Sport, 4*, 24-33.

Maurer, C., Mergner, T. & Peterka, R. J. (2006). Multisensory control of human upright stance. *Journal of Experimental Brain Research, 171*, 231-250.

Mazaheri, M., Negahban, H., Salavati, M., Sanjari, M. A. & Parnianpour, M. (2010). Reliability of recurrence quantification analysis measures of the centre of pressure during standing in individuals with musculoskeletal disorders. *Medical Engineering and Physics, 32*, 808-812.

McGuine, T. A., Greene, J. J., Best, T. & Leverson, G. (2000). Balance as a predictor of ankle injuries in high school basketball players. *Clinical Journal of Sports Medicine, 10*, 239-244.

McHugh, M. P., Tyler, T. F., Tetro, D. T., Mullaney, M. J. & Nicholas, S. J. (2006). Risk factors for noncontact ankle sprains in high school athletes – the role of hip strength and balance ability. *American Journal of Sports Medicine, 34*, 464-470.

McKay, G. D., Goldie, P. A., Payne, W. R. & Oakes, B. W. (2001). Ankle injuries in basketball: injury rate and risk factors. *British Journal of Sports Medicine, 35*, 103-108.

McKinley, P. & Pedotti, A. (1992). Motor strategies in landing from a jump: the role of skill in task execution. *Experimental Brain Research, 90*, 427-440.

McNair, P. J., Depledge, J., Brettkelly, M. & Stanley, S. N. (1996). Verbal encouragement: effects on maximum effort voluntary muscle action. *British Journal of Sports Medicine, 30*, 243-245.

Messina, D. F., Farney, W. C. & DeLee, J. C. (1999). The incidence of injury in Texas high school basketball: a prospective study among male and female athletes. *American Journal of Sports Medicine, 27*, 294-299.

Michaut, A., Pousson, M., Babault, N. & Van Hoecke, J. (2002). Is eccentric exercise-induced torque decrease contraction type dependent? *Medicine and Science in Sports and Exercise, 34*, 1003-1008.

Miller, P. K. & Bird, A. M. (1976). Localized muscle fatigue and dynamic balance. *Perceptual and Motor Skills, 42*, 135-138.

Mitchell, A., Dyson, R., Hale, T. & Abraham, C. (2008a). Biomechanics of ankle instability. Part 1: Reaction time to simulated ankle sprain. *American College of Sports Medicine, 40*(8)*,* 1515-1521.

Mitchell, A., Dyson, R., Hale, T. & Abraham, C. (2008b). Biomechanics of ankle instability. Part 2: Postural sway – reaction time relationship. *American College of Sports Medicine, 40*(8)*,* 1522-1528.

Monaghan, K., Delahunt, E. & Caulfield, B. (2006). Ankle function during gait in patients with chronic ankle instability compared to controls. *Clinical Biomechanics, 21*, 168-174.

Montgomery, L. C., Douglas, L. W. & Deuster, P. A. (1989). Reliability of an isokinetic test of muscle strength and endurance. *Journal of Orthopaedic and Sports Physical Therapy, 10*, 315-322

Morey-Klapsing, G., Arampatzis, A. & Bruggemann, G. P. (2004). Choosing EMG parameters: comparison of different algorithms and EMG intergrals in a joint stability study. *Clinical Biomechanics, 19*, 196-201.

Moseley, J. B. & Chimenti, B. T. (1995). Foot and ankle injuries in the professional athlete. In D Baxter (Ed.), *The Foot and Ankle in Sport*. (pp. 321-328). St. Louis, MO: Mosby-Year Book.

Mulligan, B. R. (1995). *Manual therapy: "NAGS", "SNAGS", and "MWMS."* (3rd ed.). Wellington, New Zealand: Plane View Services Ltd.

Mulloy-Forkin, D., Koczur, C., Battle, R. & Newton, R. A. (1996). Evaluation of kinaesthetic deficits indicative of balance control in gymnasts with unilateral chronic ankle sprain. *Journal of Orthopaedic and Sports Physical Therapy, 23*, 245-250.

Munn, J., Beard, D. J., Refshauge, K. M. & Lee, R. Y. (2003). Eccentric muscle strength in functional ankle instability. *Medicine and Science in Sports and Exercise, 3*, 245-250.

Munro, B. (1997). *Statistical Methods for Health Care Research*. (3rd ed.). New York: JB Lippincott.

Murray, M. P. (1967). Gait as a total pattern of movement. *American Journal of Physical Medicine, 46*, 290-333.

Murray, D. P., Brown, L. E., Zinder, S. M., Noffal, G. J., Bera, S. G. & Garrett, N. M. (2007). Effects of velocity specific training on rate of velocity development,

peak torque and performance. *Journal of Strength and Conditioning Research, 21,* 870-874.

Myers, J. B., Riemann, B. L., Hwang, J. H., Fu, F. H., Lephart, S. M. (2003).Effect of peripheral afferent alteration of the lateral ankle ligaments on dynamic stability. *American Journal of Sports Medicine, 31*, 498-506.

Nadler, S. F., Malanga, G. A., DePrince, M., Stitik, T. P. & Feinberg, J. H. (2000). The relationship between lower extremity injury, low back pain and hip muscle strength in male and female collegiate athletes. *Clinical Journal of Sports Medicine, 10,* 89-97.

Nardone, A., Tarantola, J., Galante, M. & Schieppati, M. (1998). Time course of stabilometric changes after a strenuous treadmill exercise. *Archives of Physical Medicine and Rehabilitation, 79*, 920-924.

Nasher, L. M. & McCollum, G. (1985). The organisation of human postural movements: a formal basis and experimental synthesis. *Behavioural Brain Sciences, 8*, 135-172.

Nasher, L. M., Shupert, C. L., Horak, F. B. & Black, F. O. (1989). Organisation of postural controls: an analysis of sensory and mechanical constraints. *Progress in Brain Research, 80*, 411-418.

Nashner, L. B., Shumway-Cook, A.V. & Marin, O. (1983). Stance posture control in select group of children with cerebral palsy: deficits in sensory organization and muscular coordination. Experimental Brain Research, 49, 393-409.

Nashner, L. M., Woollacott, M. & Tuma, G. (1979). Organization of rapid responses topostural and locomotor-like perturbations of standing man. Experimental Brain Research, 36, 463-476.

Nawoczenski, D. A., Owen, M. G., Ecker, M. L., Altman, B. & Epler, M. (1985). Objective evaluation of peroneal response to sudden inversion stress. *Journal of Orthopaedic and Sports Physical Therapy, 7*(3), 107-109.

Neafsey, E. J., Hull, C. D. & Buchwald, N. A. (1978). Preparation for movement in the cat, II: unit activity in the basal ganglia and thalamus. *Electroenceph Clinical Neurophysiology, 44*, 714-723.

Nelson, J. K. & Johnson, B. L. (1973). Effects of local and general fatigue on static balance. *Peceptual Motor Skills, 37*, 615-618.

Nicholas, S. J. & Tyler, T. F. (2002). Adductor muscle strains in sport. Journal of *Sports Medicine, 32*, 339-344.

Nickols-Richardson, S. M., Miller, L. E., Wooten, D. F., Ramp, W. K. & Herbert, W. G. (2007). Concentric and eccentric isokinetic resistance training similarly increases muscular strength, fat-free soft tissue mass, and specific bone mineral measurements in young women. *Osteoporosis International, 18*, 789-796.

Nummela, A. T., Heath, K. A. & Paavolainen, L. M. (2008). Fatigue during a running time trial. *International Journal of Sports Medicine, 29*, 738-745.

Ochsendorf, D. T., Mattacola, C. G. & Arnold, B. L. (2000). Effect of orthotics on postural sway after fatigue of the plantar flexors and dorsi flexors. *Journal of Athletic Training, 35*, 26-30.

O'Connor, D. M. (2004). Groin injuries in professional rugby league players: a prospective study. *Journal of Sports Science, 22*, 629-636.

Oda, S. & Kida, N. (2001). Neuromuscular fatigue during maximal concurrent hand grip and elbow flexion or extension. *Journal of Electromyography and Kinesiology, 11*, 281-289.

Oddsson, L. & Thorstensson, A. (1987). Fast voluntary trunk flexion movements in standing: motor patterns. *Acta Physiologica Scandinavica, 129*, 93-106.

O'Dwyer, C., Sainsbury, D. & O'Sullivan, K. (2011). Gluteus medius muscle activation during isometric muscle contractions. *Journal of Sports Rehabilitation, 20*, 174-186.

Olmsted, L. C., Carcia, C. R., Hertel, J. & Shultz, S. J. (2002). Efficacy of the star excursion balance tests in detecting reach deficits in subjects with chronic ankle instability. *Journal of Athletic Training, 37*, 501-506.

Orri, J. C. & Darden, G. F. (2008). Technical report: reliability and validity of the iSAM 9000 isokinetic dynamometer. *Journal of Strength and Conditioning Research, 22*, 310-317.

Orteza, L. C., Vogelbach, W. D., & Denegar, C. R. (1992). The effect of moulded and unmoulded orthotics on balance and pain while jogging following inversion ankle sprain. *Journal of Athletic Training, 27*, 80-84.

Osternig, L. R. (1975). Optimal isokinetic loads and velocities producing muscular power in human subjects. *Archives in Physical Medicine and Rehabilitation, 56*, 152-155.

Osternig, L. R. (1986). Isokinetic dynamometry: implications for muscle testing and rehabilitation. In K. Pandolf (Ed.), *Exercise and Sport Sciences Reviews*. (pp. 45-80). New York, USA: Macmillan Publishing Company.

Osternig, L. R. (2000). Assessing human performance. In L. Brown (Ed.), *Isokinetics in Human Performance*. (pp. 77-96). Champaign, IL: Human Kinetics Publishers.

Osternig, L. R., Sawhill, J. A., Bates, B. T. & Hamill, J. (1983). Function of limb speed on torque patterns of antagonist muscles. In H. Matsui & K. Kobayashi. (Eds.), *Biomechanics*. (pp. 63-75). Champaign, IL: Human Kinetics Publishers.

Ottaviani, R. A., Ashton-Miller, J. A., Kothari, S. U. & Wojtys, E. M. (1995). Basketball shoe height and the maximal muscular resistance to applied ankle inversion and eversion moments. *American Journal of Sports Medicine, 23*, 418-423.

Palastanga, N., Field, D. & Soames, R. (2002). *Anatomy of human movement.* (4th ed.). Edinburgh: Elsevier Science.

Papadopoulos, E. S., Nicolopoulos, C., Anderson, E. G., Curran, M., & Athanasopoulos, S. (2005). The role of ankle bracing in injury prevention, athletic performance and neuromuscular control: a review of the literature. *Journal of the Foot, 15*, 1-6.

Papadopoulos, E. S., Nicolopoulos, C., Baldoukas, A., Anderson, E. G., & Athanasopoulos, S. (2005). The effect of different ankle brace-skin interference application pressures on the electromyographic peroneus longus reaction time. *Journal of the Foot, 15*, 175-179.

Pasquet, B., Carpentier, A., Duchateau, J. & Hainaut, K. (2000). Muscle fatigue during concentric and eccentric contractions. *Muscle Nerve, 23*, 1727-1735.

Patterson, M. E., Nelson, S. G. & Duncan, P. W. (1984). Effects of stabilising the nontested lower extremity during isokinetic evaluation of the quadriceps and hamstrings. *Journal of Orthopaedic Sports Physical Therapy, 6*, 18-20.

Pincevero, D. M., Gear, W. S. & Sterner, R. L. (2001). Assessment of the reliability of high-intensity quadriceps femoris muscle fatigue. *Medicine and Science in Sports and Exercise, 33*, 334-338.

Pincevero, D. M., Lephart, S. M. & Karunakara, A. (1997). Reliability and precision of isokinetic strength and muscular endurance for the quadriceps and hamstrings. *International Journal of Sports Medicine, 18*, 113-117.

Piva, S. R., Teixeira, P. E., Almeida, G. J., Gil, A. B., DiGioia, A. M., Levison, T. J. & Fitzgerald, G. K. (2011). Contribution of hip abductor strength to physical function in patients with total knee arthroplasty. *Physical Therapy, 91*, 225-233.

Pope, R., Herbert, R. & Kirwan, J. (1998). Effects of ankle dorsiflexion range and pre-exercise calf muscle stretching on injury risk in Army recruits. *Australian Journal of Physiotherapy, 44*, 165-172.

Porter, G. K. & Kaminski, T. W. (2004). Knee angle does not affect ankle eversion to inversion peak torque ratios derived from an isokinetic dynamometer. *Isokinetics in Exercise Science, 12*, 99-104.

Powers, S. K. & Howley, E. T. (2004). *Exercise physiology: theory and application to fitness and performance.* (5th ed.). New York: McGraw-Hill.

Puffer, J. C. (2001). The sprained ankle. *Journal of Sports Medicine, 3*, 38-49.

Rahnama, N., Lees, A. & Reilly, T. (2006). Electromyography of selected lower limb muscles fatigued by exercise at the intensity of soccer match play. *Journal of Electromyograhy and Kinesiology, 16*, 257-263.

Rahnama, N., Reilly, T. & Lees, A. (2005). Science and football: the proceeding of the Fifth World Congress on Science and Football. In T. Reilly, J. Cabri & D. Arau'jo (Eds.), *Electromyographic analysis of selected muscles during a simulated soccer game*. (pp. 509-515). London: Routledge.

Rahnama, N., Reilly, T. & Lees, A. (2002). Injury risk associated with playing actions during competitive soccer. *British Journal of Sports Medicine, 36*, 354-359.

Rahnama, N., Reilly, T., Lees, A. & Graham-Smith, P. (2003). Muscle fatigue induced by exercise stimulating the work rate of competitive soccer. *Journal of Sports Sciences, 21*, 933-942.

Ramsdell, K. M., Mattacola, C. G., Uhl, T. L., McCrory, J. L. & Malone, T. R. (2001). Effects of two ankle fatigue models on the duration of postural stability dysfunction. *Journal of Athletic Training, 36*, 33-38.

Reid, D. (1992). *Assessment and rehabilitation of sports injury*. New York: Churchill Livingstone.

Reilly, T. & Thomas, V. (1976). A motion analysis of work rate in different positional roles in pro football match play. *Journal of Human Movement Studies, 2*, 87-97.

Richie, D. H. (2001). Functional instability of the ankle and the role of neuromuscular control: a comprehensive review. *The Journal of Foot and Ankle Surgery, 40*, 240-251.

Riemann, B. L. (2002). Is there a link between chronic ankle instability and postural instability? *Journal of Athletic Training, 37*, 386-393.

Riemann, B. L., Guskiewicz, K. M. & Shields, W. W. (1999). Relationship between clinical and forceplate measures of postural stability. *Journal of Sports Rehabilitation, 8*, 71-82.

Riemann, B. L. & Lephart, S. M. (2002). The sensorimotor system part 1: the physiologic basis of functional joint stability. *Journal of Athletic Training, 37*(1), 71-79.

Riemann, B. L., Myers, J. B., Stone, D. A. & Lephart, S. M. (2004). Effect of lateral ankle ligament anaesthesia on single leg stance stability. *Medicine and Science in Sports and Exercise, 36*, 388-396.

Rijke, A. M., Jones, B. & Vierhout, P. A. (1988). Injury to the lateral ankle ligaments of athletes: a post-traumatic follow up. *American Journal of Sports Medicine, 16*, 256-259.

Robbins, S., Waked, E. & Rappel, R. (1995). Ankle taping improves proprioception before and after exercise in young men. *British Journal of Sports Medicine, 29*, 242-247.

Roos, E. M., Brandsson, S. & Karlsson, J. (2001). Validation of the foot and ankle outcome score for ankle ligament reconstruction. *Foot and Ankle International, 22*, 788-794.

Ross, S. E. & Guskiewicz, K. M. (2004). Examination of static and dynamic postural stability in individuals with functionally stable and unstable ankles. *Clinical Journal of Sports Medicine, 14*, 332-338.

Ross, S. E., Guskiewicz, K. M. & Yu, B. (2005). Single-leg jump-landing stabilisation times in subjects with functionally unstable ankles. *Journal of Athletic Training, 40*, 298-304.

Rozzi, S., Yuktanandana, P., Pincivero, D. & Lephart, S. (2000). Role of fatigue on neuromuscular control. In S. Lephart & F. Fu (Eds.), *Proprioception and neuromuscular control in joint stability*. (pp. 375-383). Champaign, IL: Human Kinetics.

Ruth, C. J. (1961). The surgical treatment of injuries of the fibular collateral ligaments of the ankle. *Journal of Bone and Joint Surgery, 43*, 229-239.

Ryan, L. (1994). Mechanical stability, muscular strength and proprioception in the functionally unstable ankle. *Australian Journal of Physiotherapy, 40*, 41-47.

Sahlin, K., Tonkonogi, M. & Soderlund, K. (1998). Energy supply and muscle fatigue in humans. *Acta Physiologica Scandinavica*, *162*, 261-266.

Sainburg, R. L., Poizner, J. & Ghez, C. (1993). Loss of proprioception produces deficits in interjoint coordination. *Journal of Neurophysiology, 70*(5), 2136-2147.

Salavati, M., Moghadam, M., Ebrahimi, I. & Arab, A. M. (2007). Changes in postural stability with fatigue of lower extremity frontal and sagittal plane movers. *Gait Posture, 26*, 214-218.

Scibelli, A., Brown, L. E., Whitehurst, M., Bryant, J. R. & Buchalter, D. N. (1993). Load time during bilateral isokinetic knee extension/flexion. *Medicine and Science in Sports and Exercise, 25*, 108.

Sepic, S. B., Murray, M. P., Mollinger, L. A., Spurr, G. B. & Gardner, G. M. (1986). Strength and range of motion in the ankle in two groups of men and women. *American Journal of Physical Medicine, 65*, 75-84.

Simoneau, G. G., Degner, R. M., Kramper, C. A. & Kittleson, K. H. (1997). Changes in ankle joint proprioception resulting from strips of athletic tape applied over the skin. *Journal of Athletic Training, 32*, 141-147.

Singh, N. B., Nussbaum, M. A., Lin, D. & Madigan, M. L. (2005). Effect of localised muscle fatigue induced at different joints on postural control. *Human Factors and Ergonomics Society, 21*, 1306-1310.

Smith, R. W. & Reischl, S. F. (1986). Treatment of ankle sprains in young athletes. *American Journal of Sports Medicine, 14*, 465-471.

Sole, G., Hamren, J., Milosavljevic, S., Nicholson, H. & Sullivan, S. J. (2007). Test-retest reliability of isokinetic knee extension and flexion. *Archives of Physical Medicine and Rehabilitation, 88*, 626-631.

Spagenburg, E. E., Ward, C. W. & Williams, J. H. (1998). Effects of lactate on force production by mouse EDL muscle: implications for the development of fatigue. *Canadian Journal of Physiology and Pharmacology, 76*, 642-648.

Staples, O. S. (1975). Ruptures of the fibular collateral ligaments of the ankle: result study of immediate surgical treatments. *Journal of Bone and Joint Surgery, 57*, 101-107.

Staples, O. S. (1972). Result study of ruptures of lateral ligaments of the ankle. *Clinical Orthopaedics, 85*, 50-58.

Stasinopoulos, D. (2004). Comparison of three preventive methods in order to reduce the incidence of ankle inversion sprains among female volleyball players. *British Journal of Sports Medicine, 38*, 182-185.

Steele, C. (1994). *Anticipatory postural reactions in clumsy children: changes after physiotherapy treatment.* Unpublished PhD Thesis, University of Queensland, Australia.

Studenski, S., Duncan, P.W. & Chandler, J. (1991). Postural Responses and effector factors in persons with unexplained falls: results and methodological issues. *Journal of American Geriatrics, 39*, 229-234.

Swearingen, R. L. & Dehne, E. (1964). A study of pathological muscle function following injury to a joint. *American Journal of Bone and Joint Surgery, 46*, 1364-1369.

Taylor, J. L., Butler, J. E. & Gandevia, S. E. (2000). Changes in muscle afferents, motoneurons and motor drive during muscle fatigue. *European Journal of Applied Physiology, 83*, 106-115.

Taylor, N. A. S., Sanders, R. H., Howick, I. E. & Stanley, S. N. (1991). Static and dynamic assessment of the Biodex dynamometer. *European Journal of Applied Physiology, 62*, 180-188.

Tesch, P., Sjodin, B., Thorstensson, A. & Karlsson, J. (1978). Muscle fatigue and its relation to lactate accumulation and LDH activity in man. *Acta Physiologica Scandinavica, 103,* 413-420.

Thompson, H. W. & McKinley, P. A. (1995). Landing from a jump: the role of vision when landing from known and unknown heights. *Neuro Report, 6*, 581-584.

Thorstensson, A. & Karlsson, J. (1976). Fatiguability and fibre composition of human skeletal muscle. *Acta Physiologica Scandinavica*, 98, 318-322.

Timm, K. E. & Fyke, D. (1993). The effect of test speed sequence on the concentric isokinetic performance of the knee extensor muscle group. *Isokinetics in Exercise Science, 3,* 123-128.

Tohyama, H., Yasuda, K., Beynnon, B. D. & Renstrom, P. A. (2006). Stabilising effects of ankle bracing under a combination of inversion and axial compression loading. *Journal of Knee Surgery, Sports Traumatology and Arthroscopy, 14*, 373-378.

Trojian, T. H. & McKeag, D. B. (2006). Single leg balance test to identify risk of ankle sprains. *British Journal of Sports Medicine, 40*, 610-613.

Tropp, H. (2002). Commentary: functional ankle instability revisited. *Journal of Athletic Training, 37*, 512-515.

Tropp, H. (1986). Pronator muscle weakness in functional instability of the ankle joint. *International Journal of Sports Medicine, 5*, 291-294.

Tropp, H., Ekstrand, J. & Gillquist, J. (1984). Stabilometry in functional instability of the ankle and its value in predicting injury. *Journal of Medical Science of Sport and Exercise, 16*, 64-66.

Tropp, H. & Odenrick, P. (1988). Postural control in a single limb stance. *Journal of Orthopaedic Research, 6*, 833-839.

Tropp, H., Odenrick, P. & Gillquist, J. (1985). Stabilometry recordings in functional and mechanical instability of the ankle joint. *International Journal of Sports Medicine, 6*, 180-182.

Tyler, T. F., Nicholas, S. J., Campell, R. J. & McHugh, M. P. (2001). The association of hip strength and flexibility with the incidence of adductor muscle strains in professional ice hockey players. *American Journal of Sports Medicine, 29*, 124-128.

Vaes, P., Duquet, W. & Van Gheluwe, B. (1999). Peroneal reaction times and eversion motor responses in healthy and unstable ankles. . *Journal of Athletic Training, 37*, 475-480.

Van Deun, S., Staes, F. F., Stappaerts, K. H., Janssens, L., Levin, I. O. & Peers, K. H. (2007). Relationship of chronic ankle instability to muscle activation patterns during the transition from double leg to single leg stance. *American Journal of Sports Medicine, 35*, 274-281.

Van Gool, D., Van Gervan, D. & Boutmans, J. (1988). The physiological load imposed on soccer players during real match play. In T. Reilly, A. Lees, K. Davids, W. Murphy (Eds.), *Science and Football*. (pp. 51-59). London: E and FN SPON.

Waddington, G. S. & Adams, R. (1999). Ability to discriminate movements at the ankle and knee is joint specific. *Perception of Motor Skills, 89*, 1037-1041.

Walmsley, R. P. & Szybbo, C. (1987). A comparative study of the torque generated by the shoulder internal and external rotator muscles in different positions and at varying speeds. *Journal of Orthopaedic Sports Physical Therapy, 9*, 217-222.

Wank, V., Frick, U., Schmidtbleicher, D. (1998). Kinematics and electromyography of lower limb muscles in overground and treadmill running. *International Journal of Sports Medicine, 19*, 455-461.

Weir, J. P. (2005). Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. *Journal of Strength and Conditioning Research*, *19*, 231-240.

Westerblad, H. & Allen, D. G. (2002). Recent advances in the understanding of skeletal muscle fatigue. *Current Opinion in Rheumatology, 14*, 648-652.

Widrick, J. J., Trappe, S. W., Costill, D. L. & Fitts, R. H. (1996). Force-velocity and force-power properties of single muscle fibres from elite master runners and sedentary men. *American Journal of Physiology, 271*, C676-C683.

Wiesler, E. R., Hunter, D. M. & Martin, D. F. (1996). Ankle flexibility and injury patterns in dancers. *American Journal of Sports Medicine, 24*, 754-757.

Wikstrom, E. A., Bishop, M., Inamdar, A. D. & Hass, C. J. (2010). Gait termination control strategies are altered in chronic ankle instability subjects. *Medicine and Science in Sports and Exercise, 42,* 197-205.

Wikstrom, E. A., Powers, M. & Tillman, M. (2004). Dynamic stabilisation time after isokinetic and functional fatigue. *Journal of Athletic Training, 39*, 247-253.

Wikstrom, E. A., Tillman, M. D., Chmielewski, T. L. & Borsa, P. A. (2006). Measurement and evaluation of dynamic joint stability of the knee and ankle after injury. *Journal of Sports Medicine, 36*, 393-410.

Wikstrom, E. A., Tillman, M. D., Chmielewski, T. L. & Cauraugh, J. H. (2007). Dynamic postural stability deficits in subjects with self-reported ankle instability. *Medicine and Science in Sports and Exercise, 39*, 397-402.

Wilk, K. E., Romaniello, W. T., Soscia, S. M., Arrigo, C. A. & Andrews, J. R. (1994). The relationship between subjective knee scores, isokinetic testing, and functional testing in the ACL reconstructed knee. *Journal of Orthopaedic and Sports Physical Therapy, 20*, 60-70.

Wilkerson, G. B. & Nitz, A. J. (1994). Dynamic ankle stability: mechanical and neuromuscular interrelationships. *Journal of Sports Rehabilitation, 3*, 43-57.

Wilkerson, G. B., Pinerola, J. & Caturano, R. (1997). Invertor versus evertor torque and power deficiencies associated with lateral ankle ligament injury. *Journal of Orthopaedic and Sports Physical Therapy, 26*, 78-86.

Wilkinson, R. T. & Allison, S. (1989). Age and simple reaction time: decade differences for 5,325 subjects. *Journal of Gerontology: Psychological Sciences, 44*(2), 29-35.

Willems, T., Witvrouw, E., Verstuyft, J., Vaes, P. & Clercq, D. D. (2002). Proprioception and muscle strength in subjects with a history of ankle sprains and chronic ankle instability. *Journal of Athletic Training, 37*, 487-493.

Wilson, E. L. & Madigan, M. L. (2007). Effets of fatigue and gender on peroneal reflexes elicited by sudden ankle inversion. *Journal of Electromyography and Kinesiology, 17*, 160-166.

Winter, D. A. (2005). Kinetics: Forces and moments of force. In D. A. Winter (Ed), *Biomechanics and Motor Control of Human Movement*. (pp. 106-110). Hoboken, New Jersey: Wiley Publishers.

Winter, D. A., Patla, A. E. & Frank, J. S. (1990). Assessment of balance control in humans. *Medical Program for Technology, 16*, 31-51.

Winter, D., Prince, F., Frank, J. S., Powell, C. & Zabjek, K. F. (1996). Unified theory regarding A/P and M/L balance in quiet stance. *Journal of Neurophysiology, 75*, 2334-2343.

Winter, D. A., Wells, R. P. & Orr, G. W. (1981). Error in the use of isokinetic dynamometers. *European Journal of Applied Physiology, 46*, 397-408.

Wolfe, M. W., Uhl., T. L., Mattacola, C. G. & McCluskey, L. C. (2001). Management of ankle sprains. *American Family Physician, 63*, 93-104.

Wong, D. L. K., Glasheen-Wray, M. & Andrews, L. F. (1984). Isokinetic evaluation of the ankle invertors and evertors. *Journal of Orthopaedic and Sports Physical Therapy, 5,* 246-252.

Woods, C., Hawkins, R., Hulse, M. & Hodson, A. (2002). The football association medical research programme: an audit of injuries in professional football – analysis of preseason injuries. *British Journal of Sports Medicine, 36*, 436-441.

Woods, C., Hawkins, R., Hulse, M. & Hodson, A. (2003). The football association medical research programme: an audit of injuries in professional football: an analysis of ankle sprains. *British Journal of Sports Medicine, 37*, 233-238.

Woollacott, M. H., Von Hosten, C. & Rosblad, B. (1988). Relation between muscle response onset and body segmental movements during postural perturbations in humans. *Experimental Brain Research, 72*, 593-604.

Yaggie, J. & Armstrong, W. J. (2004). Effects of lower extremity fatigue on indices of balance. *Journal of Sports Rehabilitation, 13*, 312-322.

Yaggie, J.A. & McGregor, S. J. (2002). Effects of isokinetic ankle fatigue on the maintenance of balance and postural limits. *Archives of Physical Medicine and Rehabilitation, 83*, 224-228.

Yamanaka, K., Haga, S., Shindo, M., Narita, J., Koeski, S., Matsuura, Y. & Eda, A. (1988). Time and motion analysis in top class soccer games. In T. Reilly, A. Lees, K. Davids, W. Murphy (Eds.), *Science and Football*. (pp. 334-340). London: E and FN SPON.

Yeung, S. S., Au, A. L. & Chow, C. C. (1999). Effects of fatigue on the temporal neuromuscular control of vastus medialis muscle in humans. *European Journal of Applied Physiology and Occupational Physiology, 80*, 379-385.

Yeung, M. S., Chan, K. M., So, C. H. & Yuan, W. Y. (1994). An epidemiological survey on ankle sprains. *British Journal of Sports Medicine, 28*, 112-116.

Youdas, J. W., Loder, E. F., Moldenhauer, J. L., Paulsen, C. R. & Hollman, J. H. (2006). Hip-abductor muscle performance in participants after 45 seconds of resisted side stepping using an elastic band. *Journal of Sports Rehabilitation, 15*, 1-11.

Zatterstrom, R., Frieden, T. & Lindstrand, A. (1994). The effect of physiotherapy on standing balance in chronic anterior cruciate ligament insufficiency. *American Journal of Sports Medicine, 22*, 531-536.

Zawadzki, J., Bober, T. & Siemienski, A. (2010). Validity analysis of the Biodex system 3 dynamometer under static and isokinetic conditions. *Acta of Bioengineering and Biomechanics, 12*, 25-32.

Appendices

UNIVERSITY OF HERTFORDSHIRE
FACULTY OF HEALTH & HUMAN SCIENCES

Subject Briefing Document

Researcher: Charlotte Gautrey

Title of Study: The biomechanics of the dynamic defense mechanism

Introduction:

Ankle sprains are a common injury in sport and physically active individuals. Following an ankle sprain many individuals will develop recurrent ankle sprains and more long standing chronic ankle problems. This study will look at the effects of globalised and localised fatigue on muscle latency and postural sway, in FAI and healthy subjects. Research on elite soccer players has shown that injury risk is highest in the last 15 minutes of the contest (Rahnama, Reilly and Lees, 2002), when fatigue has set in. The current project will research whether a globalised fatiguing protocol and localised fatiguing protocols lead to an increase in muscle latency and postural sway, and therefore may be detrimental for the sports person.

Am I eligible to take part in the study?

If you are male and you are a member of the University of Hertfordshire then you are eligible to take part in this study. If you have suffered at least two lateral ankle sprains, and these sprains have occurred on the same ankle, you may be assigned to the functional ankle instability (FAI) group. If you have not suffered from an ankle sprain on either of your ankles you may be assigned to the healthy control group.

What is involved?

You will be required to come into the laboratory at the University (either G111 or H260). You will be required to attend the laboratory on 4 different occasions, approximately 1 hour long each time, at least 7 days apart.

Before either of the procedures begins you will be required to read this subject briefing form, fill out the health screen questionnaire, and FAI questionnaire, and sign the consent form. After filling out these forms the investigator will then perform a short clinical assessment on your ankle (this will involve the examiner performing a couple of ligament tests to check for mechanical instability). You will then begin by performing a 5 minute warm up on the cycle ergometer.

You will then be prepared for electromyography (EMG) set up. You will be required to shave two small patches on your lower leg (disposable razors and shaving gel will be provided). The shaved skin will then be cleaned with an alcohol wipe before application of the electrodes. Electrodes will be applied to the peroneus longus, extensor digitorum longus, tibialis anterior and gluteus medius muscles. Electrodes will be applied to both of your legs (Figure 1).

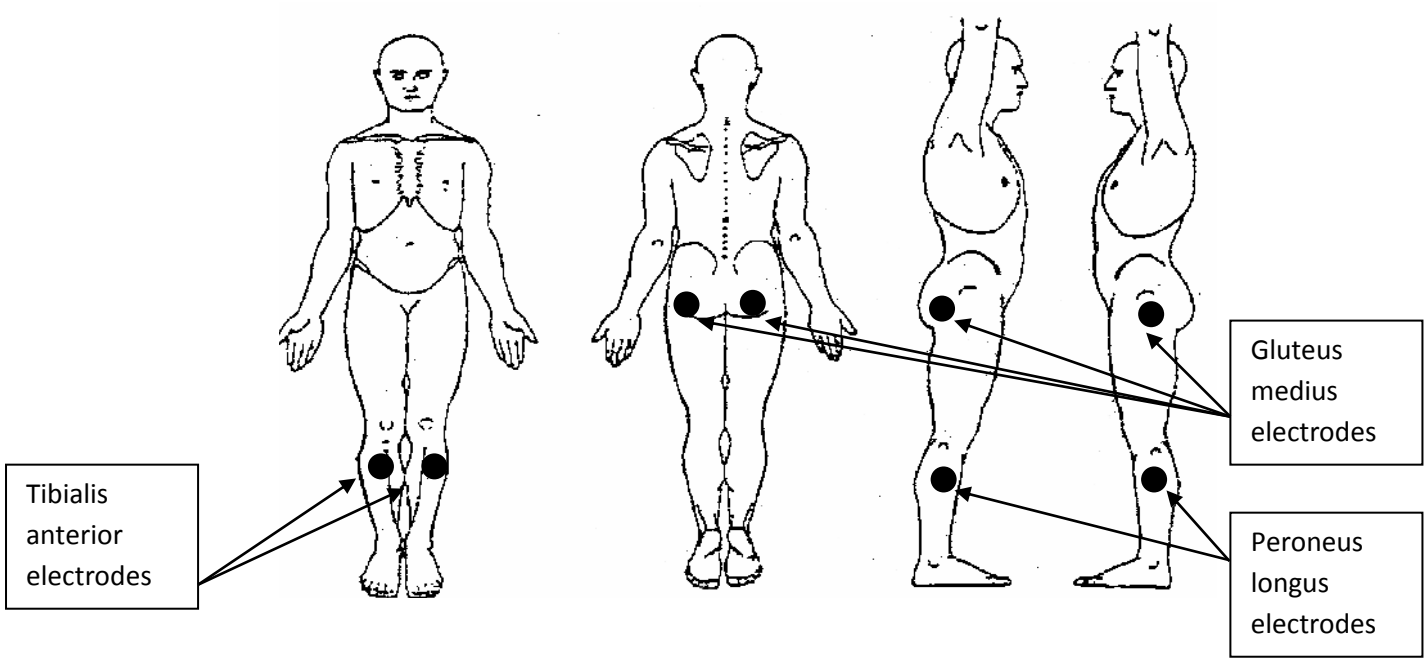


Figure 1. Anterior, posterior and lateral views of where the electrodes will be placed.

Procedure: The effects of globalised and localised fatigue on muscle latency and postural sway.

The globalised fatigue procedure will involve you coming into the laboratory and performing a running football-specific protocol on the treadmill which will last 105 minutes (45 minutes exercise, 15 minutes rest, 45 minutes exercise). The localised fatigue protocols will involve you coming into the laboratory and performing a hip and ankle fatigue protocol on the isokinetic dynamometer. The final protocol is a control test where you will be required to sit quietly for 105 minutes. Following the completion of a fatigue protocol, muscle latency and postural sway will be measured.

To measure muscle latency you will be required to stand on a tilt platform. You will have your eyes open and will be told that at some point in the next 30 seconds the tilt platform will tilt. For the postural sway procedure you will be required to perform a 30 cm drop jump from a wooden bench onto a force plate. You will be required to stand on a bench with the test leg relaxed and non-weight bearing. You will then use the other leg to propel yourself from the bench and balance your landing on the test leg on the centre of the force plate. Each procedure will be repeated up to five times on each leg. The procedures will be performed before the fatiguing protocols, and immediately after the fatiguing protocols. After performing the procedures you will be required to perform a 5 minute cool down on the cycle ergometer.

How will subjects be briefed?

You will be given written information about what the study will involve. If you have any questions about the study you are welcome to ask questions which will be answered by the investigator Charlotte Gautrey.

Participation details

You will be asked to wear a polo t-shirt and sports shorts. A few small sections of hair on your lower limbs will need to be shaved by yourself in order to improve the conduction of the EMG electrodes. A new disposable razor blade will be used for each subject to prevent cross infection.

When should I refuse to take part?

If you suffer from any of the below criteria you will be excluded from the study:

- If you are under the influence of alcohol or any other psycho-active substance
- If you have had a cold, flu, inner ear or sinus infection in the last two weeks
- If you suffer from any musculo-skeletal injuries
- If you have a history of fractures to the lower limbs
- If you suffer from any visual impairments that may affect your balance
- If you suffer from any ear problems that may affect your balance (such as ear infections, hearing problems or vertigo)
- If you have any signs of injury such as pain, tenderness, soft tissue inflammation or acute trauma in your ankles
- If you have ever been told by a doctor that you should not exercise
- If you do not take part in regular ($\geq$ 2 x wk) aerobic exercise
- If you do not feel fully fit, and eager to act as a subject

You are not obliged to participate in this study and you may withdraw from the study at any stage without prejudice or having to give a reason.

What are the adverse effects?

You may feel aching in your lower limb muscles after participating in the fatigue section of this study; this will be temporary and will reduce within a few days. The investigator is first aid trained and is able to provide initial treatment of any injuries.

Consent

Consent will be obtained by signing the LEC2 consent form. You should be aware that your participation in the study is voluntary, and you may discontinue at any time, without prejudice. If, after consenting to participate you withdraw your consent, any information already obtained will be removed from the study.

Personal data

Personal data will be collected to ensure that you match the criterion that is required for this study. Name, age, gender, weight and height will be required. You will also fill out a health and FAI questionnaire to determine whether you suffer from FAI.

All paper and questionnaires containing personal data will be securely locked away in a desk drawer. No names will be identified in the write up of this study. When the project has been completed, written up, and marked, all data will be destroyed by the means of a paper shredder, and any computer data will be deleted.

UNIVERSITY OF HERTFORDSHIRE **Form LEC2**
FACULTY OF HEALTH & HUMAN SCIENCES

School of Life Sciences Ethics

CONSENT FORM

I, the undersigned, agree to take part in:

Approved Protocol Number or Registered Project Number **LS6/11/07P(R2/08)**

Title of Study The biomechanics of the dynamic defense mechanism

to be carried out by

Name of Investigator(s) Charlotte Gautrey

I confirm that the purpose of the study has been explained to me by the investigator and that I have been informed of the details of my involvement in the study.

I confirm that my questions regarding involvement with this study have been answered to my satisfaction.

I confirm that I understand that I am not obliged to participate in this study and that I may withdraw from the study at any stage without the need to justify my decision and without personal disadvantage.

I understand that any personal information I consent to provide will be treated as confidential and will not be made publicly available without seeking any further consent.

Name of subject.....

Signature of subject.....**Date.**.....

Name of investigator.....

Signature of investigator.....**Date**.....

THIS FORM TOGETHER WITH THE PROTOCOL MONITORING FORM (LEC5), SHOULD BE GIVEN TO THE UNIVERSITY SUPERVISOR ON COMPLETION OF THE PROJECT WORK. SUPERVISORS ARE RESPONSIBLE FOR FORWARDING BOTH THE CONSENT FORMS AND A PROTOCOL MONITORING FORM TO THE SECRETARY OF THE LIFE SCIENCES ETHICS COMMITTEE.

UNIVERSITY OF HERTFORDSHIRE
FACULTY OF HEALTH & HUMAN SCIENCES

HEALTH SCREEN QUESTIONNAIRE

Name: **Date:**

It is important when having volunteered as subject for this study, and having read the subject briefing sheet that you answer the following questions.

Are you under the influence of alcohol or any other psycho-active substance?

Yes	No

Have you had a cold, flu, inner ear or sinus infection in the last two weeks?

Yes	No

Are you suffering from any musculo-skeletal injury?

Yes	No

4) Do you have a history of fractures to the lower limbs?

Yes	No

5) Do you suffer from any visual impairments that may affect your balance?

Yes	No

6) Do you suffer from any ear problems that may affect your balance (such as hearing problems, ear infection or vertigo)?

Yes	No

7) Do you have any signs of injury such as pain, tenderness, soft tissue inflammation or acute trauma in your ankles?

Yes	No

8) Have you ever been told by a doctor that you should not exercise?

Yes	No

9) Are you engaged in regular (≥ 2 x week) aerobic exercise?

Yes	No

10) Do you feel fully fit, and eager to act as subject?

Yes	No

Subjects
Signature……Date……..……

Checked by
(Name)…….....…...Date……...……

UNIVERSITY OF HERTFORDSHIRE
FACULTY OF HEALTH AND HUMAN SCIENCES

FUNCTIONAL ANKLE INSTABILITY QUESTIONNAIRE

Name: **Date:**

Part 1: Functional Ankle Instability Questionnaire

1.	Concerning your purported ankle instability, does this injury involve only one ankle? If yes, did the initial episode involve your ankle "rolling inwards"? If no, do not continue to fill out this questionnaire?	Y Y	N N
2.	Which ankle suffers the instability?	L	R
3.	Did the initial injury to your ankle require crutches, immobilisation, or both, of any form (cast, braces, etc)?	Y	N
4.	Have you had any fractures (breaks) in either of your ankles?	Y	N
5.	Is the injured/unstable ankle functionally weaker, more painful, "looser," and less functional than your uninvolved ankle?	Y	N
6.	Do you ever have episodes of your ankle "giving way" or "rolling over" during daily activity (athletic or otherwise)?	Y	N
7.	Do you attribute you current instability to past injuries to the affected ankle?	Y	N
8.	Have you had an episode of injury ("your ankle was hurt," "you were in great pain") to the affected ankle in the last 3 months?	Y	N
9.	Have you been walking around unassisted without a "limp," for at least the last 3 months?	Y	N
10.	Are you currently involved in a "formal" rehabilitation programme for the affected ankle? If you answered yes, please describe here. ----------	Y	N
11.	Can you describe a symptom(s) of your ankle "giving way"? ----------		

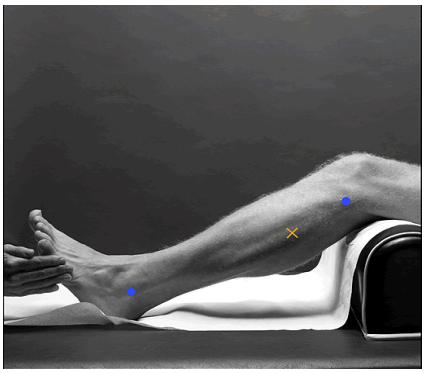
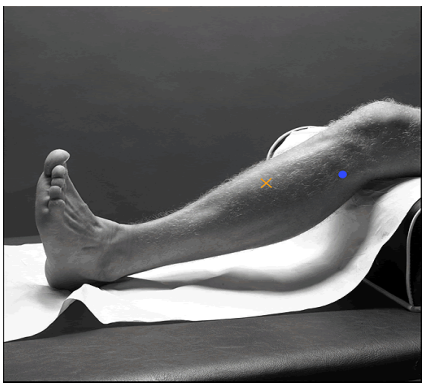
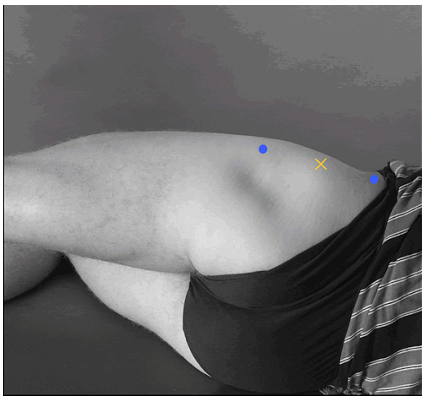
Part 2: Clinical Examination of Ankle Stability

Is there swelling present?	Y N
Is there ecchymosis present?	Y N
Anterior Drawer Test:	
Right Ankle	+pos -neg
Left Ankle	+pos -neg
Talar Tilt Test:	
Right Ankle	+pos -neg
Left Ankle	+pos -neg
Cleared for participation in the study?	Y N

Investigators signature_____

Note: To qualify as functional ankle instability, questions 3, 5, 6, 7 and 9 should be answered "yes." Questions 4, 8 and 10 should be answered "no," and no clinical signs of mechanical instability can be present.

SENIUM Guidelines for Electrode Placement

Muscle	Electrode Placement	
Peroneus Longus	Electrodes need to be placed at 25% on the line between the tip of the head of the fibula to the tip of the lateral malleolus; in the direction of the line between the two points.	
Tibialis Anterior	The electrodes need to be placed at a 1/3 on the line between the tip of the fibula and the tip of the medial malleolus; in the direction of the line between the two points.	
Gluteus Medius	Electrodes need to be placed at 50% on the line from the crista iliaca to the greater trochanter; in the direction of the line between the two points.	

SENIUM: www.senium.org; indicating anatomical landmarks and positioning of surface electrodes.

Muscle Latency Determination Example (only showing left gluteus medius, peroneus longus and tibialis anterior muscles)

CODE	LWtilt2Right								
DATE	23/07/2008								
Mean		0.001887			0.002177			0.002626	
SD		0.000905			0.001206			0.001529	
3SD		0.002715			0.003619			0.004586	
Mean+3SD		0.004602			0.005796			0.007212	
Tilt		14.335			14.335			14.335	
Onset		14.394			14.395			14.397	
RT (ms)		0.059			0.060			0.062	

Time (ms)	Mean+3SD	Glutmed L	0/1	Mean+3SD	PL L	0/1	Mean+3SD	Tibant L	0/1
0.001	0.0046018	0.0022	0	0.0057958	0.0007	0	0.0072122	0.0015	0
0.002	0.0046018	0.0015	0	0.0057958	0.0007	0	0.0072122	0.0015	0
0.003	0.0046018	0.0015	0	0.0057958	0.0007	0	0.0072122	0.0015	0
0.004	0.0046018	0.0015	0	0.0057958	0.0022	0	0.0072122	0.0015	0
Time 0.005 to 14.38 missing									
14.39	0.0046018	0.0015	0	0.0057958	0.0022	0	0.0072122	0.0045	0
14.391	0.0046018	0.003	0	0.0057958	0.0015	0	0.0072122	0.0045	0
14.392	0.0046018	0.0037	0	0.0057958	0.0015	0	0.0072122	0.0045	0
14.393	0.0046018	0.0037	0	0.0057958	0.003	0	0.0072122	0.0052	0
14.394	0.0046018	0.0052	1	0.0057958	0.003	0	0.0072122	0.0052	0
14.395	0.0046018	0.0082	1	0.0057958	0.009	1	0.0072122	0.0052	0
14.396	0.0046018	0.0105	1	0.0057958	0.0202	1	0.0072122	0.0052	0
14.397	0.0046018	0.0135	1	0.0057958	0.0255	1	0.0072122	0.0112	1
14.398	0.0046018	0.0172	1	0.0057958	0.0277	1	0.0072122	0.0187	1
14.399	0.0046018	0.0195	1	0.0057958	0.0285	1	0.0072122	0.0202	1
14.4	0.0046018	0.018	1	0.0057958	0.0277	1	0.0072122	0.0202	1

Where the green columns start represents the onset of muscle latency, also highlighted by the green numbers.

The muscle reaction time was determined by subtracting tilt onset time (above in yellow) from muscle onset time (above page in green). These values were inserted into the subject's raw data tables, mean values were then calculated.

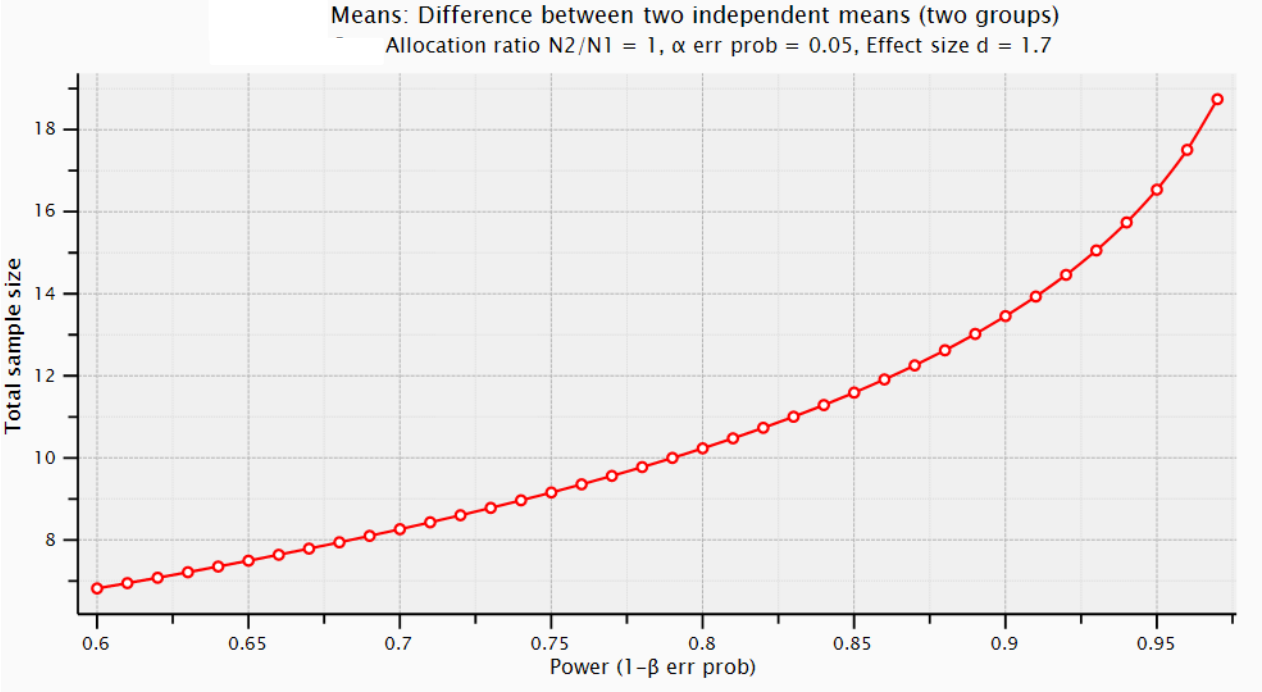
Muscle Latency Determination Example Showing Tilt Onset

CODE	LWtilt2Right			
DATE	23/07/2008			
Mean		0.001887	0.002177	0.002626
SD		0.000905	0.001206	0.001529
3SD		0.002715	0.003619	0.004586
Mean+3SD		0.004602	0.005796	0.007212
Tilt		14.335	14.335	14.335
Onset		14.394	14.395	14.397
RT (ms)		0.059	0.060	0.062

Time (ms)	Mean+3SD	Glutmed L	0/1	Mean+3SD	PL L	0/1	Mean+3SD	Tibant L	0/1	Time(ms)	Tilt
0.001	0.0046018	0.0022	0	0.0057958	0.0007	0	0.0072122	0.0015	0	0.001	31
0.002	0.0046018	0.0015	0	0.0057958	0.0007	0	0.0072122	0.0015	0	0.002	31
0.003	0.0046018	0.0015	0	0.0057958	0.0007	0	0.0072122	0.0015	0	0.003	31
0.004	0.0046018	0.0015	0	0.0057958	0.0022	0	0.0072122	0.0015	0	0.004	31
Time 0.005 to 2.863 missing											
2.864	0.0046018	0.0015	0	0.0057958	0.0007	0	0.0072122	0.003	0	14.332	31
2.865	0.0046018	0.0015	0	0.0057958	0.0007	0	0.0072122	0.003	0	14.333	31
2.866	0.0046018	0.0007	0	0.0057958	0.0007	0	0.0072122	0.003	0	14.334	31
2.867	0.0046018	0.0007	0	0.0057958	0.0022	0	0.0072122	0.003	0	14.335	23
2.868	0.0046018	0.0007	0	0.0057958	0.003	0	0.0072122	0.0015	0	14.336	23
2.869	0.0046018	0.0007	0	0.0057958	0.003	0	0.0072122	0.0015	0	14.337	23
2.87	0.0046018	0.0007	0	0.0057958	0.0037	0	0.0072122	0.0015	0	14.338	23
2.871	0.0046018	0.0007	0	0.0057958	0.0045	0	0.0072122	0.0015	0	14.339	23
2.872	0.0046018	0.0007	0	0.0057958	0.0037	0	0.0072122	0.0015	0	14.340	23

Tilt onset was determined when the numbers changed from 31 to 23 – also highlighted by when the column changes to yellow. The time of onset was then confirmed by reading the adjacent time in ms – also highlighted in yellow.

The muscle reaction time was determined by subtracting tilt onset time (above in yellow) from muscle onset time (above page in green). These values were inserted into the subject's raw data tables, mean values were then calculated.



The Effect Size Generator (Version 2.3.0, Australia) software was used to calculate Cohen's d effect size. As three muscles were tested in Pilot Study One, the muscle and condition that generated the smallest effect size was used for the calculation of power. The above plot represents the tibialis anterior muscle, when acting as a tilt limb. An effect size of d = 1.7 was calculated from the healthy groups NDA's muscle latency (mean = 49.56 ms, SD = 3.1) and the FAI groups SA's muscle latency (mean = 54.87, SD = 3.2). The plot produced by G*Power (Version 3.1.5, Germany) (Faul, Erdfelder, Lang & Buchner, 2007) indicated that a total sample size of 18 subjects would be needed to achieve a power of >95% between two independent groups. Therefore, in Study One and Study Three of this thesis a minimum of 9 subjects will be required in each group.

Muscle Latencies for the Control Group's Dominant Ankle when Acting as the Tilt Limb.

COMBINATION (Sample Rate–Analysis Method–Smoothing)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
1000 Hz–RMS–2 ms	48.45(4.2)	48.40(4.5)	0.91	2.25	49.55(4.8)	50.57(4.6)	0.90	2.28	52.11(4.8)	52.12(4.6)	0.85	2.35
1000 Hz–RMS–5 ms	48.47(4.3)	48.50(4.4)	0.84	2.32	49.56(4.3)	50.55(4.9)	0.83	2.34	52.14(4.1)	52.11(4.4)	0.81	2.37
1000 Hz–RMS–10 ms	48.46(4.2)	49.46(4.5)	0.80	2.45	48.58(3.1)	48.59(3.7)	0.78	2.43	52.15(4.7)	52.13(4.3)	0.77	2.48
1000 Hz–AVR–2 ms	48.47(4.1)	49.49(4.2)	0.81	2.53	48.57(3.3)	48.59(3.8)	0.80	2.54	52.14(4.1)	52.13(4.6)	0.77	2.57
1000 Hz–AVR–5 ms	49.50(3.2)	49.48(3.5)	0.80	2.62	49.59(3.1)	49.60(3.7)	0.78	2.68	52.15(4.7)	52.16(4.0)	0.75	2.70
1000 Hz–AVR–10 ms	49.53(3.7)	49.54(3.4)	0.77	2.65	49.60(4.3)	49.59(3.7)	0.75	2.67	52.17(4.2)	52.18(4.6)	0.73	2.68
2500 Hz–RMS–2 ms	49.55(3.3)	49.57(3.5)	0.84	2.35	49.63(4.9)	49.64(3.2)	0.81	2.36	52.18(3.7)	52.20(3.7)	0.78	2.30
2500 Hz–RMS–5 ms	49.56(3.7)	49.58(3.9)	0.83	2.39	49.63(3.3)	49.63(3.6)	0.81	2.35	52.20(3.2)	52.25(3.6)	0.84	2.39
2500 Hz–RMS–10 ms	49.58(3.3)	49.57(4.5)	0.80	2.41	49.64(3.2)	49.66(4.0)	0.80	2.42	52.23(3.9)	52.25(4.0)	0.70	2.48
2500 Hz–AVR–2 ms	49.57(3.9)	49.59(4.0)	0.81	2.60	49.63(3.3)	49.65(4.5)	0.77	2.66	52.25(3.3)	52.22(4.6)	0.73	2.54
2500 Hz–AVR–5 ms	49.58(3.3)	49.60(3.5)	0.80	2.63	49.65(4.0)	49.67(3.7)	0.78	2.65	52.23(3.5)	52.27(4.3)	0.72	2.65
2500 Hz–AVR–10 ms	49.60(3.1)	49.61(3.7)	0.77	2.53	49.67(4.3)	49.69(3.5)	0.75	2.53	52.26(3.3)	52.28(4.5)	0.69	2.60
5000 Hz–RMS–2 ms	49.60(3.3)	49.63(3.4)	0.76	2.60	49.69(3.7)	49.71(4.1)	0.73	2.66	52.26(3.9)	52.29(3.0)	0.70	2.62
5000 Hz–RMS–5 ms	49.61(3.5)	49.62(3.3)	0.75	2.65	49.70(3.3)	49.69(4.2)	0.73	2.65	52.28(3.4)	52.25(3.5)	0.68	2.67
5000 Hz–RMS–10 ms	49.66(3.4)	49.65(4.3)	0.72	2.67	49.71(4.0)	49.70(3.6)	0.71	2.64	52.27(3.6)	52.30(3.8)	0.63	2.66
5000 Hz–AVR–2 ms	49.61(3.9)	49.60(4.1)	0.72	2.66	49.70(4.3)	49.74(3.2)	0.69	2.63	52.30(3.4)	52.28(3.4)	0.62	2.65
5000 Hz–AVR–5 ms	49.63(3.4)	49.62(4.3)	0.67	2.71	49.72(4.3)	49.73(4.4)	0.65	2.74	52.31(4.7)	52.30(3.8)	0.58	2.73
5000 Hz–AVR–10 ms	49.64(3.9)	49.65(4.0)	0.63	2.63	49.72(4.2)	49.74(4.2)	0.65	2.65	52.33(4.2)	52.32(3.4)	0.55	2.72

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement

Muscle Latencies for the Control Group's Dominant Ankle when Acting as the Support Limb.

COMBINATION (Sample Rate–Analysis Method–Smoothing)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
1000 Hz–RMS–2 ms	64.16(4.1)	64.18(4.3)	0.81	2.35	65.34(3.3)	65.33(2.4)	0.80	2.30	66.48(4.3)	66.47(3.7)	0.85	2.45
1000 Hz–RMS–5 ms	64.17(4.9)	64.15(4.7)	0.74	2.42	65.35(3.7)	65.34(2.8)	0.73	2.33	66.49(4.9)	66.48(3.9)	0.71	2.47
1000 Hz–RMS–10 ms	64.20(4.2)	64.16(4.3)	0.70	2.55	65.37(3.4)	65.39(2.3)	0.68	2.43	66.50(3.3)	66.49(3.7)	0.67	2.58
1000 Hz–AVR–2 ms	64.18(4.8)	64.22(4.4)	0.71	2.63	65.34(3.9)	65.33(2.7)	0.70	2.44	66.48(3.9)	66.45(3.8)	0.67	2.47
1000 Hz–AVR–5 ms	64.20(3.1)	64.26(4.4)	0.70	2.65	65.36(3.4)	65.37(3.4)	0.68	2.58	66.50(4.3)	66.51(4.8)	0.65	2.60
1000 Hz–AVR–10 ms	64.22(3.8)	64.28(4.8)	0.67	2.65	65.37(3.9)	65.39(3.0)	0.65	2.57	66.51(4.2)	66.52(4.5)	0.63	2.58
2500 Hz–RMS–2 ms	64.22(3.3)	64.24(3.3)	0.74	2.45	65.38(4.4)	65.40(4.3)	0.71	2.56	66.55(3.3)	66.56(4.7)	0.68	2.40
2500 Hz–RMS–5 ms	64.23(3.1)	64.25(3.7)	0.73	2.49	65.39(4.9)	65.41(4.5)	0.71	2.55	66.54(3.7)	66.52(4.0)	0.74	2.49
2500 Hz–RMS–10 ms	64.25(3.1)	64.27(3.4)	0.70	2.51	65.37(3.4)	65.39(3.4)	0.70	2.42	66.57(3.3)	66.58(4.7)	0.60	2.58
2500 Hz–AVR–2 ms	64.23(3.0)	64.28(3.9)	0.71	2.60	65.39(3.0)	65.43(3.1)	0.67	2.56	66.60(3.9)	66.65(4.5)	0.63	2.44
2500 Hz–AVR–5 ms	64.26(3.3)	64.27(3.4)	0.70	2.73	65.41(3.4)	66.43(3.4)	0.68	2.55	66.61(3.5)	66.62(4.8)	0.62	2.55
2500 Hz–AVR–10 ms	64.28(3.9)	64.24(3.0)	0.67	2.63	65.40(3.1)	66.42(3.8)	0.65	2.43	66.53(3.7)	66.55(4.6)	0.59	2.50
5000 Hz–RMS–2 ms	64.30(3.3)	64.32(3.1)	0.66	2.70	65.43(3.4)	65.46(3.4)	0.63	2.56	66.60(2.7)	66.63(3.8)	0.60	2.52
5000 Hz–RMS–5 ms	64.28(3.1)	64.29(3.2)	0.65	2.75	65.45(3.7)	65.47(3.0)	0.63	2.55	66.62(2.9)	66.65(3.7)	0.58	2.57
5000 Hz–RMS–10 ms	64.32(3.3)	64.33(3.3)	0.62	2.77	65.47(3.3)	65.43(4.3)	0.61	2.54	66.64(3.5)	66.62(3.8)	0.53	2.56
5000 Hz–AVR–2 ms	64.30(3.9)	64.35(3.8)	0.62	2.76	65.45(3.4)	65.46(4.8)	0.59	2.53	66.65(3.7)	66.66(3.9)	0.52	2.55
5000 Hz–AVR–5 ms	64.31(3.2)	64.33(3.2)	0.57	2.81	65.48(3.3)	65.44(4.2)	0.55	2.64	66.66(4.4)	66.69(3.8)	0.55	2.63
5000 Hz–AVR–10 ms	64.30(3.8)	64.33(3.5)	0.53	2.73	65.49(3.9)	65.46(4.8)	0.55	2.55	66.66(4.9)	66.67(3.7)	0.51	2.62

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Control Group's Non-Dominant Ankle when Acting as the Tilt Limb.

COMBINATION (Sample Rate–Analysis Method–Smoothing)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
1000 Hz–RMS–2 ms	49.65(4.3)	49.60(4.4)	0.88	2.35	50.86(4.6)	50.85(4.5)	0.81	2.30	53.34(4.9)	53.38(4.9)	0.80	2.40
1000 Hz–RMS–5 ms	49.67(4.2)	49.62(4.5)	0.83	2.42	50.87(4.8)	50.89(4.6)	0.80	2.36	53.35(4.7)	53.30(4.6)	0.77	2.47
1000 Hz–RMS–10 ms	49.69(4.1)	49.65(4.3)	0.78	2.55	50.86(3.6)	50.87(3.4)	0.73	2.40	53.39(4.6)	53.41(4.3)	0.76	2.58
1000 Hz–AVR–2 ms	49.67(4.2)	49.66(4.4)	0.78	2.63	50.87(3.1)	50.88(3.7)	0.76	2.64	53.38(4.8)	53.44(4.6)	0.75	2.67
1000 Hz–AVR–5 ms	49.69(3.7)	49.71(3.2)	0.75	2.72	50.88(4.6)	50.90(3.4)	0.74	2.58	53.39(4.7)	53.42(4.2)	0.73	2.50
1000 Hz–AVR–10 ms	49.71(3.2)	49.69(3.4)	0.74	2.70	50.89(4.5)	50.85(3.5)	0.73	2.57	53.41(4.2)	53.43(4.6)	0.72	2.68
2500 Hz–RMS–2 ms	49.70(3.5)	49.75(3.4)	0.80	2.55	50.88(3.6)	50.89(3.4)	0.78	2.46	53.42(3.5)	53.43(3.7)	0.75	2.50
2500 Hz–RMS–5 ms	49.73(3.2)	49.76(3.9)	0.80	2.59	50.89(3.9)	50.92(3.2)	0.75	2.45	53.45(3.7)	53.46(3.6)	0.80	2.59
2500 Hz–RMS–10 ms	49.71(3.5)	49.72(4.4)	0.74	2.61	50.90(3.6)	50.93(4.4)	0.74	2.52	53.46(3.5)	53.48(3.3)	0.72	2.68
2500 Hz–AVR–2 ms	49.73(3.2)	49.74(4.0)	0.73	2.70	50.91(3.2)	50.95(4.0)	0.75	2.56	53.49(3.3)	53.51(3.6)	0.70	2.64
2500 Hz–AVR–5 ms	49.75(3.1)	49.76(3.3)	0.70	2.73	50.92(4.5)	50.90(3.4)	0.74	2.55	53.48(3.3)	53.53(3.3)	0.70	2.75
2500 Hz–AVR–10 ms	49.76(3.2)	49.78(3.5)	0.74	2.63	50.95(4.0)	50.94(3.7)	0.74	2.63	53.49(3.5)	53.54(3.2)	0.67	2.70
5000 Hz–RMS–2 ms	49.75(3.5)	49.74(3.2)	0.72	2.70	50.93(3.6)	50.98(4.4)	0.73	2.76	53.51(3.4)	53.55(3.0)	0.67	2.62
5000 Hz–RMS–5 ms	49.77(3.4)	49.79(3.3)	0.73	2.75	50.95(3.7)	50.99(4.1)	0.72	2.75	53.52(3.9)	53.54(3.3)	0.65	2.67
5000 Hz–RMS–10 ms	49.78(3.9)	49.80(4.1)	0.70	2.77	50.97(4.7)	51.02(3.4)	0.70	2.74	53.53(3.4)	53.58(3.8)	0.64	2.66
5000 Hz–AVR–2 ms	49.77(3.4)	49.81(4.3)	0.69	2.76	50.99(4.0)	50.98(3.6)	0.66	2.73	53.55(3.6)	53.59(3.3)	0.64	2.65
5000 Hz–AVR–5 ms	49.79(3.4)	49.83(4.4)	0.65	2.81	51.04(4.6)	51.01(3.6)	0.64	2.84	53.58(4.5)	53.63(3.8)	0.60	2.63
5000 Hz–AVR–10 ms	49.81(3.2)	49.86(4.2)	0.62	2.53	51.07(4.3)	51.08(3.4)	0.64	2.75	53.57(4.3)	53.61(3.3)	0.58	2.62

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Control Group's Non-Dominant Ankle when Acting as the Support Limb.

COMBINATION (Sample Rate–Analysis Method–Smoothing)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
1000 Hz–RMS–2 ms	65.53(4.3)	65.57(3.3)	0.80	2.39	66.13(3.7)	66.17(2.8)	0.81	2.36	67.78(4.2)	67.79(3.3)	0.83	2.47
1000 Hz–RMS–5 ms	65.54(4.4)	65.59(3.5)	0.73	2.43	66.15(3.1)	66.20(2.2)	0.74	2.42	67.79(4.3)	67.80(3.4)	0.70	2.49
1000 Hz–RMS–10 ms	65.55(4.8)	65.56(3.5)	0.65	2.53	66.20(3.2)	66.21(2.7)	0.70	2.55	67.75(4.2)	67.77(3.5)	0.66	2.58
1000 Hz–AVR–2 ms	65.58(4.5)	65.60(3.4)	0.68	2.54	66.21(3.9)	66.22(2.2)	0.71	2.63	67.76(4.9)	67.74(3.3)	0.66	2.57
1000 Hz–AVR–5 ms	65.61(3.4)	65.64(2.8)	0.65	2.68	66.22(3.2)	66.21(3.0)	0.69	2.65	67.77(4.2)	67.79(4.5)	0.66	2.50
1000 Hz–AVR–10 ms	65.64(3.8)	65.63(2.3)	0.67	2.67	66.23(3.6)	66.25(3.1)	0.67	2.72	67.79(4.7)	67.82(4.3)	0.63	2.68
2500 Hz–RMS–2 ms	65.63(3.3)	65.67(2.7)	0.70	2.66	66.25(4.1)	66.24(4.5)	0.70	2.45	67.81(3.3)	67.83(4.0)	0.62	2.50
2500 Hz–RMS–5 ms	65.64(3.1)	65.65(2.3)	0.71	2.65	66.23(4.9)	66.22(4.1)	0.71	2.49	67.82(3.7)	67.83(4.4)	0.71	2.69
2500 Hz–RMS–10 ms	65.65(3.3)	65.69(3.9)	0.68	2.62	66.27(3.2)	66.24(3.2)	0.69	2.51	67.80(3.3)	67.84(4.5)	0.62	2.58
2500 Hz–AVR–2 ms	65.64(3.4)	65.70(3.3)	0.67	2.56	66.27(3.0)	66.27(3.1)	0.71	2.60	67.82(3.9)	67.83(4.4)	0.62	2.54
2500 Hz–AVR–5 ms	65.66(3.4)	65.73(3.0)	0.68	2.65	66.29(3.2)	66.29(3.2)	0.70	2.73	67.84(3.4)	67.79(4.6)	0.64	2.65
2500 Hz–AVR–10 ms	65.67(3.9)	65.73(3.3)	0.64	2.63	66.28(3.1)	66.30(3.8)	0.67	2.63	67.85(3.7)	67.83(4.5)	0.60	2.60
5000 Hz–RMS–2 ms	65.68(3.4)	65.72(3.5)	0.63	2.36	66.31(3.1)	66.32(3.1)	0.66	2.70	67.86(2.5)	67.82(3.7)	0.55	2.62
5000 Hz–RMS–5 ms	65.69(3.1)	65.75(3.3)	0.62	2.45	66.35(3.6)	66.30(3.2)	0.65	2.75	67.83(2.6)	67.86(3.4)	0.54	2.67
5000 Hz–RMS–10 ms	65.72(3.4)	65.73(3.8)	0.61	2.44	66.37(3.1)	66.29(4.2)	0.62	2.77	67.88(3.5)	67.84(3.3)	0.53	2.66
5000 Hz–AVR–2 ms	65.74(3.9)	65.73(3.3)	0.55	2.53	66.34(3.4)	66.33(4.6)	0.61	2.76	67.90(3.7)	67.85(3.7)	0.51	2.65
5000 Hz–AVR–5 ms	65.73(3.3)	65.75(3.5)	0.52	2.54	66.36(3.2)	66.37(4.2)	0.57	2.81	67.85(3.4)	67.88(3.4)	0.48	2.73
5000 Hz–AVR–10 ms	65.76(3.5)	65.76(3.3)	0.50	2.45	66.38(3.6)	66.40(4.6)	0.52	2.73	67.91(3.9)	67.90(3.7)	0.44	2.72

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Functional Ankle Instability Group's Unstable Ankle when Acting as the Tilt Limb.

COMBINATION (Sample Rate–Analysis Method–Smoothing)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
1000 Hz–RMS–2 ms	55.55(4.3)	55.54(3.5)	0.82	2.45	53.75(4.8)	53.73(4.6)	0.80	2.49	59.34(4.8)	59.39(4.6)	0.81	2.51
1000 Hz–RMS–5 ms	55.56(4.2)	55.55(3.3)	0.73	2.53	53.74(4.3)	53.74(4.2)	0.77	2.59	59.36(4.2)	59.40(4.4)	0.77	2.57
1000 Hz–RMS–10 ms	55.59(4.1)	55.56(3.2)	0.70	2.53	53.76(3.1)	53.77(3.7)	0.73	2.53	59.38(4.7)	59.41(4.3)	0.76	2.68
1000 Hz–AVR–2 ms	55.56(4.2)	55.55(4.3)	0.68	2.64	53.78(3.4)	53.80(3.2)	0.70	2.71	59.37(4.2)	59.39(4.5)	0.75	2.57
1000 Hz–AVR–5 ms	55.58(3.7)	55.59(3.5)	0.65	2.58	53.79(4.1)	53.81(3.7)	0.68	2.64	59.38(4.9)	59.42(4.0)	0.73	2.50
1000 Hz–AVR–10 ms	55.60(3.2)	55.62(3.3)	0.67	2.57	53.80(4.4)	53.80(3.3)	0.65	2.58	59.39(4.2)	59.42(4.2)	0.73	2.58
2500 Hz–RMS–2 ms	55.61(3.5)	55.59(3.9)	0.70	2.56	53.79(3.3)	53.82(3.2)	0.67	2.57	59.41(3.7)	59.45(3.3)	0.74	2.60
2500 Hz–RMS–5 ms	55.63(3.2)	55.60(3.3)	0.71	2.55	53.80(3.4)	53.85(3.3)	0.70	2.56	59.40(3.2)	59.46(3.2)	0.78	2.69
2500 Hz–RMS–10 ms	55.62(3.9)	55.63(4.0)	0.70	2.52	53.82(3.2)	53.86(4.3)	0.72	2.55	59.43(3.3)	59.48(4.0)	0.72	2.78
2500 Hz–AVR–2 ms	55.63(3.2)	55.65(4.2)	0.67	2.66	53.83(3.4)	53.84(4.2)	0.68	2.62	59.45(3.4)	59.49(4.2)	0.73	2.74
2500 Hz–AVR–5 ms	55.64(3.1)	55.65(3.7)	0.70	2.55	53.80(4.0)	53.82(3.7)	0.66	2.66	59.44(3.5)	59.50(4.3)	0.73	2.65
2500 Hz–AVR–10 ms	55.65(3.2)	55.66(3.2)	0.64	2.53	53.83(4.4)	53.85(3.2)	0.68	2.55	59.47(3.4)	59.47(4.2)	0.70	2.60
5000 Hz–RMS–2 ms	55.67(3.3)	55.69(3.3)	0.68	2.46	53.84(3.7)	53.86(4.1)	0.64	2.63	59.47(3.9)	59.49(3.0)	0.67	2.62
5000 Hz–RMS–5 ms	55.69(3.2)	55.70(3.2)	0.68	2.55	53.85(3.4)	53.87(4.2)	0.63	2.66	59.48(3.4)	59.51(3.3)	0.66	2.67
5000 Hz–RMS–10 ms	55.65(3.9)	55.69(4.1)	0.67	2.54	53.86(4.0)	53.89(3.6)	0.62	2.55	59.50(3.6)	59.53(3.8)	0.64	2.66
5000 Hz–AVR–2 ms	55.67(3.2)	55.69(4.2)	0.61	2.63	53.87(4.4)	53.90(3.2)	0.62	2.54	59.51(3.4)	59.55(3.3)	0.69	2.75
5000 Hz–AVR–5 ms	55.68(3.9)	55.71(4.0)	0.62	2.64	53.88(4.3)	53.93(3.4)	0.60	2.53	59.53(4.7)	59.56(4.8)	0.70	2.73
5000 Hz–AVR–10 ms	55.70(3.2)	55.74(4.2)	0.64	2.55	53.90(4.4)	53.97(3.2)	0.61	2.55	59.56(4.4)	59.58(4.3)	0.68	2.72

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Functional Ankle Instability Group's Unstable Ankle when Acting as the Support Limb.

COMBINATION (Sample Rate–Analysis Method–Smoothing)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
1000 Hz–RMS–2 ms	66.56(4.9)	66.57(3.7)	0.82	2.51	67.10(3.7)	67.09(2.8)	0.84	2.47	68.78(4.9)	68.75(3.9)	0.89	2.36
1000 Hz–RMS–5 ms	66.57(4.4)	66.59(3.4)	0.76	2.69	67.13(3.4)	67.15(2.3)	0.70	2.49	68.79(4.2)	68.73(3.6)	0.74	2.42
1000 Hz–RMS–10 ms	66.58(4.8)	66.59(4.4)	0.73	2.63	67.18(3.9)	67.16(2.7)	0.66	2.58	68.78(3.9)	68.87(3.8)	0.70	2.55
1000 Hz–AVR–2 ms	66.59(4.4)	66.61(4.5)	0.70	2.63	67.15(3.4)	67.18(2.3)	0.66	2.57	68.79(3.2)	68.81(3.5)	0.71	2.63
1000 Hz–AVR–5 ms	66.61(3.8)	66.62(2.8)	0.67	2.70	67.18(3.9)	67.21(3.0)	0.66	2.50	68.81(4.2)	68.82(4.5)	0.69	2.65
1000 Hz–AVR–10 ms	66.63(3.4)	66.65(2.4)	0.65	2.68	67.20(3.4)	67.22(3.3)	0.63	2.68	68.81(4.3)	68.83(4.4)	0.67	2.71
2500 Hz–RMS–2 ms	66.62(3.1)	66.66(2.6)	0.67	2.67	67.21(4.9)	67.23(4.5)	0.62	2.50	68.83(3.8)	68.80(4.8)	0.70	2.45
2500 Hz–RMS–5 ms	66.64(3.3)	66.65(2.4)	0.70	2.66	67.22(4.3)	67.26(4.3)	0.71	2.69	68.84(3.3)	68.84(4.3)	0.71	2.49
2500 Hz–RMS–10 ms	66.66(3.0)	66.68(3.6)	0.72	2.65	67.24(3.7)	67.23(3.1)	0.62	2.58	68.83(3.9)	68.84(4.5)	0.69	2.51
2500 Hz–AVR–2 ms	66.65(3.3)	66.69(3.4)	0.68	2.72	67.22(3.3)	67.24(3.3)	0.62	2.54	68.85(3.3)	68.85(4.3)	0.71	2.60
2500 Hz–AVR–5 ms	66.66(3.9)	66.69(3.0)	0.68	2.76	67.26(3.1)	67.28(3.8)	0.64	2.65	68.86(3.7)	68.88(4.6)	0.70	2.73
2500 Hz–AVR–10 ms	66.68(3.4)	66.71(3.4)	0.69	2.75	67.28(3.4)	67.35(3.3)	0.60	2.60	68.88(3.3)	68.90(4.5)	0.67	2.63
5000 Hz–RMS–2 ms	66.67(3.1)	66.70(3.6)	0.66	2.63	67.27(3.7)	67.32(3.0)	0.55	2.62	68.89(2.9)	68.91(2.8)	0.66	2.70
5000 Hz–RMS–5 ms	66.69(3.4)	66.72(3.3)	0.65	2.56	67.29(3.4)	67.31(3.3)	0.54	2.67	68.90(2.4)	68.88(2.5)	0.65	2.75
5000 Hz–RMS–10 ms	66.72(3.9)	66.74(3.8)	0.66	2.65	67.30(3.4)	67.36(4.7)	0.53	2.66	68.93(3.7)	68.90(3.9)	0.62	2.77
5000 Hz–AVR–2 ms	66.70(3.4)	66.74(3.3)	0.64	2.64	67.29(3.3)	67.27(4.3)	0.51	2.65	68.90(3.4)	68.91(3.5)	0.61	2.76
5000 Hz–AVR–5 ms	66.74(3.8)	66.75(3.6)	0.64	2.63	67.32(3.7)	67.31(4.8)	0.48	2.73	68.91(4.8)	68.94(3.7)	0.57	2.81
5000 Hz–AVR–10 ms	66.76(3.3)	66.73(3.3)	0.62	2.65	67.34(3.3)	67.30(4.3)	0.44	2.72	68.91(4.4)	68.96(3.5)	0.52	2.73

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Functional Ankle Instability Group's Stable Ankle when Acting as the Tilt Limb.

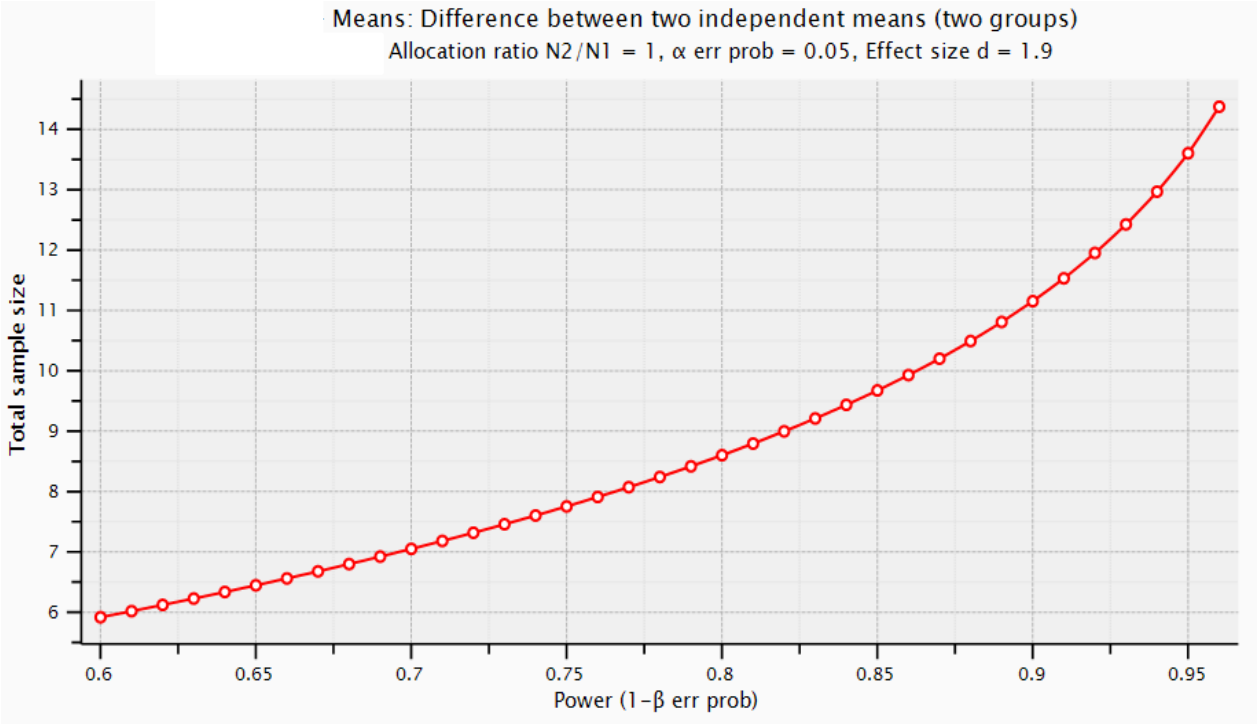
COMBINATION (Sample Rate–Analysis Method–Smoothing)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
1000 Hz–RMS–2 ms	54.64(4.7)	54.65(4.5)	0.81	2.39	52.80(4.2)	52.77(4.4)	0.80	2.30	58.38(4.3)	58.41(4.1)	0.81	2.30
1000 Hz–RMS–5 ms	54.68(4.4)	54.66(4.4)	0.73	2.43	52.82(4.8)	52.80(4.9)	0.73	2.33	58.39(4.8)	58.40(4.4)	0.80	2.36
1000 Hz–RMS–10 ms	54.66(4.3)	54.64(4.2)	0.65	2.53	52.83(3.1)	52.82(3.7)	0.68	2.43	58.37(4.3)	58.41(4.5)	0.73	2.40
1000 Hz–AVR–2 ms	54.65(4.1)	54.66(4.4)	0.68	2.54	52.82(3.8)	52.88(3.8)	0.70	2.44	58.39(4.7)	58.42(4.6)	0.76	2.64
1000 Hz–AVR–5 ms	54.66(3.3)	54.67(3.5)	0.65	2.68	52.80(4.1)	52.84(3.7)	0.68	2.58	58.40(4.3)	58.43(4.0)	0.74	2.58
1000 Hz–AVR–10 ms	54.68(3.7)	54.69(3.4)	0.67	2.67	52.85(4.7)	52.85(3.8)	0.65	2.57	58.42(4.9)	58.39(4.6)	0.73	2.57
2500 Hz–RMS–2 ms	54.69(3.3)	54.73(3.9)	0.70	2.66	52.87(3.3)	52.87(3.2)	0.71	2.56	58.43(3.2)	58.44(3.3)	0.78	2.46
2500 Hz–RMS–5 ms	54.71(3.5)	54.68(3.3)	0.71	2.65	52.86(3.7)	52.89(3.7)	0.71	2.55	58.40(3.7)	58.45(3.5)	0.75	2.45
2500 Hz–RMS–10 ms	54.69(3.4)	54.73(4.0)	0.68	2.62	52.88(3.2)	52.92(4.3)	0.70	2.42	58.44(3.3)	58.47(4.0)	0.74	2.52
2500 Hz–AVR–2 ms	54.68(3.9)	54.74(4.3)	0.67	2.56	52.89(3.8)	52.94(4.6)	0.67	2.56	58.45(3.3)	58.44(4.5)	0.75	2.56
2500 Hz–AVR–5 ms	54.71(3.4)	54.76(3.7)	0.68	2.65	52.90(4.0)	52.95(3.5)	0.68	2.55	58.46(3.4)	58.49(3.3)	0.74	2.55
2500 Hz–AVR–10 ms	54.72(3.1)	54.75(3.4)	0.64	2.63	52.87(4.8)	52.98(3.6)	0.65	2.43	58.47(3.5)	58.49(3.4)	0.74	2.63
5000 Hz–RMS–2 ms	54.74(2.4)	54.77(3.0)	0.63	2.36	52.91(3.6)	52.95(4.2)	0.63	2.56	58.50(3.5)	58.47(3.0)	0.73	2.76
5000 Hz–RMS–5 ms	54.72(2.3)	54.79(3.4)	0.62	2.45	52.92(3.2)	52.97(4.7)	0.63	2.55	58.49(3.9)	58.52(3.3)	0.72	2.75
5000 Hz–RMS–10 ms	54.75(3.3)	54.80(4.1)	0.61	2.44	52.93(4.6)	52.99(3.6)	0.61	2.54	58.52(3.6)	58.54(3.8)	0.70	2.74
5000 Hz–AVR–2 ms	54.77(3.4)	54.83(4.4)	0.55	2.53	52.94(4.0)	52.95(3.8)	0.59	2.53	58.53(3.7)	58.57(3.3)	0.66	2.73
5000 Hz–AVR–5 ms	54.79(3.9)	54.82(4.0)	0.52	2.54	52.94(4.5)	52.94(4.2)	0.55	2.64	58.55(4.8)	58.59(4.8)	0.64	2.84
5000 Hz–AVR–10 ms	54.78(3.4)	54.85(4.3)	0.50	2.45	52.97(4.8)	52.96(4.8)	0.55	2.55	58.59(4.2)	58.62(4.2)	0.64	2.75

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Functional Ankle Instability Group's Stable Ankle when Acting as the Support Limb.

COMBINATION (Sample Rate–Analysis Method–Smoothing)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
1000 Hz–RMS–2 ms	66.11(4.0)	66.16(3.4)	0.85	2.45	66.84(3.3)	66.83(2.4)	0.80	2.30	68.28(4.4)	68.24(3.0)	0.82	2.45
1000 Hz–RMS–5 ms	66.14(4.2)	66.13(3.0)	0.71	2.47	66.85(3.7)	66.84(2.9)	0.73	2.33	68.27(4.3)	68.26(3.5)	0.73	2.53
1000 Hz–RMS–10 ms	66.16(4.8)	66.20(4.3)	0.67	2.58	66.87(3.4)	66.86(2.5)	0.68	2.43	68.29(3.9)	68.38(3.2)	0.70	2.53
1000 Hz–AVR–2 ms	66.13(4.2)	66.23(4.4)	0.67	2.47	66.86(3.6)	66.83(2.7)	0.70	2.44	68.30(3.3)	68.28(3.5)	0.68	2.64
1000 Hz–AVR–5 ms	66.20(3.3)	66.25(3.3)	0.65	2.60	66.88(3.5)	66.89(3.5)	0.68	2.58	68.31(4.2)	68.33(4.5)	0.65	2.58
1000 Hz–AVR–10 ms	66.22(3.2)	66.24(3.8)	0.63	2.58	66.87(3.3)	66.90(3.0)	0.65	2.57	68.32(4.5)	68.34(4.3)	0.67	2.57
2500 Hz–RMS–2 ms	66.25(3.5)	66.27(3.2)	0.68	2.40	66.89(3.9)	66.86(4.6)	0.71	2.56	68.33(3.8)	68.40(4.8)	0.70	2.56
2500 Hz–RMS–5 ms	66.24(3.3)	66.28(3.6)	0.74	2.49	66.90(3.3)	66.88(4.5)	0.71	2.55	68.35(3.5)	68.35(4.3)	0.71	2.55
2500 Hz–RMS–10 ms	66.26(3.0)	66.29(3.2)	0.60	2.58	66.92(3.7)	66.87(3.6)	0.70	2.42	68.32(3.9)	68.36(4.5)	0.70	2.52
2500 Hz–AVR–2 ms	66.26(3.3)	66.30(3.6)	0.63	2.44	66.90(3.3)	66.89(3.1)	0.67	2.56	68.33(3.4)	68.37(4.3)	0.67	2.66
2500 Hz–AVR–5 ms	66.27(3.9)	66.29(3.2)	0.62	2.55	66.91(3.1)	66.90(3.4)	0.68	2.55	68.36(3.7)	68.35(4.6)	0.70	2.55
2500 Hz–AVR–10 ms	66.25(3.2)	66.31(3.3)	0.59	2.50	66.92(3.3)	66.93(3.8)	0.65	2.43	68.32(3.3)	68.33(4.2)	0.64	2.53
5000 Hz–RMS–2 ms	66.27(3.1)	66.30(3.4)	0.60	2.52	66.93(3.0)	66.91(3.4)	0.63	2.56	68.37(2.9)	68.40(2.8)	0.68	2.46
5000 Hz–RMS–5 ms	66.29(3.2)	66.33(3.5)	0.58	2.57	66.91(3.3)	66.95(3.2)	0.63	2.55	68.38(2.3)	68.44(2.2)	0.68	2.55
5000 Hz–RMS–10 ms	66.30(3.9)	66.32(3.4)	0.53	2.56	66.93(3.4)	66.94(4.4)	0.61	2.54	68.40(3.7)	68.42(3.9)	0.67	2.54
5000 Hz–AVR–2 ms	66.31(3.3)	66.33(3.3)	0.52	2.55	66.94(3.3)	66.95(4.7)	0.59	2.53	68.39(3.2)	68.41(3.3)	0.61	2.63
5000 Hz–AVR–5 ms	66.33(3.8)	66.34(3.6)	0.55	2.63	66.96(3.7)	66.92(4.3)	0.55	2.64	68.41(4.3)	68.42(3.1)	0.62	2.64
5000 Hz–AVR–10 ms	66.30(3.2)	66.36(3.3)	0.51	2.62	66.94(3.3)	66.94(4.2)	0.55	2.55	68.43(4.2)	68.46(3.2)	0.64	2.55

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement



The Effect Size Generator (Version 2.3.0, Australia) software was used to calculate Cohen's d effect size. As three muscles were tested in Pilot Study Two, the muscle and condition that generated the smallest effect size was used for the calculation of power. The above plot represents the tibialis anterior muscle, when acting as a tilt limb. An effect size of d = 1.9 was calculated from the healthy groups NDA's muscle latency (mean = 49.75 ms, SD = 2.3) and the FAI groups UA's muscle latency (mean = 53.70, SD = 1.9). The plot produced by G*Power (Version 3.1.5, Germany) (Faul et al., 2007) indicated that a total sample size of 14 subjects would be needed to achieve a power of >95% between two independent groups. Therefore, in Study One and Study Three of this thesis a minimum of 7 subjects will be required in each group.

Muscle Latencies for the Control Group's Dominant Ankle when Acting as the Tilt Limb.

COMBINATION (Baseline–Deviation–Number of Samples)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
50 ms–1 SD–50 ms	48.85(4.3)	48.90(4.5)	0.85	2.34	49.85(4.8)	50.10(4.6)	0.80	2.66	52.34(4.8)	52.41(4.6)	0.76	2.86
50 ms–1 SD–100 ms	48.85(4.3)	48.90(4.5)	0.85	2.34	49.85(4.8)	50.10(4.6)	0.80	2.66	52.34(4.8)	52.41(4.6)	0.76	2.86
100 ms–1 SD–50 ms	48.97(4.1)	49.10(4.2)	0.81	2.53	48.86(.1)	48.99(3.7)	0.83	2.30	52.55(4.7)	52.76(4.3)	0.74	2.89
100 ms–1 SD–100 ms	48.97(4.1)	49.10(4.2)	0.81	2.53	48.86(3.1)	48.99(3.7)	0.83	2.30	52.55(4.7)	52.76(4.3)	0.74	2.89
500 ms–1 SD–50 ms	49.05(3.7)	49.22(3.4)	0.77	2.62	49.10(4.1)	49.18(3.7)	0.79	2.58	52.63(4.7)	52.87(4.0)	0.68	3.02
500 ms–1 SD–100 ms	49.05(3.7)	49.22(3.4)	0.77	2.62	49.10(4.1)	49.18(3.7)	0.79	2.58	52.63(4.7)	52.87(4.0)	0.68	3.02
50 ms–2 SD–50 ms	49.05(3.7)	49.15(3.9)	0.83	2.35	49.87(3.9)	49.93(3.2)	0.80	2.45	52.65(3.7)	52.63(3.7)	0.71	2.86
50 ms–2 SD–100 ms	49.05(3.7)	49.15(3.9)	0.83	2.35	49.87(3.9)	49.93(3.2)	0.80	2.45	52.65(3.7)	52.63(3.7)	0.71	2.86
100 ms–2 SD–50 ms	49.11(3.9)	49.23(4.0)	0.82	2.41	49.10(3.2)	49.21(4.0)	0.85	2.33	52.63(3.9)	52.67(4.0)	0.74	2.96
100 ms–2 SD–100 ms	49.11(3.9)	49.23(4.0)	0.82	2.41	49.10(3.2)	49.21(4.0)	0.85	2.33	52.63(3.9)	52.67(4.0)	0.74	2.96
500 ms–2 SD–50 ms	49.35(3.1)	49.41(3.7)	0.80	2.43	49.31(4.0)	49.37(3.7)	0.79	2.45	52.67(3.5)	52.70(4.3)	0.64	2.76
500 ms–2 SD–100 ms	49.35(3.1)	49.41(3.7)	0.80	2.43	49.31(4.0)	49.37(3.7)	0.79	2.45	52.67(3.5)	52.70(4.3)	0.64	2.76
50 ms–3 SD–50 ms	49.51(3.5)	49.60(3.3)	0.95	2.20	49.75(3.7)	49.69(4.1)	0.93	2.22	52.78(3.9)	52.81(3.0)	0.90	2.55
50 ms–3 SD–100 ms	49.51(3.5)	49.60(3.3)	0.95	2.20	49.75(3.7)	49.69(4.1)	0.93	2.22	52.78(3.9)	52.81(3.0)	0.90	2.55
100 ms–3 SD–50 ms	49.48(3.9)	49.39(4.1)	0.90	2.37	49.40(4.0)	49.33(3.6)	0.87	2.45	52.72(3.6)	52.85(3.8)	0.82	2.76
100 ms–3 SD–100 ms	49.48(3.9)	49.39(4.1)	0.90	2.37	49.40(4.0)	49.33(3.6)	0.87	2.45	52.72(3.6)	52.85(3.8)	0.82	2.76
500 ms–3 SD–50 ms	49.53(3.9)	49.49(4.0)	0.88	2.41	49.59(4.3)	49.62(4.4)	0.85	2.47	52.77(4.7)	52.68(3.8)	0.78	2.79
500 ms–3 SD–100 ms	49.53(3.9)	49.49(4.0)	0.88	2.41	49.59(4.3)	49.62(4.4)	0.85	2.47	52.77(4.7)	52.68(3.8)	0.78	2.79

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Control Group's Dominant Ankle when Acting as the Support Limb.

COMBINATION (Baseline–Deviation–Number of Samples)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
50 ms–1 SD–50 ms	64.56(4.9)	64.76(3.7)	0.74	2.71	65.10(3.7)	65.43(2.8)	0.72	2.66	66.78(4.9)	66.73(3.9)	0.71	2.71
50 ms–1 SD–100 ms	64.56(4.9)	64.76(3.7)	0.74	2.71	65.10(3.7)	65.43(2.8)	0.72	2.66	66.78(4.9)	66.73(3.9)	0.71	2.71
100 ms–1 SD–50 ms	64.58(4.8)	64.68(3.4)	0.70	2.78	65.21(3.9)	65.16(2.7)	0.72	2.78	66.75(3.9)	66.79(3.8)	0.69	2.99
100 ms–1 SD–100 ms	64.58(4.8)	64.68(3.4)	0.70	2.78	65.21(3.9)	65.16(2.7)	0.72	2.78	66.75(3.9)	66.79(3.8)	0.69	2.99
500 ms–1 SD–50 ms	64.61(3.8)	64.65(3.8)	0.66	3.05	65.34(3.9)	65.45(3.0)	0.69	2.94	66.71(4.2)	66.82(4.5)	0.69	2.87
500 ms–1 SD–100 ms	64.61(3.8)	64.65(3.8)	0.66	3.05	65.34(3.9)	65.45(3.0)	0.69	2.94	66.71(4.2)	66.82(4.5)	0.69	2.87
50 ms–2 SD–50 ms	64.62(3.1)	64.65(3.7)	0.76	2.76	65.38(4.9)	65.46(4.5)	0.78	2.65	66.75(3.7)	66.80(4.0)	0.80	2.70
50 ms–2 SD–100 ms	64.62(3.1)	64.65(3.7)	0.76	2.76	65.38(4.9)	65.46(4.5)	0.78	2.65	66.75(3.7)	66.80(4.0)	0.80	2.70
100 ms–2 SD–50 ms	64.64(3.0)	64.68(3.9)	0.69	2.98	65.39(3.0)	65.48(3.1)	0.70	2.78	66.72(3.9)	66.81(4.5)	0.69	2.96
100 ms–2 SD–100 ms	64.64(3.0)	64.68(3.9)	0.69	2.98	65.39(3.0)	65.48(3.1)	0.70	2.78	66.72(3.9)	66.81(4.5)	0.69	2.96
500 ms–2 SD–50 ms	64.66(3.9)	64.69(3.0)	0.61	2.87	65.41(3.1)	66.43(3.8)	0.64	2.88	66.73(3.7)	66.79(4.6)	0.62	3.12
500 ms–2 SD–100 ms	64.66(3.9)	64.69(3.0)	0.61	2.87	65.41(3.1)	66.43(3.8)	0.64	2.88	66.73(3.7)	66.79(4.6)	0.62	3.12
50 ms–3 SD–50 ms	64.68(3.1)	64.70(3.2)	0.85	2.35	65.45(3.7)	65.49(3.0)	0.82	2.40	66.75(2.9)	66.80(2.7)	0.85	2.65
50 ms–3 SD–100 ms	64.68(3.1)	64.70(3.2)	0.85	2.35	65.45(3.7)	65.49(3.0)	0.82	2.40	66.75(2.9)	66.80(2.7)	0.85	2.65
100 ms–3 SD–50 ms	64.69(3.9)	64.72(3.8)	0.73	2.65	65.47(3.4)	65.44(3.8)	0.75	2.56	66.77(3.7)	66.79(3.9)	0.74	2.78
100 ms–3 SD–100 ms	64.69(3.9)	64.72(3.8)	0.73	2.65	65.47(3.4)	65.44(3.8)	0.75	2.56	66.77(3.7)	66.79(3.9)	0.74	2.78
500 ms–3 SD–50 ms	64.70(3.8)	64.73(3.5)	0.68	2.86	65.49(3.9)	65.48(3.8)	0.68	2.65	66.78(4.9)	66.82(3.7)	0.69	2.98
500 ms–3 SD–100 ms	64.70(3.8)	64.73(3.5)	0.68	2.86	65.49(3.9)	65.48(3.8)	0.68	2.65	66.78(4.9)	66.82(3.7)	0.69	2.98

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Control Group's Non-Dominant Ankle when Acting as the Tilt Limb.

COMBINATION (Baseline–Deviation–Number of Samples)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
50 ms–1 SD–50 ms	49.75(4.3)	47.80(4.5)	0.85	2.40	50.85(4.8)	51.10(4.6)	0.78	2.56	53.34(4.9)	53.45(3.9)	0.79	2.76
50 ms–1 SD–100 ms	49.75(4.3)	47.80(4.5)	0.85	2.40	50.85(4.8)	51.10(4.6)	0.78	2.56	53.34(4.9)	53.45(3.9)	0.79	2.76
100 ms–1 SD–50 ms	49.78(4.1)	48.15(4.4)	0.82	2.43	49.76(3.1)	49.98(3.7)	0.81	2.40	53.55(4.7)	53.76(4.3)	0.72	2.79
100 ms–1 SD–100 ms	49.78(4.1)	48.15(4.4)	0.82	2.43	49.76(3.1)	49.98(3.7)	0.81	2.40	53.55(4.7)	53.76(4.3)	0.72	2.79
500 ms–1 SD–50 ms	50.09(3.7)	50.24(3.4)	0.74	2.52	50.15(4.5)	50.18(3.5)	0.75	2.55	53.63(4.2)	53.87(4.2)	0.66	2.87
500 ms–1 SD–100 ms	50.09(3.7)	50.24(3.4)	0.74	2.52	50.15(4.5)	50.18(3.5)	0.75	2.55	53.63(4.2)	53.87(4.2)	0.66	2.87
50 ms–2 SD–50 ms	50.05(3.5)	50.15(3.9)	0.82	2.35	50.87(3.9)	50.93(3.2)	0.77	2.55	53.65(3.7)	53.63(3.7)	0.71	2.76
50 ms–2 SD–100 ms	50.05(3.5)	50.15(3.9)	0.82	2.35	50.87(3.9)	50.93(3.2)	0.77	2.55	53.65(3.7)	53.63(3.7)	0.71	2.76
100 ms–2 SD–50 ms	50.11(3.5)	50.25(4.0)	0.74	2.41	50.10(3.2)	50.21(4.0)	0.71	2.53	53.63(3.3)	53.64(4.3)	0.69	2.73
100 ms–2 SD–100 ms	50.11(3.5)	50.25(4.0)	0.74	2.41	50.10(3.2)	50.21(4.0)	0.71	2.53	53.63(3.3)	53.64(4.3)	0.69	2.73
500 ms–2 SD–50 ms	50.35(3.1)	50.41(3.5)	0.65	2.43	50.31(4.0)	50.37(3.7)	0.69	2.55	53.67(3.5)	53.70(4.3)	0.62	2.76
500 ms–2 SD–100 ms	50.35(3.1)	50.41(3.5)	0.65	2.43	50.31(4.0)	50.37(3.7)	0.69	2.55	53.67(3.5)	53.70(4.3)	0.62	2.76
50 ms–3 SD–50 ms	50.55(3.5)	50.61(3.3)	0.88	2.25	50.75(3.7)	50.79(4.1)	0.90	2.21	53.78(3.9)	53.81(3.0)	0.88	2.54
50 ms–3 SD–100 ms	50.55(3.5)	50.61(3.3)	0.88	2.25	50.75(3.7)	50.79(4.1)	0.90	2.21	53.78(3.9)	53.81(3.0)	0.88	2.54
100 ms–3 SD–50 ms	50.48(3.9)	50.39(4.1)	0.80	2.37	50.40(4.0)	50.33(3.6)	0.79	2.45	53.72(3.6)	53.85(3.8)	0.75	2.75
100 ms–3 SD–100 ms	50.48(3.9)	50.39(4.1)	0.80	2.37	50.40(4.0)	50.33(3.6)	0.79	2.45	53.72(3.6)	53.85(3.8)	0.75	2.75
500 ms–3 SD–50 ms	50.53(3.4)	50.49(4.4)	0.76	2.43	50.57(4.3)	50.62(4.4)	0.73	2.43	53.77(3.3)	53.63(3.8)	0.71	2.71
500 ms–3 SD–100 ms	50.53(3.4)	50.49(4.4)	0.76	2.43	50.57(4.3)	50.62(4.4)	0.73	2.43	53.77(3.3)	53.63(3.8)	0.71	2.71

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Control Group's Non-Dominant Ankle when Acting as the Support Limb.

COMBINATION (Baseline–Deviation–Number of Samples)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
50 ms–1 SD–50 ms	65.53(4.3)	65.73(3.3)	0.73	2.73	66.13(3.7)	66.33(3.8)	0.70	2.63	67.78(4.3)	67.73(3.3)	0.64	2.51
50 ms–1 SD–100 ms	65.53(4.3)	65.73(3.3)	0.73	2.73	66.13(3.7)	66.33(3.8)	0.70	2.63	67.78(4.3)	67.73(3.3)	0.64	2.51
100 ms–1 SD–50 ms	65.58(4.8)	65.68(4.4)	0.71	2.75	66.21(3.9)	66.16(3.7)	0.70	2.76	67.75(3.9)	67.79(3.5)	0.67	2.79
100 ms–1 SD–100 ms	65.58(4.8)	65.68(4.4)	0.71	2.75	66.21(3.9)	66.16(3.7)	0.70	2.76	67.75(3.9)	67.79(3.5)	0.67	2.79
500 ms–1 SD–50 ms	65.61(3.8)	65.65(3.8)	0.65	2.87	66.34(3.6)	66.45(3.0)	0.66	2.94	67.71(4.7)	67.72(4.5)	0.63	2.88
500 ms–1 SD–100 ms	65.61(3.8)	65.65(3.8)	0.65	2.87	66.34(3.6)	66.45(3.0)	0.66	2.94	67.71(4.7)	67.72(4.5)	0.63	2.88
50 ms–2 SD–50 ms	65.62(3.1)	65.65(3.7)	0.76	2.76	66.38(3.9)	66.46(4.5)	0.77	2.65	67.75(3.7)	67.80(4.0)	0.77	2.70
50 ms–2 SD–100 ms	65.62(3.1)	65.65(3.7)	0.76	2.76	66.38(3.9)	66.46(4.5)	0.77	2.65	67.75(3.7)	67.80(4.0)	0.77	2.70
100 ms–2 SD–50 ms	65.64(3.0)	65.68(3.9)	0.69	2.98	66.39(3.0)	66.48(3.1)	0.70	2.78	67.72(3.9)	67.81(4.5)	0.67	2.96
100 ms–2 SD–100 ms	65.64(3.0)	65.68(3.9)	0.69	2.98	66.39(3.0)	66.48(3.1)	0.70	2.78	67.72(3.9)	67.81(4.5)	0.67	2.96
500 ms–2 SD–50 ms	65.66(3.9)	65.69(3.0)	0.61	2.77	66.41(3.1)	66.43(3.8)	0.64	2.88	67.73(3.7)	67.79(4.6)	0.62	3.04
500 ms–2 SD–100 ms	65.66(3.9)	65.69(3.0)	0.61	2.77	66.41(3.1)	66.43(3.8)	0.64	2.88	67.73(3.7)	67.79(4.6)	0.62	3.04
50 ms–3 SD–50 ms	65.68(3.1)	65.70(3.5)	0.86	2.36	66.45(3.6)	66.46(3.2)	0.88	2.47	67.75(2.6)	67.80(2.7)	0.87	2.65
50 ms–3 SD–100 ms	65.68(3.1)	65.70(3.5)	0.86	2.36	66.45(3.6)	66.46(3.2)	0.88	2.47	67.75(2.6)	67.80(2.7)	0.87	2.65
100 ms–3 SD–50 ms	65.69(3.9)	65.72(3.8)	0.75	2.63	66.47(3.4)	66.46(4.6)	0.76	2.53	67.75(3.7)	67.79(3.7)	0.75	2.75
100 ms–3 SD–100 ms	65.69(3.9)	65.72(3.8)	0.75	2.63	66.47(3.4)	66.46(4.6)	0.76	2.53	67.75(3.7)	67.79(3.7)	0.75	2.75
500 ms–3 SD–50 ms	65.70(3.5)	65.75(3.5)	0.69	2.83	66.49(3.6)	66.48(4.6)	0.68	2.66	67.78(3.9)	67.77(3.7)	0.68	2.95
500 ms–3 SD–100 ms	65.70(3.5)	65.75(3.5)	0.69	2.83	66.49(3.6)	66.48(4.6)	0.68	2.66	67.78(3.9)	67.77(3.7)	0.68	2.95

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Functional Ankle Instability Group's Unstable Ankle when Acting as the Tilt Limb.

COMBINATION (Baseline–Deviation–Number of Samples)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
50 ms–1 SD–50 ms	55.85(4.3)	55.90(4.5)	0.79	2.43	53.85(4.8)	53.10(4.6)	0.80	2.45	59.34(4.8)	59.41(4.6)	0.76	2.56
50 ms–1 SD–100 ms	55.85(4.3)	55.90(4.5)	0.79	2.43	53.85(4.8)	53.10(4.6)	0.80	2.45	59.34(4.8)	59.41(4.6)	0.76	2.56
100 ms–1 SD–50 ms	55.97(4.1)	55.10(4.2)	0.77	2.53	53.86(3.1)	53.99(3.7)	0.73	2.34	59.55(4.7)	59.79(4.3)	0.74	2.86
100 ms–1 SD–100 ms	55.97(4.1)	55.10(4.2)	0.77	2.53	53.86(3.1)	53.99(3.7)	0.73	2.34	59.55(4.7)	59.79(4.3)	0.74	2.86
500 ms–1 SD–50 ms	55.05(3.7)	55.25(3.5)	0.75	2.65	53.10(4.1)	53.13(3.7)	0.72	2.52	59.63(4.9)	59.80(4.0)	0.68	2.87
500 ms–1 SD–100 ms	55.05(3.7)	55.25(3.5)	0.75	2.65	53.10(4.1)	53.13(3.7)	0.72	2.52	59.63(4.9)	59.80(4.0)	0.68	2.87
50 ms–2 SD–50 ms	55.05(3.5)	55.15(3.9)	0.80	2.35	53.87(3.3)	53.93(3.2)	0.78	2.45	59.65(3.7)	59.63(3.3)	0.75	2.86
50 ms–2 SD–100 ms	55.05(3.5)	55.15(3.9)	0.80	2.35	53.87(3.3)	53.93(3.2)	0.78	2.45	59.65(3.7)	59.63(3.3)	0.75	2.86
100 ms–2 SD–50 ms	55.11(3.9)	55.23(3.0)	0.77	2.41	53.10(3.2)	53.21(4.3)	0.73	2.35	59.63(3.3)	59.64(4.0)	0.67	2.96
100 ms–2 SD–100 ms	55.11(3.9)	55.23(3.0)	0.77	2.41	53.10(3.2)	53.21(4.3)	0.73	2.35	59.63(3.3)	59.64(4.0)	0.67	2.96
500 ms–2 SD–50 ms	55.35(3.1)	55.41(3.7)	0.72	2.41	53.31(4.0)	53.37(3.7)	0.68	2.49	59.67(3.5)	59.70(4.3)	0.62	2.76
500 ms–2 SD–100 ms	55.35(3.1)	55.41(3.7)	0.72	2.41	53.31(4.0)	53.37(3.7)	0.68	2.49	59.67(3.5)	59.70(4.3)	0.62	2.76
50 ms–3 SD–50 ms	55.50(3.3)	55.54(3.3)	0.90	2.35	53.70(3.7)	53.67(4.1)	0.88	2.42	59.79(3.9)	59.83(3.0)	0.88	2.52
50 ms–3 SD–100 ms	55.50(3.3)	55.54(3.3)	0.90	2.35	53.70(3.7)	53.67(4.1)	0.88	2.42	59.79(3.9)	59.83(3.0)	0.88	2.52
100 ms–3 SD–50 ms	55.48(3.9)	55.39(4.1)	0.81	2.47	53.40(4.0)	53.33(3.6)	0.78	2.49	59.72(3.6)	59.85(3.8)	0.80	2.76
100 ms–3 SD–100 ms	55.48(3.9)	55.39(4.1)	0.81	2.47	53.40(4.0)	53.33(3.6)	0.78	2.49	59.72(3.6)	59.85(3.8)	0.80	2.76
500 ms–3 SD–50 ms	55.53(3.9)	55.49(4.0)	0.75	2.54	53.59(4.3)	53.62(4.4)	0.71	2.65	59.77(4.7)	59.68(4.8)	0.74	2.70
500 ms–3 SD–100 ms	55.53(3.9)	55.49(4.0)	0.75	2.54	53.59(4.3)	53.62(4.4)	0.71	2.65	59.77(4.7)	59.68(4.8)	0.74	2.70

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Functional Ankle Instability Group's Unstable Ankle when Acting as the Support Limb.

COMBINATION (Baseline–Deviation–Number of Samples)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
50 ms–1 SD–50 ms	66.56(4.9)	66.76(3.7)	0.73	2.73	67.10(3.7)	67.43(2.8)	0.71	2.61	68.78(4.9)	68.73(3.9)	0.69	2.81
50 ms–1 SD–100 ms	66.56(4.9)	66.76(3.7)	0.73	2.73	67.10(3.7)	67.43(2.8)	0.71	2.61	68.78(4.9)	68.73(3.9)	0.69	2.81
100 ms–1 SD–50 ms	66.58(4.8)	66.68(4.4)	0.70	2.78	67.21(3.9)	67.16(2.7)	0.70	2.77	68.78(3.9)	68.89(3.8)	0.67	2.90
100 ms–1 SD–100 ms	66.58(4.8)	66.68(4.4)	0.70	2.78	67.21(3.9)	67.16(2.7)	0.70	2.77	68.78(3.9)	68.89(3.8)	0.67	2.90
500 ms–1 SD–50 ms	66.61(3.8)	66.65(2.8)	0.66	3.15	67.34(3.9)	67.45(3.0)	0.67	2.97	68.71(4.2)	68.82(4.5)	0.68	2.88
500 ms–1 SD–100 ms	66.61(3.8)	66.65(2.8)	0.66	3.15	67.34(3.9)	67.45(3.0)	0.67	2.97	68.71(4.2)	68.82(4.5)	0.68	2.88
50 ms–2 SD–50 ms	66.62(3.1)	66.65(2.6)	0.77	2.74	67.38(4.9)	67.46(4.5)	0.75	2.63	68.75(3.8)	68.80(4.8)	0.73	2.73
50 ms–2 SD–100 ms	66.62(3.1)	66.65(2.6)	0.77	2.74	67.38(4.9)	67.46(4.5)	0.75	2.63	68.75(3.8)	68.80(4.8)	0.73	2.73
100 ms–2 SD–50 ms	66.66(3.0)	66.68(3.6)	0.70	2.95	67.39(3.7)	67.47(3.1)	0.69	2.76	68.72(3.9)	68.81(4.5)	0.67	2.98
100 ms–2 SD–100 ms	66.66(3.0)	66.68(3.6)	0.70	2.95	67.39(3.7)	67.47(3.1)	0.69	2.76	68.72(3.9)	68.81(4.5)	0.67	2.98
500 ms–2 SD–50 ms	66.66(3.9)	66.69(3.0)	0.63	2.83	67.41(3.1)	67.43(3.8)	0.65	2.85	68.73(3.7)	68.78(4.6)	0.63	2.90
500 ms–2 SD–100 ms	66.66(3.9)	66.69(3.0)	0.63	2.83	67.41(3.1)	67.43(3.8)	0.65	2.85	68.73(3.7)	68.78(4.6)	0.63	2.90
50 ms–3 SD–50 ms	66.67(3.1)	66.70(3.6)	0.86	2.34	67.45(3.7)	67.47(3.0)	0.85	2.45	68.78(2.9)	68.80(2.8)	0.88	2.68
50 ms–3 SD–100 ms	66.67(3.1)	66.70(3.6)	0.86	2.34	67.45(3.7)	67.47(3.0)	0.85	2.45	68.78(2.9)	68.80(2.8)	0.88	2.68
100 ms–3 SD–50 ms	66.69(3.9)	66.72(3.8)	0.74	2.63	67.47(3.4)	67.44(4.7)	0.77	2.57	68.78(3.7)	68.82(3.9)	0.72	2.70
100 ms–3 SD–100 ms	66.69(3.9)	66.72(3.8)	0.74	2.63	67.47(3.4)	67.44(4.7)	0.77	2.57	68.78(3.7)	68.82(3.9)	0.72	2.70
500 ms–3 SD–50 ms	66.76(3.8)	66.73(3.6)	0.69	2.86	67.49(3.7)	67.50(4.8)	0.68	2.67	68.78(4.8)	68.82(3.7)	0.68	2.91
500 ms–3 SD–100 ms	66.76(3.8)	66.73(3.6)	0.69	2.86	67.49(3.7)	67.50(4.8)	0.68	2.67	68.78(4.8)	68.82(3.7)	0.68	2.91

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Functional Ankle Instability Group's Stable Ankle when Acting as the Tilt Limb.

COMBINATION (Baseline–Deviation–Number of Samples)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
50 ms–1 SD–50 ms	54.84(4.4)	54.94(4.5)	0.74	2.44	52.82(4.2)	52.70(4.6)	0.70	2.40	58.38(4.8)	58.41(4.6)	0.76	2.60
50 ms–1 SD–100 ms	54.84(4.4)	54.94(4.5)	0.74	2.44	52.82(4.2)	52.70(4.6)	0.70	2.40	58.38(4.8)	58.41(4.6)	0.76	2.60
100 ms–1 SD–50 ms	54.96(4.1)	54.90(4.2)	0.71	2.53	52.82(3.1)	52.92(3.7)	0.68	2.30	58.55(4.7)	58.79(4.8)	0.73	2.82
100 ms–1 SD–100 ms	54.96(4.1)	54.90(4.2)	0.71	2.53	52.82(3.1)	52.92(3.7)	0.68	2.30	58.55(4.7)	58.79(4.8)	0.73	2.82
500 ms–1 SD–50 ms	54.91(3.7)	54.95(3.5)	0.68	2.60	52.80(4.1)	52.93(3.7)	0.65	2.52	58.60(4.9)	58.82(4.0)	0.64	2.82
500 ms–1 SD–100 ms	54.91(3.7)	54.95(3.5)	0.68	2.60	52.80(4.1)	52.93(3.7)	0.65	2.52	58.60(4.9)	58.82(4.0)	0.64	2.82
50 ms–2 SD–50 ms	54.92(3.5)	54.95(3.9)	0.75	2.35	52.87(3.3)	52.93(3.2)	0.77	2.47	58.65(3.7)	58.63(3.3)	0.75	2.83
50 ms–2 SD–100 ms	54.92(3.5)	54.95(3.9)	0.75	2.35	52.87(3.3)	52.93(3.2)	0.77	2.47	58.65(3.7)	58.63(3.3)	0.75	2.83
100 ms–2 SD–50 ms	54.91(3.9)	54.93(4.0)	0.70	2.41	52.86(3.2)	52.82(4.3)	0.72	2.32	58.63(3.3)	58.64(4.0)	0.68	2.96
100 ms–2 SD–100 ms	54.91(3.9)	54.93(4.0)	0.70	2.41	52.86(3.2)	52.82(4.3)	0.72	2.32	58.63(3.3)	58.64(4.0)	0.68	2.96
500 ms–2 SD–50 ms	54.93(3.1)	54.91(3.7)	0.65	2.45	52.87(3.0)	52.87(3.7)	0.68	2.42	58.67(3.5)	58.70(4.3)	0.62	2.79
500 ms–2 SD–100 ms	54.93(3.1)	54.91(3.7)	0.65	2.45	52.87(3.0)	52.87(3.7)	0.68	2.42	58.67(3.5)	58.70(4.3)	0.62	2.79
50 ms–3 SD–50 ms	54.91(2.3)	54.95(3.0)	0.88	2.30	52.85(3.2)	52.86(4.2)	0.87	2.40	58.80(3.9)	58.83(3.0)	0.89	2.51
50 ms–3 SD–100 ms	54.91(2.3)	54.95(3.0)	0.88	2.30	52.85(3.2)	52.86(4.2)	0.87	2.40	58.80(3.9)	58.83(3.0)	0.89	2.51
100 ms–3 SD–50 ms	54.93(3.9)	54.90(4.1)	0.80	2.47	52.87(3.0)	52.89(3.6)	0.79	2.48	58.77(3.6)	58.85(3.8)	0.77	2.77
100 ms–3 SD–100 ms	54.93(3.9)	54.90(4.1)	0.80	2.47	52.87(3.0)	52.89(3.6)	0.79	2.48	58.77(3.6)	58.85(3.8)	0.77	2.77
500 ms–3 SD–50 ms	54.95(3.9)	54.98(4.0)	0.73	2.53	52.89(4.0)	52.90(4.2)	0.70	2.64	58.74(4.7)	58.88(3.8)	0.71	2.73
500 ms–3 SD–100 ms	54.95(3.9)	54.98(4.0)	0.73	2.53	52.89(4.0)	52.90(4.2)	0.70	2.64	58.74(4.7)	58.88(3.8)	0.71	2.73

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies for the Functional Ankle Instability Group's Stable Ankle when Acting as the Support Limb.

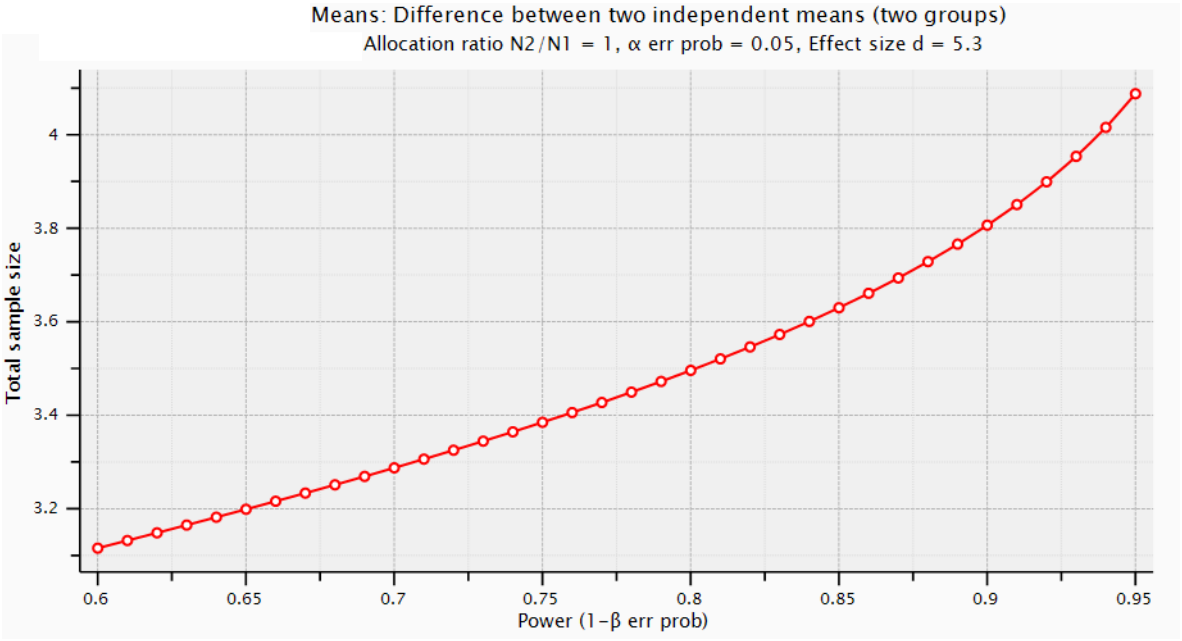
COMBINATION (Baseline–Deviation–Number of Samples)	Peroneus Longus				Tibialis Anterior				Gluteus Medius			
	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)	Test 1 (ms)	Test 2 (ms)	ICC	SEM (%)
50 ms–1 SD–50 ms	66.16(4.0)	66.26(3.0)	0.71	2.71	66.85(3.7)	66.83(3.9)	0.73	2.64	68.28(4.9)	68.23(3.0)	0.70	2.78
50 ms–1 SD–100 ms	66.16(4.0)	66.26(3.0)	0.71	2.71	66.85(3.7)	66.83(3.9)	0.73	2.64	68.28(4.9)	68.23(3.0)	0.70	2.78
100 ms–1 SD–50 ms	66.19(4.8)	66.23(4.4)	0.68	2.78	66.87(3.9)	66.86(3.7)	0.69	2.78	68.29(3.9)	68.31(3.2)	0.67	2.91
100 ms–1 SD–100 ms	66.19(4.8)	66.23(4.4)	0.68	2.78	66.87(3.9)	66.86(3.7)	0.69	2.78	68.29(3.9)	68.31(3.2)	0.67	2.91
500 ms–1 SD–50 ms	66.22(3.3)	66.25(2.8)	0.62	3.10	66.88(3.9)	66.89(3.0)	0.68	2.97	68.31(4.2)	68.33(4.5)	0.66	2.89
500 ms–1 SD–100 ms	66.22(3.3)	66.25(2.8)	0.62	3.10	66.88(3.9)	66.89(3.0)	0.68	2.97	68.31(4.2)	68.33(4.5)	0.66	2.89
50 ms–2 SD–50 ms	66.25(3.5)	66.27(2.6)	0.78	2.70	66.89(3.9)	66.92(3.5)	0.74	2.64	68.35(3.8)	68.40(4.8)	0.73	2.72
50 ms–2 SD–100 ms	66.25(3.5)	66.27(2.6)	0.78	2.70	66.89(3.9)	66.92(3.5)	0.74	2.64	68.35(3.8)	68.40(4.8)	0.73	2.72
100 ms–2 SD–50 ms	66.26(3.0)	66.28(3.6)	0.68	2.90	66.92(3.7)	66.87(3.1)	0.67	2.79	68.32(3.9)	68.41(4.5)	0.67	3.03
100 ms–2 SD–100 ms	66.26(3.0)	66.28(3.6)	0.68	2.90	66.92(3.7)	66.87(3.1)	0.67	2.79	68.32(3.9)	68.41(4.5)	0.67	3.03
500 ms–2 SD–50 ms	66.27(3.9)	66.29(3.0)	0.62	2.83	66.91(3.1)	66.93(3.8)	0.61	2.81	68.33(3.7)	68.35(4.6)	0.63	2.93
500 ms–2 SD–100 ms	66.27(3.9)	66.29(3.0)	0.62	2.83	66.91(3.1)	66.93(3.8)	0.61	2.81	68.33(3.7)	68.35(4.6)	0.63	2.93
50 ms–3 SD–50 ms	66.29(3.1)	66.30(3.5)	0.82	2.33	66.93(3.0)	66.95(3.2)	0.83	2.46	68.38(2.9)	68.40(2.8)	0.83	2.63
50 ms–3 SD–100 ms	66.29(3.1)	66.30(3.5)	0.82	2.33	66.93(3.0)	66.95(3.2)	0.83	2.46	68.38(2.9)	68.40(2.8)	0.83	2.63
100 ms–3 SD–50 ms	66.30(3.9)	66.32(3.8)	0.72	2.62	66.93(3.4)	66.94(3.7)	0.75	2.55	68.40(3.7)	68.42(3.9)	0.73	2.73
100 ms–3 SD–100 ms	66.30(3.9)	66.32(3.8)	0.72	2.62	66.93(3.4)	66.94(3.7)	0.75	2.55	68.40(3.7)	68.42(3.9)	0.73	2.73
500 ms–3 SD–50 ms	66.33(3.8)	66.34(3.6)	0.69	2.80	66.94(3.7)	66.92(3.8)	0.70	2.65	68.43(4.3)	68.42(3.7)	0.68	2.80
500 ms–3 SD–100 ms	66.33(3.8)	66.34(3.6)	0.69	2.80	66.94(3.7)	66.92(3.8)	0.70	2.65	68.43(4.3)	68.42(3.7)	0.68	2.80

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

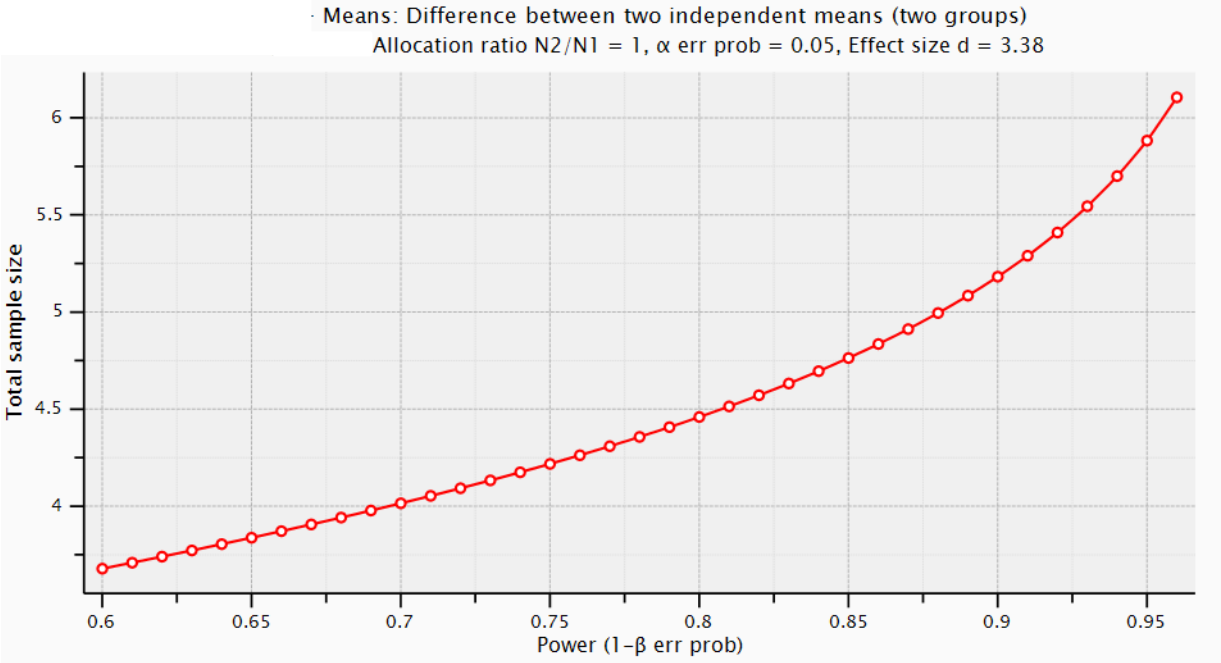
Muscle Latencies (ms) Following an Inversion and Plantarflexion Perturbation.

Condition	Control Group		FAI Group	
	DA	NDA	UA	SA
***TILT LIMB* (ms)**				
Peroneus Longus	48.1 (4.3)	49.0 (4.2)	55.5 (4.9)*	54.0 (4.1)†
Tibialis Anterior	46.7 (4.0)	48.1 (4.7)	53.7 (4.9)*	52.5 (3.5)†
Gluteus Medius	52.4 (4.4)	53.0 (4.0)	59.0 (4.1)*	57.7 (3.9)†
***SUPPORT LIMB* (ms)**				
Peroneus Longus	64.5 (5.4)	65.3 (5.9)	66.8 (6.3)	67.3 (6.1)
Tibialis Anterior	65.1 (6.2)	66.3 (5.9)	67.9 (5.7)	68.5 (6.4)
Gluteus Medius	66.7 (6.0)	67.5 (5.8)	68.7 (6.2)	69.6 (6.4)

Results are presented as Mean (SD). DA: Dominant Ankle, NDA: Non-Dominant Ankle, UA: Unstable Ankle, SA: Stable Ankle. * UA significantly (P<0.025) slower than DA and NDA. † SA significantly (P<0.025) slower than DA and NDA.



The Effect Size Generator (Version 2.3.0, Australia) software was used to calculate Cohen's d effect size. The ankle condition during the 200 ms analysis that generated the smallest effect size was used for the calculation of power. An effect size of d = 5.3 was calculated from the healthy groups NDA's lateral sway distance (mean = 2.76 cm, SD = 0.30) and the FAI groups SA's lateral sway distance (mean = 4.42 cm, SD = 0.35). The plot produced by G*Power (Version 3.1.5, Germany) (Faul et al., 2007) indicated that a total sample size of 4 subjects would be needed to achieve a power of 94% between two independent groups. Therefore, in Study Two and Study Four of this thesis a minimum of 2 subjects will be required in each group.



The Effect Size Generator (Version 2.3.0, Australia) software was used to calculate Cohen's d effect size. The ankle condition during the 200 ms analysis that generated the smallest effect size was used for the calculation of power. An effect size of d = 3.38 was calculated from the healthy groups NDA's mediolateral sway distance (mean = 4.30 cm, SD = 0.41) and the FAI groups SA's mediolateral sway distance (mean = 6.12 cm, SD = 0.64). The plot produced by G*Power (Version 3.1.5, Germany) (Faul et al., 2007) indicated that a total sample size of 6 subjects would be needed to achieve a power of >95% between two independent groups. Therefore, in Study Two and Study Four of this thesis a minimum of 3 subjects will be required in each group.

Anterior and Posterior Sway Distance for the Control Group During a Single Leg Drop Jump Landing.

COMBINATION (Sampling Rate – Balance Duration)	Control Group: Dominant Ankle				Control Group: Non-Dominant Ankle			
	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)
ANTERIOR SWAY								
200 Hz – 200 ms	3.38 (0.18)	3.45 (0.23)	0.87	2.03	3.22 (0.23)	3.37 (0.22)	0.83	2.11
500 Hz – 200 ms	3.39 (0.23)	3.47 (0.25)	0.85	2.12	3.35 (0.25)	3.39 (0.32)	0.79	2.28
1000 Hz – 200 ms	3.33 (0.24)	3.31 (0.19)	0.79	2.32	3.31 (0.33)	3.27 (0.20)	0.72	2.39
200 Hz – 3 secs	7.74 (0.83)	7.65 (0.75)	0.91	5.03	8.56 (0.91)	8.67 (1.01)	0.89	5.56
500 Hz – 3 secs	7.80 (0.90)	7.92 (0.83)	0.86	5.23	7.98 (0.85)	8.59 (0.89)	0.82	5.76
1000 Hz – 3 secs	7.79 (0.87)	7.84 (0.92)	0.77	5.61	8.23 (0.93)	8.40 (0.93)	0.79	5.81
POSTERIOR SWAY								
200 Hz – 200 ms	5.12 (0.36)	5.20 (0.32)	0.85	2.13	4.98 (0.32)	5.03 (0.37)	0.80	2.22
500 Hz – 200 ms	5.10 (0.31)	5.16 (0.41)	0.83	2.24	5.09 (0.34)	4.90 (0.30)	0.76	2.34
1000 Hz – 200 ms	5.21 (0.39)	5.19 (0.33)	0.76	2.38	5.04 (0.37)	4.95 (0.28)	0.70	2.42
200 Hz – 3 secs	12.53 (1.12)	12.43 (1.03)	0.89	5.14	12.81 (1.21)	12.93 (1.19)	0.84	5.67
500 Hz – 3 secs	12.46 (1.15)	12.55 (1.17)	0.84	5.38	12.89 (1.26)	12.75 (1.12)	0.80	5.88
1000 Hz – 3 secs	12.55 (1.21)	12.56 (1.18)	0.75	5.73	12.74 (1.22)	12.80 (1.09)	0.77	5.93

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Medial and Lateral Sway Distance for the Control Group During a Single Leg Drop Jump Landing.

COMBINATION (Sampling Rate – Balance Duration)	Control Group: Dominant Ankle				Control Group: Non-Dominant Ankle			
	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)
MEDIAL SWAY								
200 Hz – 200 ms	1.60 (0.09)	1.63 (0.11)	0.85	2.25	1.67 (0.12)	1.63 (0.10)	0.82	2.21
500 Hz – 200 ms	1.67 (0.13)	1.62 (0.12)	0.82	2.34	1.60 (0.09)	1.68 (0.12)	0.78	2.28
1000 Hz – 200 ms	1.61 (0.08)	1.59 (0.08)	0.78	2.43	1.71 (0.14)	1.73 (0.13)	0.70	2.40
200 Hz – 3 secs	5.23 (0.46)	5.37 (0.45)	0.88	5.23	5.57 (0.52)	5.49 (0.49)	0.89	5.32
500 Hz – 3 secs	5.32 (0.53)	5.30 (0.51)	0.84	5.37	5.53 (0.50)	5.60 (0.53)	0.80	5.42
1000 Hz – 3 secs	5.27 (0.49)	5.37 (0.55)	0.76	5.43	5.59 (0.56)	5.62 (0.59)	0.75	5.39
LATERAL SWAY								
200 Hz – 200 ms	2.58 (0.16)	2.62 (0.19)	0.84	2.45	2.61 (0.18)	2.67 (0.20)	0.80	2.31
500 Hz – 200 ms	2.55 (0.11)	2.52 (0.10)	0.81	2.54	2.59 (0.17)	2.56 (0.16)	0.75	2.48
1000 Hz – 200 ms	2.61 (0.18)	2.58 (0.17)	0.75	2.63	2.55 (0.17)	2.60 (0.17)	0.71	2.50
200 Hz – 3 secs	6.21 (0.71)	6.27 (0.73)	0.86	5.43	6.32 (0.73)	6.29 (0.73)	0.84	5.42
500 Hz – 3 secs	6.28 (0.78)	6.22 (0.71)	0.83	5.57	6.27 (0.70)	6.35 (0.77)	0.77	5.52
1000 Hz – 3 secs	6.19 (0.71)	6.31 (0.80)	0.72	5.63	6.33 (0.76)	6.25 (0.67)	0.70	5.69

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Anteroposterior and Mediolateral Sway Distance for the Control Group During a Single Leg Drop Jump Landing.

COMBINATION (Sampling Rate – Balance Duration)	Control Group: Dominant Ankle				Control Group: Non-Dominant Ankle			
	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)
A/P SWAY								
200 Hz – 200 ms	8.50 (1.21)	8.65 (1.30)	0.82	4.01	8.20 (1.22)	8.40 (1.25)	0.80	4.11
500 Hz – 200 ms	8.49 (1.20)	8.63 (1.35)	0.80	4.14	8.44 (1.30)	8.29 (1.21)	0.79	4.34
1000 Hz – 200 ms	8.54 (1.34)	8.50 (1.37)	0.72	4.22	8.35 (1.29)	8.22 (1.16)	0.73	4.65
200 Hz – 3 secs	20.27 (2.53)	20.08 (2.65)	0.84	9.83	21.37 (2.56)	21.60 (2.69)	0.85	9.72
500 Hz – 3 secs	20.26 (2.45)	20.47 (2.59)	0.80	10.23	20.87 (2.61)	21.34 (2.65)	0.80	10.21
1000 Hz – 3 secs	20.34 (2.58)	20.40 (2.61)	0.73	10.83	20.97 (2.55)	21.20 (2.51)	0.74	10.32
M/L SWAY								
200 Hz – 200 ms	4.18 (0.32)	4.25 (0.35)	0.81	4.11	4.28 (0.30)	4.30 (0.31)	0.80	4.21
500 Hz – 200 ms	4.22 (0.38)	4.14 (0.36)	0.78	4.24	4.19 (0.32)	4.24 (0.34)	0.77	4.34
1000 Hz – 200 ms	4.22 (0.29)	4.17 (0.33)	0.72	4.32	4.26 (0.35)	4.33 (0.30)	0.73	4.45
200 Hz – 3 secs	11.44 (0.98)	11.64 (0.95)	0.83	9.96	11.89 (0.90)	11.78 (0.94)	0.84	9.82
500 Hz – 3 secs	11.60 (1.10)	11.52 (0.98)	0.79	10.33	11.80 (0.95)	11.95 (0.99)	0.81	10.31
1000 Hz – 3 secs	11.46 (1.04)	11.68 (1.05)	0.72	10.63	11.92 (1.03)	11.87 (0.96)	0.77	10.42

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Anterior and Posterior Sway Distance for the Functional Ankle Instability Group During a Single Leg Drop Jump Landing.

COMBINATION (Sampling Rate – Balance Duration)	FAI Group: Unstable Ankle				FAI Group: Stable Ankle			
	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)
ANTERIOR SWAY								
200 Hz – 200 ms	3.45 (0.21)	3.42 (0.23)	0.86	2.14	3.39 (0.18)	3.41 (0.20)	0.82	2.22
500 Hz – 200 ms	3.44 (0.19)	3.45 (0.28)	0.83	2.32	3.41 (0.19)	3.50 (0.24)	0.76	2.38
1000 Hz – 200 ms	3.52 (0.23)	3.49 (0.25)	0.80	2.42	3.45 (0.24)	3.46 (0.27)	0.72	2.49
200 Hz – 3 secs	8.46 (1.02)	8.42 (0.98)	0.87	5.13	7.87 (0.84)	7.94 (0.72)	0.85	5.66
500 Hz – 3 secs	8.40 (0.90)	8.38 (0.89)	0.84	5.33	8.01 (0.87)	8.12 (0.89)	0.80	5.86
1000 Hz – 3 secs	8.52 (1.06)	8.45 (0.94)	0.78	5.71	8.23 (0.92)	8.35 (0.98)	0.73	5.91
POSTERIOR SWAY								
200 Hz – 200 ms	5.34 (0.43)	5.30 (0.40)	0.83	2.24	5.21 (0.41)	5.16 (0.44)	0.80	2.32
500 Hz – 200 ms	5.31 (0.41)	5.38 (0.45)	0.81	2.34	5.17 (0.38)	5.21 (0.40)	0.71	2.44
1000 Hz – 200 ms	5.39 (0.45)	5.43 (0.47)	0.74	2.48	5.26 (0.43)	5.29 (0.43)	0.66	2.52
200 Hz – 3 secs	13.42 (1.04)	13.40 (1.10)	0.87	5.24	13.12 (0.97)	13.12 (0.94)	0.80	5.77
500 Hz – 3 secs	13.37 (0.98)	13.35 (0.97)	0.82	5.48	13.18 (1.02)	13.20 (0.98)	0.72	5.98
1000 Hz – 3 secs	13.33 (0.92)	13.34 (0.91)	0.72	5.84	13.21 (0.99)	13.25 (1.05)	0.68	5.99

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Medial and Lateral Sway Distance for the Functional Ankle Instability Group During a Single Leg Drop Jump Landing.

COMBINATION (Sampling Rate – Balance Duration)	FAI Group: Unstable Ankle				FAI Group: Stable Ankle			
	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)
MEDIAL SWAY								
200 Hz – 200 ms	1.74 (0.11)	1.80 (0.17)	0.83	2.31	1.70 (0.09)	1.72 (0.10)	0.81	2.35
500 Hz – 200 ms	1.72 (0.14)	1.75 (0.16)	0.80	2.44	1.72 (0.11)	1.75 (0.12)	0.79	2.38
1000 Hz – 200 ms	1.77 (0.17)	1.75 (0.15)	0.74	2.53	1.76 (0.14)	1.77 (0.16)	0.72	2.53
200 Hz – 3 secs	5.62 (0.57)	5.60 (0.51)	0.85	5.33	5.45 (0.50)	5.49 (0.53)	0.85	5.42
500 Hz – 3 secs	5.57 (0.51)	5.51 (0.48)	0.82	5.47	5.55 (0.56)	5.58 (0.60)	0.77	5.52
1000 Hz – 3 secs	5.52 (0.50)	5.48 (0.49)	0.72	5.53	5.49 (0.54)	5.52 (0.52)	0.70	5.49
LATERAL SWAY								
200 Hz – 200 ms	4.50 (0.23)	4.54 (0.26)	0.83	2.41	4.42 (0.25)	4.47 (0.28)	0.81	2.36
500 Hz – 200 ms	4.45 (0.30)	4.48 (0.32)	0.78	2.52	4.41 (0.23)	4.48 (0.30)	0.72	2.53
1000 Hz – 200 ms	4.48 (0.28)	4.41 (0.29)	0.74	2.57	4.40 (0.30)	4.41 (0.24)	0.66	2.52
200 Hz – 3 secs	6.62 (0.67)	6.69 (0.70)	0.85	5.49	6.75 (0.70)	6.69 (0.73)	0.81	5.48
500 Hz – 3 secs	6.69 (0.71)	6.71 (0.68)	0.82	5.60	6.70 (0.67)	6.67 (0.69)	0.73	5.59
1000 Hz – 3 secs	6.70 (0.68)	6.73 (0.72)	0.68	5.69	6.78 (0.72)	6.73 (0.75)	0.67	5.72

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Anteroposterior and Mediolateral Sway Distance for the Functional Ankle Instability Group During a Single Leg Drop Jump.

COMBINATION (Sampling Rate – Balance Duration)	FAI Group: Unstable Ankle				FAI Group: Stable Ankle			
	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)	Test 1 (cm)	Test 2 (cm)	ICC	SEM (%)
A/P SWAY								
200 Hz – 200 ms	8.79 (0.92)	8.72 (0.90)	0.80	4.11	8.60 (0.90)	8.57 (0.89)	0.80	4.24
500 Hz – 200 ms	8.75 (0.93)	8.83 (0.91)	0.78	4.24	8.58 (0.92)	8.71 (0.93)	0.75	4.44
1000 Hz – 200 ms	8.91 (0.98)	8.92 (0.96)	0.70	4.32	8.71 (0.99)	8.75 (0.94)	0.70	4.75
200 Hz – 3 secs	21.88 (2.51)	21.82 (2.59)	0.82	9.93	20.99 (2.48)	21.06 (2.49)	0.83	9.82
500 Hz – 3 secs	21.77 (2.55)	21.73 (2.49)	0.77	10.43	21.19 (2.54)	21.32 (2.59)	0.75	10.31
1000 Hz – 3 secs	21.85 (2.59)	21.79 (2.51)	0.69	10.93	21.44 (2.57)	21.60 (2.56)	0.74	10.42
M/L SWAY								
200 Hz – 200 ms	6.24 (0.63)	6.34 (0.62)	0.80	4.22	6.12 (0.64)	6.19 (0.62)	0.81	4.32
500 Hz – 200 ms	6.17 (0.65)	6.23 (0.67)	0.76	4.34	6.13 (0.62)	6.23 (0.67)	0.73	4.44
1000 Hz – 200 ms	6.25 (0.66)	6.16 (0.70)	0.67	4.42	6.16 (0.60)	6.18 (0.62)	0.67	4.53
200 Hz – 3 secs	12.24 (1.12)	12.29 (1.03)	0.81	9.98	12.20 (1.16)	12.18 (1.09)	0.81	9.90
500 Hz – 3 secs	12.26 (1.11)	12.22 (1.10)	0.74	10.43	12.25 (1.20)	12.25 (1.14)	0.77	10.40
1000 Hz – 3 secs	12.22 (1.05)	12.21 (0.99)	0.70	10.73	12.27 (1.05)	12.25 (1.02)	0.72	10.52

Results are presented as Mean (SD). ICC: Intraclass Correlation Coefficient, SEM: Standard Error of Measurement.

Muscle Latencies (ms) for the Tilting Limb

Condition	Control Group		FAI Group	
	DA	NDA	UA	SA
***PERONEUS LONGUS* (ms)**				
Pre Test	48.5 (4.7)	49.4 (4.6)	55.9 (4.7)*	54.3 (4.0)†
Ankle IsoK Fatigue	49.9 (4.6)	50.5 (4.6)	57.3 (4.9)*	55.0 (4.3)†
Hip IsoK Fatigue	49.5 (4.5)	49.9 (4.9)	56.6 (4.8)*	55.1 (4.1)†
Football-Specific Fatigue	50.0 (4.4)	50.3 (4.3)	57.3 (4.5)*	55.4 (4.0)†
Control	48.2 (4.7)	49.1 (4.8)	55.4 (4.1)*	54.1 (3.8)†
***TIBIALIS ANTERIOR* (ms)**				
Pre Test	46.9 (4.6)	48.2 (4.6)	53.2 (4.5)*	52.6 (4.9)†
Ankle IsoK Fatigue	48.0 (4.5)	49.5 (4.2)	55.3 (4.7)*	53.9 (4.0)†
Hip IsoK Fatigue	48.3 (4.4)	48.9 (4.2)	53.9 (4.0)*	53.3 (4.7)†
Football-Specific Fatigue	48.4 (4.2)	49.6 (4.1)	55.2 (4.6)*	54.2 (4.0)†
Control	46.2 (4.5)	48.0 (3.6)	52.9 (4.0)*	52.1 (4.8)†
***GLUTEUS MEDIUS* (ms)**				
Pre Test	53.1 (4.6)	53.4 (4.3)	59.2 (4.8)*	57.7 (4.0)†
Ankle IsoK Fatigue	54.9 (4.5)	55.0 (4.3)	60.4 (4.9)*	58.8 (4.4)†
Hip IsoK Fatigue	55.3 (4.6)	55.3 (4.9)	60.6 (4.7)*	59.4 (4.2)†
Football-Specific Fatigue	55.0 (4.4)	54.9 (4.1)	60.7 (4.1)*	59.2 (4.1)†
Control	53.1 (4.6)	53.2 (4.9)	58.8 (4.0)*	57.4 (4.4)†

Results are presented as Mean (SD). DA: Dominant Ankle, NDA: Non-Dominant Ankle, UA: Unstable Ankle, SA: Stable Ankle. * FAI groups UA significantly (P<0.025) slower for each condition, than the control groups DA and NDA. † FAI groups SA significantly (P<0.025) slower for each condition, than the control groups DA and NDA.

Muscle Latencies (ms) for the Support Limb.

Condition	Control Group		FAI Group	
	DA	**NDA**	**UA**	**SA**
***PERONEUS LONGUS* (ms)**				
Pre Test	64.7 (4.5)	65.4 (4.2)	66.9 (4.3)	67.5 (4.7)
Ankle IsoK Fatigue	65.6 (4.3)	66.7 (4.4)	67.8 (4.7)	68.7 (4.0)
Hip IsoK Fatigue	65.3 (4.3)	66.4 (4.5)	67.5 (4.6)	68.5 (4.8)
Football-Specific Fatigue	65.9 (4.2)	66.9 (4.0)	68.0 (4.4)	68.9 (4.5)
Control	64.9 (4.3)	65.6 (4.5)	66.8 (4.9)	67.3 (4.3)
***TIBIALIS ANTERIOR* (ms)**				
Pre Test	65.6 (4.5)	66.5 (4.2)	68.2 (4.3)	68.8 (4.7)
Ankle IsoK Fatigue	66.7 (4.2)	67.4 (4.4)	69.4 (4.5)	69.5 (4.8)
Hip IsoK Fatigue	66.6 (4.1)	67.2 (4.6)	69.1 (4.4)	69.3 (4.1)
Football-Specific Fatigue	66.8 (4.1)	67.8 (4.9)	69.8 (4.2)	69.0 (4.2)
Control	65.2 (4.4)	66.1 (4.5)	68.0 (4.8)	68.4 (4.2)
***GLUTEUS MEDIUS* (ms)**				
Pre Test	66.9 (4.3)	67.7 (4.0)	68.9 (4.0)	69.9 (4.6)
Ankle IsoK Fatigue	67.7 (4.0)	68.4 (4.2)	69.7 (4.5)	70.5 (4.0)
Hip IsoK Fatigue	67.3 (4.2)	68.2 (4.6)	69.3 (4.8)	70.1 (4.4)
Football-Specific Fatigue	67.9 (4.1)	68.8 (4.1)	69.9 (4.1)	70.6 (4.8)
Control	66.7 (4.2)	67.5 (4.3)	68.6 (4.8)	69.5 (4.4)

Results are presented as Mean (SD). DA: Dominant Ankle, NDA: Non-Dominant Ankle, UA: Unstable Ankle, SA: Stable Ankle.

SPSS Statistical Outputs

Tests of Normality

	SubjectCondition	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
LatencyDAY1	DA	.203	10	.200	.923	10	.378
	NDA	.329	10	.063	.746	10	.073
	UA	.186	10	.200*	.856	10	.068
	SA	.266	10	.200*	.891	10	.172
LatencyDAY2	DA	.276	10	.060	.852	10	.062
	NDA	.249	10	.079	.877	10	.119
	UA	.170	10	.200*	.944	10	.604
	SA	.141	10	.200*	.981	10	.969

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

Day	Dependent Variable
1	LatencyDAY1
2	LatencyDAY2

Between-Subjects Factors

		Value Label	N
Ankle	1.00	DA	10
	2.00	NDA	10
	3.00	UA	10
	4.00	SA	10

Descriptive Statistics

	Ankle	Mean	Std. Deviation	N
LatencyDAY1	DA	48.5390	.37394	10
	NDA	48.3990	.23163	10
	UA	54.7500	.58405	10
	SA	54.4020	.58627	10
	Total	51.5225	3.12788	40
LatencyDAY2	DA	48.3980	.54002	10
	NDA	48.2490	.53488	10
	UA	54.8380	.74055	10
	SA	54.4472	.56611	10
	Total	51.4831	3.25499	40

Box's Test of Equality of Covariance Matrices[a]

Box's M	17.957
F	1.794
df1	9
df2	14851.910
Sig.	.064

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

a. Design: Intercept + Ankle

Within Subjects Design: Day

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
Day	1.000	.000	0	.000	1.000	1.000	1.000

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept + Ankle

Within Subjects Design: Day

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
LatencyDAY1	2.374	3	36	.069
LatencyDAY2	.750	3	36	.530

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

a. Design: Intercept + Ankle

Within Subjects Design: Day

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Day	Pillai's Trace	.007	.257[b]	1.000	36.000	.615	.007
	Wilks' Lambda	.993	.257[b]	1.000	36.000	.615	.007
	Hotelling's Trace	.007	.257[b]	1.000	36.000	.615	.007
	Roy's Largest Root	.007	.257[b]	1.000	36.000	.615	.007
Day * Ankle	Pillai's Trace	.050	.633[b]	3.000	36.000	.598	.050
	Wilks' Lambda	.950	.633[b]	3.000	36.000	.598	.050
	Hotelling's Trace	.053	.633[b]	3.000	36.000	.598	.050
	Roy's Largest Root	.053	.633[b]	3.000	36.000	.598	.050

a. Design: Intercept + Ankle

Within Subjects Design: Day

b. Exact statistic

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	212202.867	1	212202.867	460980.049	.000	1.000
Ankle	773.613	3	257.871	560.188	.000	.979
Error	16.572	36	.460			

Multiple Comparisons

Measure: MEASURE_1

Tukey HSD

(I) Ankle	(J) Ankle	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
	NDA	.1445	.21455	.906	-.4333	.7223
DA	UA	-6.3255*	.21455	.000	-6.9033	-5.7477
	SA	-5.9561*	.21455	.000	-6.5339	-5.3783
	DA	-.1445	.21455	.906	-.7223	.4333
NDA	UA	-6.4700*	.21455	.000	-7.0478	-5.8922
	SA	-6.1006*	.21455	.000	-6.6784	-5.5228
	DA	6.3255*	.21455	.000	5.7477	6.9033
UA	NDA	6.4700*	.21455	.000	5.8922	7.0478
	SA	.3694	.21455	.328	-.2084	.9472
	DA	5.9561*	.21455	.000	5.3783	6.5339
SA	NDA	6.1006*	.21455	.000	5.5228	6.6784
	UA	-.3694	.21455	.328	-.9472	.2084

Based on observed means.

The error term is Mean Square(Error) = .230.

*. The mean difference is significant at the .05 level.

Intraclass Correlation Coefficient

	Intraclass Correlation	95% Confidence Interval		F Test with True Value 0			
		Lower Bound	Upper Bound	Value	df1	df2	Sig
Single Measures	.666	.138	.904	4.994	9	10	.010
Average Measures	.800	.243	.949	4.994	9	10	.010

One-way random effects model where people effects are random.

Tests of Normality

	SubjectCondition	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
LatencyDAY1	DA	.242	10	.102	.842	10	.056
	NDA	.301	10	.051	.699	10	.061
	UA	.185	10	.190*	.887	10	.157
	SA	.169	10	.190*	.954	10	.721
LatencyDAY2	DA	.149	10	.190*	.928	10	.430
	NDA	.171	10	.190*	.920	10	.360
	UA	.176	10	.190*	.952	10	.696
	SA	.112	10	.190*	.988	10	.994

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

Day	Dependent Variable
1	LatencyDAY1
2	LatencyDAY2

Between-Subjects Factors

		Value Label	N
Ankle	1.00	DA	10
	2.00	NDA	10
	3.00	UA	10
	4.00	SA	10

Descriptive Statistics

	Ankle	Mean	Std. Deviation	N
LatencyDAY1	DA	48.4260	.25264	10
	NDA	48.3990	.23163	10
	UA	54.7830	.53821	10
	SA	54.4020	.58627	10
	Total	51.5025	3.15989	40
LatencyDAY2	DA	48.3250	.49232	10
	NDA	48.3540	.45817	10
	UA	54.8380	.74055	10
	SA	54.5330	.43828	10
	Total	51.5125	3.25782	40

Box's Test of Equality of Covariance Matrices[a]

Box's M	20.986
F	2.097
df1	9
df2	14851.910
Sig.	.026

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

a. Design: Intercept + Ankle

Within Subjects Design: Day

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
Day	1.000	.000	0	.000	1.000	1.000	1.000

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept + Ankle

Within Subjects Design: Day

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
LatencyDAY1	2.477	3	36	.109
LatencyDAY2	1.734	3	36	.177

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

a. Design: Intercept + Ankle

Within Subjects Design: Day

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Day	Pillai's Trace	.000	.017[b]	1.000	36.000	.897	.000
	Wilks' Lambda	1.000	.017[b]	1.000	36.000	.897	.000
	Hotelling's Trace	.000	.017[b]	1.000	36.000	.897	.000
	Roy's Largest Root	.000	.017[b]	1.000	36.000	.897	.000
Day * Ankle	Pillai's Trace	.036	.452[b]	3.000	36.000	.718	.036
	Wilks' Lambda	.964	.452[b]	3.000	36.000	.718	.036
	Hotelling's Trace	.038	.452[b]	3.000	36.000	.718	.036
	Roy's Largest Root	.038	.452[b]	3.000	36.000	.718	.036

a. Design: Intercept + Ankle

Within Subjects Design: Day

b. Exact statistic

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	212241.805	1	212241.805	576987.058	.000	1.000
Ankle	785.680	3	261.893	711.966	.000	.983
Error	13.242	36	.368			

Multiple Comparisons

Measure: MEASURE_1

Tukey HSD

(I) Ankle	(J) Ankle	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
	NDA	-.0010	.19179	1.000	-.5175	.5155
DA	UA	-6.4350*	.19179	.000	-6.9515	-5.9185
	SA	-6.0920*	.19179	.000	-6.6085	-5.5755
	DA	.0010	.19179	1.000	-.5155	.5175
NDA	UA	-6.4340*	.19179	.000	-6.9505	-5.9175
	SA	-6.0910*	.19179	.000	-6.6075	-5.5745
	DA	6.4350*	.19179	.000	5.9185	6.9515
UA	NDA	6.4340*	.19179	.000	5.9175	6.9505
	SA	.3430	.19179	.295	-.1735	.8595
	DA	6.0920*	.19179	.000	5.5755	6.6085
SA	NDA	6.0910*	.19179	.000	5.5745	6.6075
	UA	-.3430	.19179	.295	-.8595	.1735

Based on observed means.

The error term is Mean Square(Error) = .184.

*. The mean difference is significant at the .05 level.

Intraclass Correlation Coefficient

	Intraclass Correlation	95% Confidence Interval		F Test with True Value 0			
		Lower Bound	Upper Bound	Value	df1	df2	Sig
Single Measures	.527	-.078	.855	3.231	9	10	.041
Average Measures	.691	-.170	.922	3.231	9	10	.041

One-way random effects model where people effects are random.

Tests of Normality

	Ankle	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
MuscleLatency	DA	.276	20	.071	.819	20	.102
	NDA	.099	20	.180*	.964	20	.618
	UA	.149	20	.190*	.912	20	.070
	SA	.189	20	.060	.955	20	.458

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Descriptives

MuscleLatency

	N	Mean	Std. Deviation	Std. Error	95% Confidence Interval for Mean		Minimum	Maximum
					Lower Bound	Upper Bound		
DA	20	48.4860	.31987	.07153	48.3363	48.6357	48.13	49.12
NDA	20	48.2026	.38764	.08668	48.0211	48.3840	47.20	48.90
UA	20	54.4850	.57669	.12895	54.2151	54.7549	53.40	55.43
SA	20	54.5325	.43200	.09660	54.3303	54.7347	53.80	55.60
Total	80	51.4265	3.13313	.35029	50.7293	52.1238	47.20	55.60

Test of Homogeneity of Variances

MuscleLatency

Levene Statistic	df1	df2	Sig.
1.858	3	76	.144

ANOVA

MuscleLatency

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	760.841	3	253.614	1314.440	.000
Within Groups	14.664	76	.193		
Total	775.505	79			

Multiple Comparisons

Dependent Variable: MuscleLatency

Tukey HSD

(I) Ankle	(J) Ankle	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
	NDA	.28345	.13890	.182	-.0814	.6483
DA	UA	-5.99900*	.13890	.000	-6.3639	-5.6341
	SA	-6.04650*	.13890	.000	-6.4114	-5.6816
	DA	-.28345	.13890	.182	-.6483	.0814
NDA	UA	-6.28245*	.13890	.000	-6.6473	-5.9176
	SA	-6.32995*	.13890	.000	-6.6948	-5.9651
	DA	5.99900*	.13890	.000	5.6341	6.3639
UA	NDA	6.28245*	.13890	.000	5.9176	6.6473
	SA	-.04750	.13890	.986	-.4124	.3174
	DA	6.04650*	.13890	.000	5.6816	6.4114
SA	NDA	6.32995*	.13890	.000	5.9651	6.6948
	UA	.04750	.13890	.986	-.3174	.4124

*. The mean difference is significant at the 0.05 level.

Tests of Normality

	Ankle	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
SwayDAY1	DA	.219	10	.191	.913	10	.303
	NDA	.215	10	.200*	.926	10	.410
	UA	.196	10	.200*	.939	10	.543
	SA	.147	10	.200*	.956	10	.745
SwayDAY2	DA	.269	10	.058	.927	10	.416
	NDA	.170	10	.200*	.913	10	.304
	UA	.262	10	.051	.933	10	.476
	SA	.219	10	.189	.865	10	.087

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

TestDay	Dependent Variable
1	SwayDAY1
2	SwayDAY2

Between-Subjects Factors

		Value Label	N
Ankle	1.00	DA	10
	2.00	NDA	10
	3.00	UA	10
	4.00	SA	10

Descriptive Statistics

	Ankle	Mean	Std. Deviation	N
SwayDAY1	DA	2.5160	.07919	10
	NDA	2.4330	.07973	10
	UA	4.4800	.07242	10
	SA	4.4390	.05043	10
	Total	3.4670	1.00803	40
SwayDAY2	DA	2.5260	.08003	10
	NDA	2.4800	.06831	10
	UA	4.3590	.09351	10
	SA	4.4450	.08223	10
	Total	3.4525	.96541	40

Box's Test of Equality of Covariance Matrices[a]

Box's M	11.236
F	1.122
df1	9
df2	14851.910
Sig.	.342

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

a. Design: Intercept + Ankle

Within Subjects Design: TestDay

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
TestDay	1.000	.000	0	.000	1.000	1.000	1.000

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept + Ankle

Within Subjects Design: TestDay

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
SwayDAY1	.555	3	36	.648
SwayDAY2	.421	3	36	.739

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

a. Design: Intercept + SubjectCondition

Within Subjects Design: TestDay

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
TestDay	Pillai's Trace	.019	.701[b]	1.000	36.000	.408	.019
	Wilks' Lambda	.981	.701[b]	1.000	36.000	.408	.019
	Hotelling's Trace	.019	.701[b]	1.000	36.000	.408	.019
	Roy's Largest Root	.019	.701[b]	1.000	36.000	.408	.019
TestDay * Ankle	Pillai's Trace	.272	4.483[b]	3.000	36.000	.009	.272
	Wilks' Lambda	.728	4.483[b]	3.000	36.000	.009	.272
	Hotelling's Trace	.374	4.483[b]	3.000	36.000	.009	.272
	Roy's Largest Root	.374	4.483[b]	3.000	36.000	.009	.272

a. Design: Intercept + Ankle

Within Subjects Design: TestDay

b. Exact statistic

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	957.590	1	957.590	166658.089	.000	1.000
Ankle	75.474	3	25.158	4378.474	.000	.997
Error	.207	36	.006			

Multiple Comparisons

Measure: MEASURE_1

Tukey HSD

(I) SubjectCondition	(J) Ankle	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
	NDA	.0645	.02397	.060	-.0001	.1291
DA	UA	-1.8985*	.02397	.000	-1.9631	-1.8339
	SA	-1.9210*	.02397	.000	-1.9856	-1.8564
	DA	-.0645	.02397	.060	-.1291	.0001
NDA	UA	-1.9630*	.02397	.000	-2.0276	-1.8984
	SA	-1.9855*	.02397	.000	-2.0501	-1.9209
	DA	1.8985*	.02397	.000	1.8339	1.9631
UA	NDA	1.9630*	.02397	.000	1.8984	2.0276
	SA	-.0225	.02397	.784	-.0871	.0421
	DA	1.9210*	.02397	.000	1.8564	1.9856
SA	NDA	1.9855*	.02397	.000	1.9209	2.0501
	UA	.0225	.02397	.784	-.0421	.0871

Based on observed means.

The error term is Mean Square(Error) = .003.

*. The mean difference is significant at the .05 level.

Intraclass Correlation Coefficient

	Intraclass Correlation	95% Confidence Interval		F Test with True Value 0			
		Lower Bound	Upper Bound	Value	df1	df2	Sig
Single Measures	.739	.276	.927	6.654	9	10	.003
Average Measures	.850	.432	.962	6.654	9	10	.003

One-way random effects model where people effects are random.

Tests of Normality

	SubjectCondition	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
TotalSway	DA	.247	20	.072	.912	20	.069
	NDA	.186	20	.068	.936	20	.199
	UA	.141	20	.200*	.955	20	.446
	SA	.180	20	.090	.936	20	.202

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Descriptives

TotalSway

	N	Mean	Std. Deviation	Std. Error	95% Confidence Interval for Mean		Minimum	Maximum
					Lower Bound	Upper Bound		
DA	20	2.5210	.07766	.01736	2.4847	2.5573	2.40	2.67
NDA	20	2.4565	.07618	.01703	2.4208	2.4922	2.30	2.59
UA	20	4.4195	.10237	.02289	4.3716	4.4674	4.21	4.60
SA	20	4.4420	.06646	.01486	4.4109	4.4731	4.32	4.54
Total	80	3.4598	.98071	.10965	3.2415	3.6780	2.30	4.60

Test of Homogeneity of Variances

TotalSway

Levene Statistic	df1	df2	Sig.
1.148	3	76	.335

ANOVA

TotalSway

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	75.474	3	25.158	3764.904	.000
Within Groups	.508	76	.007		
Total	75.982	79			

Multiple Comparisons

Dependent Variable: TotalSway

Tukey HSD

(I) SubjectCondition	(J) SubjectCondition	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
	NDA	.06450	.02585	.069	-.0034	.1324
DA	UA	-1.89850*	.02585	.000	-1.9664	-1.8306
	SA	-1.92100*	.02585	.000	-1.9889	-1.8531
	DA	-.06450	.02585	.069	-.1324	.0034
NDA	UA	-1.96300*	.02585	.000	-2.0309	-1.8951
	SA	-1.98550*	.02585	.000	-2.0534	-1.9176
	DA	1.89850*	.02585	.000	1.8306	1.9664
UA	NDA	1.96300*	.02585	.000	1.8951	2.0309
	SA	-.02250	.02585	.820	-.0904	.0454
	DA	1.92100*	.02585	.000	1.8531	1.9889
SA	NDA	1.98550*	.02585	.000	1.9176	2.0534
	UA	.02250	.02585	.820	-.0454	.0904

*. The mean difference is significant at the 0.05 level.

Tests of Normality

	SubjectType	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
Speed60Test1	Healthy	.226	10	.159	.877	10	.121
	FAI	.225	10	.163	.886	10	.155
Speed60Test2	Healthy	.237	10	.116	.864	10	.086
	FAI	.225	10	.163	.886	10	.155
Speed90Test1	Healthy	.215	10	.190	.894	10	.188
	FAI	.225	10	.163	.886	10	.155
Speed90Test2	Healthy	.214	10	.190	.887	10	.155
	FAI	.195	10	.190	.900	10	.222
Speed120Test1	Healthy	.236	10	.120	.878	10	.123
	FAI	.238	10	.115	.887	10	.158
Speed120Test2	Healthy	.222	10	.177	.877	10	.120
	FAI	.238	10	.115	.887	10	.158
Speed180Test1	Healthy	.222	10	.177	.888	10	.161
	FAI	.238	10	.115	.887	10	.158
Speed180Test2	Healthy	.158	10	.190	.940	10	.549
	FAI	.237	10	.117	.892	10	.180

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

Speed	Day	Dependent Variable
1	1	Speed60Test1
	2	Speed60Test2
2	1	Speed90Test1
	2	Speed90Test2
3	1	Speed120Test1
	2	Speed120Test2
4	1	Speed180Test1
	2	Speed180Test2

Between-Subjects Factors

		Value Label	N
SubjectType	1.00	Healthy	10
	2.00	FAI	10

Descriptive Statistics

	SubjectType	Mean	Std. Deviation	N
Speed60Test1	Healthy	-.7810	.10765	10
	FAI	-.7420	.10912	10
	Total	-.7615	.10737	20
Speed60Test2	Healthy	-.7260	.10772	10
	FAI	-.7120	.10912	10
	Total	-.7190	.10578	20
Speed90Test1	Healthy	-.7500	.11155	10
	FAI	-.7620	.10912	10
	Total	-.7560	.10758	20
Speed90Test2	Healthy	-.7550	.11750	10
	FAI	-.8010	.11060	10
	Total	-.7780	.11354	20
Speed120Test1	Healthy	-.7240	.10617	10
	FAI	-.7250	.10659	10
	Total	-.7245	.10354	20
Speed120Test2	Healthy	-.7080	.10612	10
	FAI	-.7050	.10659	10
	Total	-.7065	.10353	20
Speed180Test1	Healthy	-.7060	.10394	10
	FAI	-.6950	.10659	10
	Total	-.7005	.10262	20
Speed180Test2	Healthy	-.6710	.09422	10
	FAI	-.6730	.10478	10
	Total	-.6720	.09699	20

Box's Test of Equality of Covariance Matrices[a]

Box's M	116.532
F	1.622
df1	36
df2	1090.213
Sig.	.062

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

a. Design: Intercept + SubjectType

Within Subjects Design: Speed + Day + Speed * Day

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
Speed	.077	42.821	5	.070	.423	.466	.333
Day	1.000	.000	0	.	1.000	1.000	1.000
Speed * Day	.381	16.122	5	.087	.748	.907	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept + SubjectType

Within Subjects Design: Speed + Day + Speed * Day

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
Speed60Test1	.003	1	18	.956
Speed60Test2	.004	1	18	.948
Speed90Test1	.000	1	18	1.000
Speed90Test2	.038	1	18	.848
Speed120Test1	.003	1	18	.959
Speed120Test2	.003	1	18	.959
Speed180Test1	.025	1	18	.875
Speed180Test2	.387	1	18	.542

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

a. Design: Intercept + SubjectType

Within Subjects Design: Speed + Day + Speed * Day

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Speed	Pillai's Trace	.950	102.078[b]	3.000	16.000	.000	.950
	Wilks' Lambda	.050	102.078[b]	3.000	16.000	.000	.950
	Hotelling's Trace	19.140	102.078[b]	3.000	16.000	.000	.950
	Roy's Largest Root	19.140	102.078[b]	3.000	16.000	.000	.950
Speed * SubjectType	Pillai's Trace	.951	1.551[b]	3.000	16.000	.550	.051
	Wilks' Lambda	.049	1.551[b]	3.000	16.000	.550	.051
	Hotelling's Trace	19.228	1.551[b]	3.000	16.000	.550	.051
	Roy's Largest Root	19.228	1.551[b]	3.000	16.000	.550	.051
Day	Pillai's Trace	.873	1.362[b]	1.000	18.000	.600	.073
	Wilks' Lambda	.127	1.362[b]	1.000	18.000	.600	.073
	Hotelling's Trace	6.853	1.362[b]	1.000	18.000	.600	.073
	Roy's Largest Root	6.853	1.362[b]	1.000	18.000	.600	.073
Day * SubjectType	Pillai's Trace	.638	1.768[b]	1.000	18.000	.678	.068
	Wilks' Lambda	.362	1.768[b]	1.000	18.000	.678	.068
	Hotelling's Trace	1.765	1.768[b]	1.000	18.000	.678	.068
	Roy's Largest Root	1.765	1.768[b]	1.000	18.000	.678	.068
Speed * Day	Pillai's Trace	.958	2.908[b]	3.000	16.000	.800	.058
	Wilks' Lambda	.042	2.908[b]	3.000	16.000	.800	.058
	Hotelling's Trace	23.045	2.908[b]	3.000	16.000	.800	.058
	Roy's Largest Root	23.045	2.908[b]	3.000	16.000	.800	.058
Speed * Day * SubjectType	Pillai's Trace	.890	2.955[b]	3.000	16.000	.920	.030
	Wilks' Lambda	.110	2.955[b]	3.000	16.000	.920	.030
	Hotelling's Trace	8.054	2.955[b]	3.000	16.000	.920	.030
	Roy's Largest Root	8.054	2.955[b]	3.000	16.000	.920	.030

a. Design: Intercept + SubjectType

Within Subjects Design: Speed + Day + Speed * Day

b. Exact statistic

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	84.623	1	84.623	924.296	.000	.981
SubjectType	2.250E-005	1	2.250E-005	.000	.988	.000
Error	1.648	18	.092			

Pairwise Comparisons

Measure: MEASURE_1

(I) Speed	(J) Speed	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.027*	.002	.000	.020	.033
	3	-.025*	.002	.000	-.029	-.020
	4	-.054*	.003	.000	-.063	-.045
2	1	-.027*	.002	.000	-.033	-.020
	3	-.052*	.003	.000	-.060	-.043
	4	-.081*	.005	.000	-.094	-.067
3	1	.025*	.002	.000	.020	.029
	2	.052*	.003	.000	.043	.060
	4	-.029*	.002	.000	-.036	-.023
4	1	.054*	.003	.000	.045	.063
	2	.081*	.005	.000	.067	.094
	3	.029*	.002	.000	.023	.036

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Intraclass Correlation Coefficient

	Intraclass Correlation	95% Confidence Interval		F Test with True Value 0			
		Lower Bound	Upper Bound	Value	df1	df2	Sig
Single Measures	.868	.580	.965	14.195	47	48	.000
Average Measures	.930	.734	.982	14.195	47	48	.000

One-way random effects model where people effects are random.

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
DAY1.60.DF	.250	16	.069	.916	16	.148
DAY2.60.DF	.172	16	.192	.924	16	.195
DAY1.120.DF	.207	16	.066	.882	16	.061
DAY2.120.DF	.289	16	.051	.869	16	.057
DAY1.180.DF	.313	16	.067	.787	16	.072
DAY2.180.DF	.313	16	.076	.787	16	.072
DAY1.240.DF	.393	16	.078	.750	16	.081
DAY2.240.DF	.300	16	.079	.794	16	.072
DAY1.300.DF	.287	16	.061	.807	16	.063
DAY2.300.DF	.250	16	.079	.820	16	.075
DAY1.360.DF	.345	16	.080	.803	16	.063
DAY2.360.DF	.356	16	.095	.748	16	.071
DAY1.60.NEUT	.276	16	.082	.895	16	.067
DAY2.60.NEUT	.188	16	.136	.919	16	.163
DAY1.120.NEUT	.220	16	.077	.892	16	.061
DAY2.120.NEUT	.324	16	.067	.831	16	.077
DAY1.180.NEUT	.257	16	.076	.814	16	.064
DAY2.180.NEUT	.330	16	.080	.778	16	.061
DAY1.240.NEUT	.349	16	.090	.814	16	.064
DAY2.240.NEUT	.356	16	.090	.748	16	.071
DAY1.300.NEUT	.287	16	.081	.807	16	.063
DAY2.300.NEUT	.313	16	.070	.787	16	.072
DAY1.360.NEUT	.412	16	.080	.682	16	.070
DAY2.360.NEUT	.393	16	.080	.750	16	.061
DAY1.60.PF	.314	16	.080	.850	16	.074
DAY2.60.PF	.201	16	.082	.925	16	.205
DAY1.120.PF	.393	16	.070	.750	16	.061
DAY2.120.PF	.256	16	.076	.827	16	.076
DAY1.180.PF	.289	16	.081	.849	16	.083
DAY2.180.PF	.236	16	.078	.809	16	.084
DAY1.240.PF	.300	16	.060	.794	16	.092
DAY2.240.PF	.271	16	.073	.793	16	.062
DAY1.300.PF	.236	16	.068	.809	16	.064
DAY2.300.PF	.236	16	.058	.809	16	.074
DAY1.360.PF	.225	16	.071	.853	16	.085
DAY2.360.PF	.215	16	.057	.894	16	.065

Within-Subjects Factors

Measure: MEASURE_1

day	speed	SetupPosition	Dependent Variable
1	1	1	DAY1.60.DF
		2	DAY1.60.NEUT
		3	DAY1.60.PF
	2	1	DAY1.120.DF
		2	DAY1.120.NEUT
		3	DAY1.120.PF
	3	1	DAY1.180.DF
		2	DAY1.180.NEUT
		3	DAY1.180.PF
	4	1	DAY1.240.DF
		2	DAY1.240.NEUT
		3	DAY1.240.PF
	5	1	DAY1.300.DF
		2	DAY1.300.NEUT
		3	DAY1.300.PF
	6	1	DAY1.360.DF
		2	DAY1.360.NEUT
		3	DAY1.360.PF
2	1	1	DAY2.60.DF
		2	DAY2.60.NEUT
		3	DAY2.60.PF
	2	1	DAY2.120.DF
		2	DAY2.120.NEUT
		3	DAY2.120.PF
	3	1	DAY2.180.DF
		2	DAY2.180.NEUT
		3	DAY2.180.PF
	4	1	DAY2.240.DF
		2	DAY2.240.NEUT
		3	DAY2.240.PF
	5	1	DAY2.300.DF
		2	DAY2.300.NEUT
		3	DAY2.300.PF
	6	1	DAY2.360.DF
		2	DAY2.360.NEUT
		3	DAY2.360.PF

Descriptive Statistics

	Mean	Std. Deviation	N
DAY1.60.DF	.3300	.01317	16
DAY1.60.NEUT	.3269	.01401	16
DAY1.60.PF	.3925	.00775	16
DAY1.120.DF	.2781	.00911	16
DAY1.120.NEUT	.2756	.00964	16
DAY1.120.PF	.3275	.00683	16
DAY1.180.DF	.2400	.00632	16
DAY1.180.NEUT	.2413	.00719	16
DAY1.180.PF	.2744	.00727	16
DAY1.240.DF	.1825	.00683	16
DAY1.240.NEUT	.1819	.00750	16
DAY1.240.PF	.1981	.00655	16
DAY1.300.DF	.1094	.00680	16
DAY1.300.NEUT	.1094	.00680	16
DAY1.300.PF	.1581	.00750	16
DAY1.360.DF	.0706	.00772	16
DAY1.360.NEUT	.0706	.00680	16
DAY1.360.PF	.1100	.00966	16
DAY2.60.DF	.3281	.01601	16
DAY2.60.NEUT	.3300	.01265	16
DAY2.60.PF	.3913	.01025	16
DAY2.120.DF	.2763	.00885	16
DAY2.120.NEUT	.2763	.00719	16
DAY2.120.PF	.3288	.00885	16
DAY2.180.DF	.2400	.00632	16
DAY2.180.NEUT	.2413	.00619	16
DAY2.180.PF	.2681	.00750	16
DAY2.240.DF	.1819	.00655	16
DAY2.240.NEUT	.1806	.00574	16
DAY2.240.PF	.2025	.00775	16
DAY2.300.DF	.1100	.00730	16
DAY2.300.NEUT	.1100	.00632	16
DAY2.300.PF	.1619	.00750	16
DAY2.360.DF	.0706	.00574	16
DAY2.360.NEUT	.0725	.00683	16
DAY2.360.PF	.1063	.00957	16

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
day	1.000	.000	0	.	1.000	1.000	1.000
speed	.410	11.664	14	.640	.779	1.000	.200
SetupPosition	.720	4.601	2	.100	.781	.856	.500
day * speed	.297	15.896	14	.328	.697	.934	.200
day * SetupPosition	.913	1.279	2	.528	.920	1.000	.500
speed * SetupPosition	.001	76.612	54	.060	.515	.817	.100
day * speed * SetupPosition	.002	71.837	54	.084	.505	.792	.100

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept

Within Subjects Design: day + speed + SetupPosition + day * speed + day * SetupPosition + speed * SetupPosition + day * speed * SetupPosition

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
day	Pillai's Trace	.000	.004[b]	1.000	15.000	.949	.000
	Wilks' Lambda	1.000	.004[b]	1.000	15.000	.949	.000
	Hotelling's Trace	.000	.004[b]	1.000	15.000	.949	.000
	Roy's Largest Root	.000	.004[b]	1.000	15.000	.949	.000
speed	Pillai's Trace	1.000	17925.753[b]	5.000	11.000	.000	1.000
	Wilks' Lambda	.000	17925.753[b]	5.000	11.000	.000	1.000
	Hotelling's Trace	8148.069	17925.753[b]	5.000	11.000	.000	1.000
	Roy's Largest Root	8148.069	17925.753[b]	5.000	11.000	.000	1.000
SetupPosition	Pillai's Trace	.997	2028.589[b]	2.000	14.000	.000	.997
	Wilks' Lambda	.003	2028.589[b]	2.000	14.000	.000	.997
	Hotelling's Trace	289.798	2028.589[b]	2.000	14.000	.000	.997
	Roy's Largest Root	289.798	2028.589[b]	2.000	14.000	.000	.997
day * speed	Pillai's Trace	.247	.722[b]	5.000	11.000	.621	.047
	Wilks' Lambda	.753	.722[b]	5.000	11.000	.621	.047
	Hotelling's Trace	.328	.722[b]	5.000	11.000	.621	.047
	Roy's Largest Root	.328	.722[b]	5.000	11.000	.621	.047
day * SetupPosition	Pillai's Trace	.086	.662[b]	2.000	14.000	.531	.086
	Wilks' Lambda	.914	.662[b]	2.000	14.000	.531	.086
	Hotelling's Trace	.095	.662[b]	2.000	14.000	.531	.086
	Roy's Largest Root	.095	.662[b]	2.000	14.000	.531	.086
speed * SetupPosition	Pillai's Trace	.986	43.074[b]	10.000	6.000	.000	.986
	Wilks' Lambda	.014	43.074[b]	10.000	6.000	.000	.986
	Hotelling's Trace	71.789	43.074[b]	10.000	6.000	.000	.986
	Roy's Largest Root	71.789	43.074[b]	10.000	6.000	.000	.986
day * speed * SetupPosition	Pillai's Trace	.637	1.055[b]	10.000	6.000	.496	.137
	Wilks' Lambda	.363	1.055[b]	10.000	6.000	.496	.137
	Hotelling's Trace	1.759	1.055[b]	10.000	6.000	.496	.137
	Roy's Largest Root	1.759	1.055[b]	10.000	6.000	.496	.137

a. Design: Intercept

Within Subjects Design: day + speed + SetupPosition + day * speed + day * SetupPosition + speed * SetupPosition + day * speed * SetupPosition

b. Exact statistic

Pairwise Comparisons

Measure: MEASURE_1

(I) speed	(J) speed	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.056*	.001	.000	.051	.061
	3	.099*	.001	.000	.094	.104
	4	.162*	.001	.000	.157	.166
	5	.223*	.001	.000	.219	.228
	6	.266*	.001	.000	.262	.271
2	1	-.056*	.001	.000	-.061	-.051
	3	.043*	.001	.000	.039	.047
	4	.106*	.001	.000	.102	.110
	5	.167*	.001	.000	.164	.170
	6	.210*	.001	.000	.206	.214
3	1	-.099*	.001	.000	-.104	-.094
	2	-.043*	.001	.000	-.047	-.039
	4	.063*	.001	.000	.060	.066
	5	.124*	.001	.000	.122	.127
	6	.167*	.001	.000	.163	.171
4	1	-.162*	.001	.000	-.166	-.157
	2	-.106*	.001	.000	-.110	-.102
	3	-.063*	.001	.000	-.066	-.060
	5	.061*	.001	.000	.059	.064
	6	.104*	.001	.000	.100	.109
5	1	-.223*	.001	.000	-.228	-.219
	2	-.167*	.001	.000	-.170	-.164
	3	-.124*	.001	.000	-.127	-.122
	4	-.061*	.001	.000	-.064	-.059
	6	.043*	.001	.000	.040	.047
6	1	-.266*	.001	.000	-.271	-.262
	2	-.210*	.001	.000	-.214	-.206
	3	-.167*	.001	.000	-.171	-.163
	4	-.104*	.001	.000	-.109	-.100
	5	-.043*	.001	.000	-.047	-.040

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Pairwise Comparisons

Measure: MEASURE_1

(I) SetupPosition	(J) SetupPosition	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.000	.001	1.000	-.003	.003
	3	-.042*	.001	.000	-.045	-.039
2	1	.000	.001	1.000	-.003	.003
	3	-.042*	.001	.000	-.044	-.040
3	1	.042*	.001	.000	.039	.045
	2	.042*	.001	.000	.040	.044

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Intraclass Correlation Coefficient

	Intraclass Correlation	95% Confidence Interval		F Test with True Value 0			
		Lower Bound	Upper Bound	Value	df1	df2	Sig
Single Measures	.844	.739	.909	11.828	47	48	.000
Average Measures	.915	.850	.953	11.828	47	48	.000

One-way random effects model where people effects are random.

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Speed60INV	.158	16	.190*	.959	16	.646
Speed120INV	.176	16	.190*	.956	16	.583
Speed180INV	.234	16	.052	.891	16	.057
Speed240INV	.171	16	.190*	.943	16	.392
Speed300INV	.202	16	.079	.871	16	.068
Speed360INV	.153	16	.180*	.948	16	.460
Speed60EVER	.182	16	.161	.905	16	.097
Speed120EVER	.255	16	.077	.813	16	.074
Speed180EVER	.243	16	.062	.903	16	.090
Speed240EVER	.238	16	.066	.906	16	.101
Speed300EVER	.226	16	.068	.912	16	.124
Speed360EVER	.233	16	.070	.832	16	.077

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

Speed	Movement	Dependent Variable
1	1	Speed60INV
	2	Speed60EVER
2	1	Speed120INV
	2	Speed120EVER
3	1	Speed180INV
	2	Speed180EVER
4	1	Speed240INV
	2	Speed240EVER
5	1	Speed300INV
	2	Speed300EVER
6	1	Speed360INV
	2	Speed360EVER

Descriptive Statistics

	Mean	Std. Deviation	N
Speed60INV	31.6469	.32836	16
Speed60EVER	31.6125	.33838	16
Speed120INV	29.7000	.21909	16
Speed120EVER	29.6306	.21089	16
Speed180INV	27.1063	.15262	16
Speed180EVER	27.0750	.14376	16
Speed240INV	24.6375	.22767	16
Speed240EVER	24.6313	.16621	16
Speed300INV	19.6813	.25091	16
Speed300EVER	19.5213	.21793	16
Speed360INV	15.6394	.21041	16
Speed360EVER	15.6144	.22295	16

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
Speed	.178	22.574	14	.072	.639	.832	.200
Movement	1.000	.000	0	.	1.000	1.000	1.000
Speed * Movement	.110	28.919	14	.062	.524	.646	.200

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept

Within Subjects Design: Speed + Movement + Speed * Movement

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Speed	Pillai's Trace	1.000	20687.643[b]	5.000	11.000	.000	1.000
	Wilks' Lambda	.000	20687.643[b]	5.000	11.000	.000	1.000
	Hotelling's Trace	9403.474	20687.643[b]	5.000	11.000	.000	1.000
	Roy's Largest Root	9403.474	20687.643[b]	5.000	11.000	.000	1.000
Movement	Pillai's Trace	.133	2.308[b]	1.000	15.000	.149	.133
	Wilks' Lambda	.867	2.308[b]	1.000	15.000	.149	.133
	Hotelling's Trace	.154	2.308[b]	1.000	15.000	.149	.133
	Roy's Largest Root	.154	2.308[b]	1.000	15.000	.149	.133
Speed * Movement	Pillai's Trace	.317	1.019[b]	5.000	11.000	.452	.317
	Wilks' Lambda	.683	1.019[b]	5.000	11.000	.452	.317
	Hotelling's Trace	.463	1.019[b]	5.000	11.000	.452	.317
	Roy's Largest Root	.463	1.019[b]	5.000	11.000	.452	.317

a. Design: Intercept

Within Subjects Design: Speed + Movement + Speed * Movement

b. Exact statistic

Pairwise Comparisons

Measure: MEASURE_1

(I) Speed	(J) Speed	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	1.964*	.067	.000	1.732	2.197
	3	4.539*	.066	.000	4.308	4.770
	4	6.995*	.061	.000	6.782	7.209
	5	12.028*	.075	.000	11.766	12.290
	6	16.003*	.080	.000	15.723	16.283
2	1	-1.964*	.067	.000	-2.197	-1.732
	3	2.575*	.053	.000	2.390	2.759
	4	5.031*	.036	.000	4.904	5.158
	5	10.064*	.070	.000	9.821	10.307
	6	14.038*	.045	.000	13.881	14.196
3	1	-4.539*	.066	.000	-4.770	-4.308
	2	-2.575*	.053	.000	-2.759	-2.390
	4	2.456*	.038	.000	2.325	2.587
	5	7.489*	.056	.000	7.294	7.685
	6	11.464*	.046	.000	11.305	11.623
4	1	-6.995*	.061	.000	-7.209	-6.782
	2	-5.031*	.036	.000	-5.158	-4.904
	3	-2.456*	.038	.000	-2.587	-2.325
	5	5.033*	.054	.000	4.845	5.222
	6	9.008*	.042	.000	8.861	9.154
5	1	-12.028*	.075	.000	-12.290	-11.766
	2	-10.064*	.070	.000	-10.307	-9.821
	3	-7.489*	.056	.000	-7.685	-7.294
	4	-5.033*	.054	.000	-5.222	-4.845
	6	3.974*	.051	.000	3.796	4.153
6	1	-16.003*	.080	.000	-16.283	-15.723
	2	-14.038*	.045	.000	-14.196	-13.881
	3	-11.464*	.046	.000	-11.623	-11.305
	4	-9.008*	.042	.000	-9.154	-8.861
	5	-3.974*	.051	.000	-4.153	-3.796

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
LoadRange60INV	.227	16	.077	.911	16	.122
Total60INV	.273	16	.072	.833	16	.078
LoadRange120INV	.314	16	.060	.750	16	.061
Total120INV	.290	16	.091	.760	16	.061
LoadRange180INV	.260	16	.105	.880	16	.059
Total180INV	.198	16	.092	.923	16	.190
LoadRange240INV	.287	16	.091	.807	16	.073
Total240INV	.236	16	.078	.809	16	.064
LoadRange300INV	.250	16	.059	.820	16	.065
Total300INV	.287	16	.061	.807	16	.053
LoadRange360INV	.239	16	.065	.796	16	.072
Total360INV	.269	16	.063	.885	16	.057
LoadRange60EVER	.195	16	.107	.869	16	.056
Total60EVER	.298	16	.070	.846	16	.052
LoadRange120EVER	.330	16	.070	.778	16	.051
Total120EVER	.355	16	.070	.746	16	.051
LoadRange180EVER	.252	16	.088	.848	16	.073
Total180EVER	.220	16	.078	.919	16	.164
LoadRange240EVER	.300	16	.080	.794	16	.102
Total240EVER	.250	16	.089	.820	16	.115
LoadRange300EVER	.287	16	.091	.807	16	.073
Total300EVER	.287	16	.091	.807	16	.073
LoadRange360EVER	.236	16	.088	.809	16	.084
Total360EVER	.250	16	.109	.895	16	.067

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

AnalysisType	Speed	AnkleMovement	Dependent Variable
1	1	1	LoadRange60INV
		2	LoadRange60EVER
	2	1	LoadRange120INV
		2	LoadRange120EVER
	3	1	LoadRange180INV
		2	LoadRange180EVER
	4	1	LoadRange240INV
		2	LoadRange240EVER
	5	1	LoadRange300INV
		2	LoadRange300EVER
	6	1	LoadRange360INV
		2	LoadRange360EVER
2	1	1	Total60INV
		2	Total60EVER
	2	1	Total120INV
		2	Total120EVER
	3	1	Total180INV
		2	Total180EVER
	4	1	Total240INV
		2	Total240EVER
	5	1	Total300INV
		2	Total300EVER
	6	1	Total360INV
		2	Total360EVER

Descriptive Statistics

	Mean	Std. Deviation	N
LoadRange60INV	.3088	.01088	16
LoadRange60EVER	.3113	.00957	16
LoadRange120INV	.2656	.00629	16
LoadRange120EVER	.2688	.00619	16
LoadRange180INV	.2163	.01088	16
LoadRange180EVER	.2150	.01033	16
LoadRange240INV	.1594	.00680	16
LoadRange240EVER	.1581	.00655	16
LoadRange300INV	.1200	.00730	16
LoadRange300EVER	.1206	.00680	16
LoadRange360INV	.0694	.00854	16
LoadRange360EVER	.0681	.00750	16
Total60INV	.3075	.01000	16
Total60EVER	.3056	.00892	16
Total120INV	.2663	.00619	16
Total120EVER	.2675	.00577	16
Total180INV	.2169	.01078	16
Total180EVER	.2175	.01065	16
Total240INV	.1881	.00750	16
Total240EVER	.1900	.00730	16
Total300INV	.1594	.00680	16
Total300EVER	.1594	.00680	16
Total360INV	.1213	.01147	16
Total360EVER	.1200	.01155	16

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
AnalysisType	1.000	.000	0	.	1.000	1.000	1.000
Speed	.078	33.480	14	.053	.520	.639	.200
AnkleMovement	1.000	.000	0	.	1.000	1.000	1.000
AnalysisType * Speed	.243	18.521	14	.191	.692	.924	.200
AnalysisType * AnkleMovement	1.000	.000	0	.	1.000	1.000	1.000
Speed * AnkleMovement	.166	23.493	14	.056	.600	.766	.200
AnalysisType * Speed * AnkleMovement	.169	23.325	14	.059	.749	1.000	.200

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept

Within Subjects Design: AnalysisType + Speed + AnkleMovement + AnalysisType * Speed + AnalysisType * AnkleMovement + Speed * AnkleMovement + AnalysisType * Speed * AnkleMovement

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
AnalysisType	Pillai's Trace	.978	665.673[b]	1.000	15.000	.000	.978
	Wilks' Lambda	.022	665.673[b]	1.000	15.000	.000	.978
	Hotelling's Trace	44.378	665.673[b]	1.000	15.000	.000	.978
	Roy's Largest Root	44.378	665.673[b]	1.000	15.000	.000	.978
Speed	Pillai's Trace	.999	4300.617[b]	5.000	11.000	.000	.999
	Wilks' Lambda	.001	4300.617[b]	5.000	11.000	.000	.999
	Hotelling's Trace	1954.826	4300.617[b]	5.000	11.000	.000	.999
	Roy's Largest Root	1954.826	4300.617[b]	5.000	11.000	.000	.999
AnkleMovement	Pillai's Trace	.009	.137[b]	1.000	15.000	.717	.009
	Wilks' Lambda	.991	.137[b]	1.000	15.000	.717	.009
	Hotelling's Trace	.009	.137[b]	1.000	15.000	.717	.009
	Roy's Largest Root	.009	.137[b]	1.000	15.000	.717	.009
AnalysisType * Speed	Pillai's Trace	.978	97.795[b]	5.000	11.000	.000	.978
	Wilks' Lambda	.022	97.795[b]	5.000	11.000	.000	.978
	Hotelling's Trace	44.452	97.795[b]	5.000	11.000	.000	.978
	Roy's Largest Root	44.452	97.795[b]	5.000	11.000	.000	.978
AnalysisType * AnkleMovement	Pillai's Trace	.007	.110[b]	1.000	15.000	.744	.007
	Wilks' Lambda	.993	.110[b]	1.000	15.000	.744	.007
	Hotelling's Trace	.007	.110[b]	1.000	15.000	.744	.007
	Roy's Largest Root	.007	.110[b]	1.000	15.000	.744	.007
Speed * AnkleMovement	Pillai's Trace	.251	.736[b]	5.000	11.000	.612	.151
	Wilks' Lambda	.749	.736[b]	5.000	11.000	.612	.151
	Hotelling's Trace	.335	.736[b]	5.000	11.000	.612	.151
	Roy's Largest Root	.335	.736[b]	5.000	11.000	.612	.151
AnalysisType * Speed * AnkleMovement	Pillai's Trace	.250	.733[b]	5.000	11.000	.614	.140
	Wilks' Lambda	.750	.733[b]	5.000	11.000	.614	.140
	Hotelling's Trace	.333	.733[b]	5.000	11.000	.614	.140
	Roy's Largest Root	.333	.733[b]	5.000	11.000	.614	.140

a. Design: Intercept

Within Subjects Design: AnalysisType + Speed + AnkleMovement + AnalysisType * Speed + AnalysisType * AnkleMovement + Speed * AnkleMovement + AnalysisType * Speed * AnkleMovement

b. Exact statistic

Pairwise Comparisons

Measure: MEASURE_1

(I) Speed	(J) Speed	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.041*	.002	.000	.036	.047
	3	.092*	.002	.000	.083	.101
	4	.134*	.002	.000	.128	.140
	5	.168*	.002	.000	.163	.174
	6	.214*	.002	.000	.205	.222
2	1	-.041*	.002	.000	-.047	-.036
	3	.051*	.002	.000	.044	.057
	4	.093*	.001	.000	.090	.097
	5	.127*	.001	.000	.124	.131
	6	.172*	.002	.000	.166	.179
3	1	-.092*	.002	.000	-.101	-.083
	2	-.051*	.002	.000	-.057	-.044
	4	.043*	.002	.000	.037	.048
	5	.077*	.002	.000	.071	.082
	6	.122*	.002	.000	.116	.127
4	1	-.134*	.002	.000	-.140	-.128
	2	-.093*	.001	.000	-.097	-.090
	3	-.043*	.002	.000	-.048	-.037
	5	.034*	.001	.000	.031	.037
	6	.079*	.001	.000	.075	.083
5	1	-.168*	.002	.000	-.174	-.163
	2	-.127*	.001	.000	-.131	-.124
	3	-.077*	.002	.000	-.082	-.071
	4	-.034*	.001	.000	-.037	-.031
	6	.045*	.001	.000	.040	.050
6	1	-.214*	.002	.000	-.222	-.205
	2	-.172*	.002	.000	-.179	-.166
	3	-.122*	.002	.000	-.127	-.116
	4	-.079*	.001	.000	-.083	-.075
	5	-.045*	.001	.000	-.050	-.040

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Tests of Normality

	SubjectType	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
Speed60Test1	Healthy	.191	10	.190	.883	10	.140
	FAI	.209	10	.190	.908	10	.270
Speed60Test2	Healthy	.184	10	.190	.919	10	.347
	FAI	.231	10	.138	.908	10	.265
Speed90Test1	Healthy	.188	10	.190	.966	10	.856
	FAI	.321	10	.064	.833	10	.066
Speed90Test2	Healthy	.362	10	.051	.808	10	.058
	FAI	.365	10	.070	.731	10	.062
Speed120Test1	Healthy	.161	10	.190	.942	10	.572
	FAI	.246	10	.088	.910	10	.280
Speed120Test2	Healthy	.293	10	.065	.778	10	.068
	FAI	.315	10	.066	.729	10	.072
Speed180Test1	Healthy	.208	10	.190	.931	10	.457
	FAI	.175	10	.190	.958	10	.760
Speed180Test2	Healthy	.216	10	.190	.905	10	.251
	FAI	.315	10	.066	.805	10	.077

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

Speed	Day	Dependent Variable
1	1	Speed60Test1
	2	Speed60Test2
2	1	Speed90Test1
	2	Speed90Test2
3	1	Speed120Test1
	2	Speed120Test2
4	1	Speed180Test1
	2	Speed180Test2

Between-Subjects Factors

		Value Label	N
SubjectType	1.00	Healthy	10
	2.00	FAI	10

Descriptive Statistics

	SubjectType	Mean	Std. Deviation	N
Speed60Test1	Healthy	-.7310	.06226	10
	FAI	-.7240	.09606	10
	Total	-.7275	.07887	20
Speed60Test2	Healthy	-.7420	.08753	10
	FAI	-.7050	.06381	10
	Total	-.7235	.07693	20
Speed90Test1	Healthy	-.7450	.09156	10
	FAI	-.7390	.06674	10
	Total	-.7420	.07804	20
Speed90Test2	Healthy	-.7210	.05021	10
	FAI	-.7190	.05486	10
	Total	-.7200	.05120	20
Speed120Test1	Healthy	-.7520	.07714	10
	FAI	-.7220	.05554	10
	Total	-.7370	.06721	20
Speed120Test2	Healthy	-.7290	.03814	10
	FAI	-.7280	.04917	10
	Total	-.7285	.04283	20
Speed180Test1	Healthy	-.7320	.06391	10
	FAI	-.7160	.02119	10
	Total	-.7240	.04706	20
Speed180Test2	Healthy	-.7140	.02836	10
	FAI	-.6980	.05160	10
	Total	-.7060	.04135	20

Box's Test of Equality of Covariance Matrices[a]

Box's M	116.532
F	1.622
df1	36
df2	1090.213
Sig.	.062

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

a. Design: Intercept + SubjectType

Within Subjects Design: Speed + Day + Speed * Day

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
Speed	.515	11.098	5	.053	.765	.931	.333
Day	1.000	.000	0	.	1.000	1.000	1.000
Speed * Day	.601	8.514	5	.131	.732	.884	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept + SubjectType

Within Subjects Design: Speed + Day + Speed * Day

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
Speed60Test1	3.160	1	18	.092
Speed60Test2	2.474	1	18	.133
Speed90Test1	1.255	1	18	.277
Speed90Test2	.005	1	18	.947
Speed120Test1	2.043	1	18	.170
Speed120Test2	.315	1	18	.582
Speed180Test1	3.754	1	18	.069
Speed180Test2	.873	1	18	.362

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

a. Design: Intercept + SubjectType

Within Subjects Design: Speed + Day + Speed * Day

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Speed	Pillai's Trace	.392	3.441[b]	3.000	16.000	.002	.392
	Wilks' Lambda	.608	3.441[b]	3.000	16.000	.002	.392
	Hotelling's Trace	.645	3.441[b]	3.000	16.000	.002	.392
	Roy's Largest Root	.645	3.441[b]	3.000	16.000	.002	.392
Speed * SubjectType	Pillai's Trace	.048	.267[b]	3.000	16.000	.848	.048
	Wilks' Lambda	.952	.267[b]	3.000	16.000	.848	.048
	Hotelling's Trace	.050	.267[b]	3.000	16.000	.848	.048
	Roy's Largest Root	.050	.267[b]	3.000	16.000	.848	.048
Day	Pillai's Trace	.150	3.165[b]	1.000	18.000	.092	.150
	Wilks' Lambda	.850	3.165[b]	1.000	18.000	.092	.150
	Hotelling's Trace	.176	3.165[b]	1.000	18.000	.092	.150
	Roy's Largest Root	.176	3.165[b]	1.000	18.000	.092	.150
Day * SubjectType	Pillai's Trace	.000	.003[b]	1.000	18.000	.960	.000
	Wilks' Lambda	1.000	.003[b]	1.000	18.000	.960	.000
	Hotelling's Trace	.000	.003[b]	1.000	18.000	.960	.000
	Roy's Largest Root	.000	.003[b]	1.000	18.000	.960	.000
Speed * Day	Pillai's Trace	.049	.277[b]	3.000	16.000	.841	.049
	Wilks' Lambda	.951	.277[b]	3.000	16.000	.841	.049
	Hotelling's Trace	.052	.277[b]	3.000	16.000	.841	.049
	Roy's Largest Root	.052	.277[b]	3.000	16.000	.841	.049
Speed * Day * SubjectType	Pillai's Trace	.078	.452[b]	3.000	16.000	.719	.078
	Wilks' Lambda	.922	.452[b]	3.000	16.000	.719	.078
	Hotelling's Trace	.085	.452[b]	3.000	16.000	.719	.078
	Roy's Largest Root	.085	.452[b]	3.000	16.000	.719	.078

a. Design: Intercept + SubjectType
Within Subjects Design: Speed + Day + Speed * Day

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	84.347	1	84.347	5081.274	.000	.996
SubjectType	.008	1	.008	.498	.489	.027
Error	.299	18	.017			

Pairwise Comparisons

Measure: MEASURE_1

(I) Speed	(J) Speed	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.006	.010	.003	-.024	.035
	3	.007	.010	.003	-.022	.036
	4	-.011	.011	.033	-.044	.023
2	1	-.006	.010	.003	-.035	.024
	3	.002	.009	.009	-.026	.030
	4	-.016	.010	.012	-.047	.015
3	1	-.007	.010	.003	-.036	.022
	2	-.002	.009	.009	-.030	.026
	4	-.018*	.005	.027	-.034	-.002
4	1	.011	.011	.033	-.023	.044
	2	.016	.010	.012	-.015	.047
	3	.018*	.005	.027	.002	.034

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Intraclass Correlation Coefficient

	Intraclass Correlation	95% Confidence Interval		F Test with True Value 0			
		Lower Bound	Upper Bound	Value	df1	df2	Sig
Single Measures	.844	.739	.909	11.828	47	48	.000
Average Measures	.915	.850	.953	11.828	47	48	.000

One-way random effects model where people effects are random.

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
DAY1.60.NS	.272	16	.062	.787	16	.052
DAY2.60.NS	.271	16	.053	.793	16	.052
DAY1.120.NS	.236	16	.057	.808	16	.063
DAY2.120.NS	.236	16	.057	.808	16	.083
DAY1.180.NS	.271	16	.073	.793	16	.082
DAY2.180.NS	.220	16	.058	.819	16	.085
DAY1.240.NS	.433	16	.090	.472	16	.100
DAY2.240.NS	.185	16	.145	.869	16	.076
DAY1.300.NS	.522	16	.081	.289	16	.070
DAY2.300.NS	.250	16	.079	.910	16	.114
DAY1.360.NS	.220	16	.058	.819	16	.075
DAY2.360.NS	.220	16	.068	.819	16	.075
DAY1.60.SS	.233	16	.080	.885	16	.066
DAY2.60.SS	.275	16	.092	.862	16	.070
DAY1.120.SS	.308	16	.100	.768	16	.061
DAY2.120.SS	.323	16	.090	.759	16	.061
DAY1.180.SS	.257	16	.106	.814	16	.054
DAY2.180.SS	.273	16	.062	.788	16	.082
DAY1.240.SS	.161	16	.190*	.918	16	.158
DAY2.240.SS	.250	16	.059	.910	16	.114
DAY1.300.SS	.222	16	.064	.883	16	.053
DAY2.300.SS	.227	16	.067	.874	16	.062
DAY1.360.SS	.255	16	.056	.873	16	.060
DAY2.360.SS	.314	16	.060	.850	16	.064

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

Day	Speed	SetupPosition	Dependent Variable
1	1	1	DAY1.60.NS
		2	DAY1.60.SS
	2	1	DAY1.120.NS
		2	DAY1.120.SS
	3	1	DAY1.180.NS
		2	DAY1.180.SS
	4	1	DAY1.240.NS
		2	DAY1.240.SS
	5	1	DAY1.300.NS
		2	DAY1.300.SS
	6	1	DAY1.360.NS
		2	DAY1.360.SS
2	1	1	DAY2.60.NS
		2	DAY2.60.SS
	2	1	DAY2.120.NS
		2	DAY2.120.SS
	3	1	DAY2.180.NS
		2	DAY2.180.SS
	4	1	DAY2.240.NS
		2	DAY2.240.SS
	5	1	DAY2.300.NS
		2	DAY2.300.SS
	6	1	DAY2.360.NS
		2	DAY2.360.SS

Descriptive Statistics

	Mean	Std. Deviation	N
DAY1.60.NS	.9081	.00834	16
DAY1.60.SS	.9931	.01138	16
DAY1.120.NS	.7688	.00806	16
DAY1.120.SS	.8463	.00719	16
DAY1.180.NS	.5975	.00775	16
DAY1.180.SS	.6613	.00719	16
DAY1.240.NS	.4006	.06027	16
DAY1.240.SS	.4806	.01340	16
DAY1.300.NS	.4044	.58560	16
DAY1.300.SS	.2969	.00873	16
DAY1.360.NS	.1494	.00772	16
DAY1.360.SS	.1856	.00814	16
DAY2.60.NS	.9075	.00775	16
DAY2.60.SS	.9925	.01125	16
DAY2.120.NS	.7688	.00806	16
DAY2.120.SS	.8469	.00602	16
DAY2.180.NS	.5994	.00772	16
DAY2.180.SS	.6631	.00704	16
DAY2.240.NS	.4169	.01493	16
DAY2.240.SS	.4800	.01095	16
DAY2.300.NS	.2600	.01095	16
DAY2.300.SS	.2975	.00931	16
DAY2.360.NS	.1506	.00772	16
DAY2.360.SS	.1875	.00775	16

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
Day	1.000	.000	0	.	1.000	1.000	1.000
Speed	.005	237.545	14	.060	.206	.208	.200
SetupPosition	1.000	.000	0	.	1.000	1.000	1.000
Day * Speed	.004	244.653	14	.58	.206	.207	.200
Day * SetupPosition	1.000	.000	0	.	1.000	1.000	1.000
Speed * SetupPosition	.005	224.184	14	.064	.206	.207	.200
Day * Speed * SetupPosition	.005	233.860	14	.069	.205	.206	.200

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept

Within Subjects Design: Day + Speed + SetupPosition + Day * Speed + Day * SetupPosition + Speed * SetupPosition + Day * Speed * SetupPosition

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Day	Pillai's Trace	.044	.694[b]	1.000	15.000	.418	.044
	Wilks' Lambda	.956	.694[b]	1.000	15.000	.418	.044
	Hotelling's Trace	.046	.694[b]	1.000	15.000	.418	.044
	Roy's Largest Root	.046	.694[b]	1.000	15.000	.418	.044
Speed	Pillai's Trace	1.000	87841.216[b]	5.000	11.000	.000	1.000
	Wilks' Lambda	.000	87841.216[b]	5.000	11.000	.000	1.000
	Hotelling's Trace	39927.825	87841.216[b]	5.000	11.000	.000	1.000
	Roy's Largest Root	39927.825	87841.216[b]	5.000	11.000	.000	1.000
SetupPosition	Pillai's Trace	.513	15.802[b]	1.000	15.000	.001	.513
	Wilks' Lambda	.487	15.802[b]	1.000	15.000	.001	.513
	Hotelling's Trace	1.053	15.802[b]	1.000	15.000	.001	.513
	Roy's Largest Root	1.053	15.802[b]	1.000	15.000	.001	.513
Day * Speed	Pillai's Trace	.183	.492[b]	5.000	11.000	.776	.183
	Wilks' Lambda	.817	.492[b]	5.000	11.000	.776	.183
	Hotelling's Trace	.223	.492[b]	5.000	11.000	.776	.183
	Roy's Largest Root	.223	.492[b]	5.000	11.000	.776	.183
Day * SetupPosition	Pillai's Trace	.049	.777[b]	1.000	15.000	.392	.049
	Wilks' Lambda	.951	.777[b]	1.000	15.000	.392	.049
	Hotelling's Trace	.052	.777[b]	1.000	15.000	.392	.049
	Roy's Largest Root	.052	.777[b]	1.000	15.000	.392	.049
Speed * SetupPosition	Pillai's Trace	.933	30.809[b]	5.000	11.000	.000	.933
	Wilks' Lambda	.067	30.809[b]	5.000	11.000	.000	.933
	Hotelling's Trace	14.004	30.809[b]	5.000	11.000	.000	.933
	Roy's Largest Root	14.004	30.809[b]	5.000	11.000	.000	.933
Day * Speed * SetupPosition	Pillai's Trace	.181	.487[b]	5.000	11.000	.780	.181
	Wilks' Lambda	.819	.487[b]	5.000	11.000	.780	.181
	Hotelling's Trace	.221	.487[b]	5.000	11.000	.780	.181
	Roy's Largest Root	.221	.487[b]	5.000	11.000	.780	.181

a. Design: Intercept

Within Subjects Design: Day + Speed + SetupPosition + Day * Speed + Day * SetupPosition + Speed * SetupPosition + Day * Speed * SetupPosition

b. Exact statistic

Pairwise Comparisons

Measure: MEASURE_1

(I) Speed	(J) Speed	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.143*	.002	.000	.137	.148
	3	.320*	.001	.000	.316	.324
	4	.506*	.004	.000	.491	.520
	5	.636*	.037	.000	.508	.763
	6	.782*	.002	.000	.776	.788
2	1	-.143*	.002	.000	-.148	-.137
	3	.177*	.001	.000	.172	.182
	4	.363*	.004	.000	.350	.377
	5	.493*	.036	.000	.366	.620
	6	.639*	.001	.000	.635	.644
3	1	-.320*	.001	.000	-.324	-.316
	2	-.177*	.001	.000	-.182	-.172
	4	.186*	.004	.000	.171	.201
	5	.316*	.036	.000	.189	.442
	6	.462*	.001	.000	.457	.467
4	1	-.506*	.004	.000	-.520	-.491
	2	-.363*	.004	.000	-.377	-.350
	3	-.186*	.004	.000	-.201	-.171
	5	.130*	.037	.044	.002	.257
	6	.276*	.005	.000	.260	.293
5	1	-.636*	.037	.000	-.763	-.508
	2	-.493*	.036	.000	-.620	-.366
	3	-.316*	.036	.000	-.442	-.189
	4	-.130*	.037	.044	-.257	-.002
	6	.146*	.037	.017	.019	.274
6	1	-.782*	.002	.000	-.788	-.776
	2	-.639*	.001	.000	-.644	-.635
	3	-.462*	.001	.000	-.467	-.457
	4	-.276*	.005	.000	-.293	-.260
	5	-.146*	.037	.017	-.274	-.019

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Intraclass Correlation Coefficient

	Intraclass Correlation	95% Confidence Interval		F Test with True Value 0			
		Lower Bound	Upper Bound	Value	df1	df2	Sig
Single Measures	.844	.739	.909	11.828	47	48	.000
Average Measures	.915	.850	.953	11.828	47	48	.000

One-way random effects model where people effects are random.

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Speed60ABD	.184	16	.149	.812	16	.054
Speed120ABD	.224	16	.052	.896	16	.070
Speed180ABD	.179	16	.180	.915	16	.142
Speed240ABD	.159	16	.190	.950	16	.496
Speed300ABD	.207	16	.065	.925	16	.201
Speed360ABD	.130	16	.190	.935	16	.291
Speed60ADD	.187	16	.138	.920	16	.171
Speed120ADD	.159	16	.190	.931	16	.256
Speed180ADD	.137	16	.190	.948	16	.463
Speed240ADD	.135	16	.190	.955	16	.570
Speed300ADD	.204	16	.074	.954	16	.558
Speed360ADD	.175	16	.190	.968	16	.811

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

Speed	HipMovement	Dependent Variable
1	1	Speed60ABD
	2	Speed60ADD
2	1	Speed120ABD
	2	Speed120ADD
3	1	Speed180ABD
	2	Speed180ADD
4	1	Speed240ABD
	2	Speed240ADD
5	1	Speed300ABD
	2	Speed300ADD
6	1	Speed360ABD
	2	Speed360ADD

Descriptive Statistics

	Mean	Std. Deviation	N
Speed60ABD	42.1188	.70258	16
Speed60ADD	41.7813	.37098	16
Speed120ABD	39.4188	.28336	16
Speed120ADD	39.4938	.22940	16
Speed180ABD	35.9375	.46314	16
Speed180ADD	36.0063	.52974	16
Speed240ABD	32.7656	.39947	16
Speed240ADD	32.7531	.29635	16
Speed300ABD	27.5363	.24982	16
Speed300ADD	27.5288	.24905	16
Speed360ABD	21.7625	.26045	16
Speed360ADD	21.7375	.21871	16

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
Speed	.100	30.166	14	.068	.645	.843	.200
HipMovement	1.000	.000	0	.	1.000	1.000	1.000
Speed * HipMovement	.407	11.786	14	.630	.719	.973	.200

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept

Within Subjects Design: Speed + HipMovement + Speed * HipMovement

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Speed	Pillai's Trace	1.000	30376.028[b]	5.000	11.000	.000	1.000
	Wilks' Lambda	.000	30376.028[b]	5.000	11.000	.000	1.000
	Hotelling's Trace	13807.285	30376.028[b]	5.000	11.000	.000	1.000
	Roy's Largest Root	13807.285	30376.028[b]	5.000	11.000	.000	1.000
HipMovement	Pillai's Trace	.023	.358[b]	1.000	15.000	.559	.023
	Wilks' Lambda	.977	.358[b]	1.000	15.000	.559	.023
	Hotelling's Trace	.024	.358[b]	1.000	15.000	.559	.023
	Roy's Largest Root	.024	.358[b]	1.000	15.000	.559	.023
Speed * HipMovement	Pillai's Trace	.246	.716[b]	5.000	11.000	.624	.246
	Wilks' Lambda	.754	.716[b]	5.000	11.000	.624	.246
	Hotelling's Trace	.326	.716[b]	5.000	11.000	.624	.246
	Roy's Largest Root	.326	.716[b]	5.000	11.000	.624	.246

a. Design: Intercept

Within Subjects Design: Speed + HipMovement + Speed * HipMovement

b. Exact statistic

Pairwise Comparisons

Measure: MEASURE_1

(I) Speed	(J) Speed	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	2.494*	.104	.000	2.131	2.857
	3	5.978*	.125	.000	5.542	6.414
	4	9.191*	.113	.000	8.798	9.583
	5	14.418*	.113	.000	14.025	14.810
	6	20.200*	.115	.000	19.799	20.601
2	1	-2.494*	.104	.000	-2.857	-2.131
	3	3.484*	.085	.000	3.187	3.782
	4	6.697*	.085	.000	6.399	6.994
	5	11.924*	.045	.000	11.767	12.081
	6	17.706*	.042	.000	17.561	17.852
3	1	-5.978*	.125	.000	-6.414	-5.542
	2	-3.484*	.085	.000	-3.782	-3.187
	4	3.213*	.118	.000	2.802	3.623
	5	8.439*	.099	.000	8.094	8.785
	6	14.222*	.101	.000	13.869	14.575
4	1	-9.191*	.113	.000	-9.583	-8.798
	2	-6.697*	.085	.000	-6.994	-6.399
	3	-3.213*	.118	.000	-3.623	-2.802
	5	5.227*	.091	.000	4.909	5.544
	6	11.009*	.079	.000	10.733	11.286
5	1	-14.418*	.113	.000	-14.810	-14.025
	2	-11.924*	.045	.000	-12.081	-11.767
	3	-8.439*	.099	.000	-8.785	-8.094
	4	-5.227*	.091	.000	-5.544	-4.909
	6	5.783*	.044	.000	5.630	5.935
6	1	-20.200*	.115	.000	-20.601	-19.799
	2	-17.706*	.042	.000	-17.852	-17.561
	3	-14.222*	.101	.000	-14.575	-13.869
	4	-11.009*	.079	.000	-11.286	-10.733
	5	-5.783*	.044	.000	-5.935	-5.630

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
LoadRange60ABD	.160	16	.190	.927	16	.220
Total60ABD	.134	16	.190	.954	16	.554
LoadRange120ABD	.214	16	.059	.874	16	.061
Total120ABD	.183	16	.158	.899	16	.077
LoadRange180ABD	.186	16	.140	.881	16	.060
Total180ABD	.245	16	.061	.832	16	.068
LoadRange240ABD	.142	16	.190	.938	16	.331
Total240ABD	.156	16	.190	.962	16	.690
LoadRange300ABD	.229	16	.065	.879	16	.067
Total300ABD	.204	16	.073	.822	16	.055
LoadRange360ABD	.197	16	.096	.880	16	.059
Total360ABD	.197	16	.099	.879	16	.068
LoadRange60ADD	.235	16	.078	.874	16	.061
Total60ADD	.231	16	.052	.844	16	.061
LoadRange120ADD	.197	16	.096	.918	16	.154
Total120ADD	.140	16	.190	.958	16	.633
LoadRange180ADD	.228	16	.066	.908	16	.108
Total180ADD	.194	16	.109	.897	16	.072
LoadRange240ADD	.154	16	.190	.932	16	.264
Total240ADD	.141	16	.190	.930	16	.244
LoadRange300ADD	.227	16	.067	.886	16	.069
Total300ADD	.254	16	.067	.884	16	.066
LoadRange360ADD	.251	16	.088	.861	16	.060
Total360ADD	.302	16	.100	.839	16	.059

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Within-Subjects Factors

Measure: MEASURE_1

AnalysisType	Speed	HipMovement	Dependent Variable
1	1	1	LoadRange60ABD
		2	LoadRange60ADD
	2	1	LoadRange120ABD
		2	LoadRange120ADD
	3	1	LoadRange180ABD
		2	LoadRange180ADD
	4	1	LoadRange240ABD
		2	LoadRange240ADD
	5	1	LoadRange300ABD
		2	LoadRange300ADD
	6	1	LoadRange360ABD
		2	LoadRange360ADD
2	1	1	Total60ABD
		2	Total60ADD
	2	1	Total120ABD
		2	Total120ADD
	3	1	Total180ABD
		2	Total180ADD
	4	1	Total240ABD
		2	Total240ADD
	5	1	Total300ABD
		2	Total300ADD
	6	1	Total360ABD
		2	Total360ADD

Descriptive Statistics

	Mean	Std. Deviation	N
LoadRange60ABD	1.1150	.06175	16
LoadRange60ADD	1.0988	.06561	16
LoadRange120ABD	.9113	.01893	16
LoadRange120ADD	.9125	.01770	16
LoadRange180ABD	.7856	.01365	16
LoadRange180ADD	.7794	.01124	16
LoadRange240ABD	.6225	.02176	16
LoadRange240ADD	.5981	.03351	16
LoadRange300ABD	.4063	.01025	16
LoadRange300ADD	.4044	.01031	16
LoadRange360ABD	.1863	.01088	16
LoadRange360ADD	.1856	.01365	16
Total60ABD	1.1200	.05854	16
Total60ADD	1.1088	.07173	16
Total120ABD	.9188	.02156	16
Total120ADD	.9175	.01732	16
Total180ABD	.7869	.02243	16
Total180ADD	.7863	.01857	16
Total240ABD	.6225	.03531	16
Total240ADD	.6263	.01893	16
Total300ABD	.5344	.01263	16
Total300ADD	.5331	.00946	16
Total360ABD	.3844	.01094	16
Total360ADD	.3813	.00957	16

Mauchly's Test of Sphericity[a]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[b]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
AnalysisType	1.000	.000	0	.	1.000	1.000	1.000
Speed	.007	64.128	14	.052	.302	.328	.200
HipMovement	1.000	.000	0	.	1.000	1.000	1.000
AnalysisType * Speed	.008	62.927	14	.060	.326	.360	.200
AnalysisType * HipMovement	1.000	.000	0	.	1.000	1.000	1.000
Speed * HipMovement	.019	51.883	14	.062	.376	.428	.200
AnalysisType * Speed * HipMovement	.008	67.151	14	.065	.330	.366	.200

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. Design: Intercept

Within Subjects Design: AnalysisType + Speed + HipMovement + AnalysisType * Speed + AnalysisType * HipMovement + Speed * HipMovement + AnalysisType * Speed * HipMovement

b. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
AnalysisType	Pillai's Trace	.958	339.285[b]	1.000	15.000	.000	.958
	Wilks' Lambda	.042	339.285[b]	1.000	15.000	.000	.958
	Hotelling's Trace	22.619	339.285[b]	1.000	15.000	.000	.958
	Roy's Largest Root	22.619	339.285[b]	1.000	15.000	.000	.958
Speed	Pillai's Trace	1.000	8871.059[b]	5.000	11.000	.000	1.000
	Wilks' Lambda	.000	8871.059[b]	5.000	11.000	.000	1.000
	Hotelling's Trace	4032.299	8871.059[b]	5.000	11.000	.000	1.000
	Roy's Largest Root	4032.299	8871.059[b]	5.000	11.000	.000	1.000
HipMovement	Pillai's Trace	.069	1.108[b]	1.000	15.000	.309	.069
	Wilks' Lambda	.931	1.108[b]	1.000	15.000	.309	.069
	Hotelling's Trace	.074	1.108[b]	1.000	15.000	.309	.069
	Roy's Largest Root	.074	1.108[b]	1.000	15.000	.309	.069
AnalysisType * Speed	Pillai's Trace	.996	535.747[b]	5.000	11.000	.000	.996
	Wilks' Lambda	.004	535.747[b]	5.000	11.000	.000	.996
	Hotelling's Trace	243.521	535.747[b]	5.000	11.000	.000	.996
	Roy's Largest Root	243.521	535.747[b]	5.000	11.000	.000	.996
AnalysisType * HipMovement	Pillai's Trace	.070	1.121[b]	1.000	15.000	.306	.070
	Wilks' Lambda	.930	1.121[b]	1.000	15.000	.306	.070
	Hotelling's Trace	.075	1.121[b]	1.000	15.000	.306	.070
	Roy's Largest Root	.075	1.121[b]	1.000	15.000	.306	.070
Speed * HipMovement	Pillai's Trace	.169	.448[b]	5.000	11.000	.806	.169
	Wilks' Lambda	.831	.448[b]	5.000	11.000	.806	.169
	Hotelling's Trace	.204	.448[b]	5.000	11.000	.806	.169
	Roy's Largest Root	.204	.448[b]	5.000	11.000	.806	.169
AnalysisType * Speed * HipMovement	Pillai's Trace	.401	1.472[b]	5.000	11.000	.275	.401
	Wilks' Lambda	.599	1.472[b]	5.000	11.000	.275	.401
	Hotelling's Trace	.669	1.472[b]	5.000	11.000	.275	.401
	Roy's Largest Root	.669	1.472[b]	5.000	11.000	.275	.401

a. Design: Intercept

Within Subjects Design: AnalysisType + Speed + HipMovement + AnalysisType * Speed + AnalysisType * HipMovement + Speed * HipMovement + AnalysisType * Speed * HipMovement

b. Exact statistic

Pairwise Comparisons

Measure: MEASURE_1

(I) Speed	(J) Speed	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
	2	.196*	.008	.000	.167	.225
	3	.326*	.010	.000	.293	.360
1	4	.493*	.010	.000	.457	.530
	5	.641*	.009	.000	.611	.672
	6	.826*	.009	.000	.794	.858
	1	-.196*	.008	.000	-.225	-.167
	3	.130*	.003	.000	.121	.140
2	4	.298*	.003	.000	.286	.309
	5	.445*	.003	.000	.436	.455
	6	.631*	.003	.000	.621	.641
	1	-.326*	.010	.000	-.360	-.293
	2	-.130*	.003	.000	-.140	-.121
3	4	.167*	.004	.000	.154	.180
	5	.315*	.003	.000	.304	.326
	6	.500*	.003	.000	.491	.509
	1	-.493*	.010	.000	-.530	-.457
	2	-.298*	.003	.000	-.309	-.286
4	3	-.167*	.004	.000	-.180	-.154
	5	.148*	.004	.000	.135	.161
	6	.333*	.004	.000	.320	.346
	1	-.641*	.009	.000	-.672	-.611
	2	-.445*	.003	.000	-.455	-.436
5	3	-.315*	.003	.000	-.326	-.304
	4	-.148*	.004	.000	-.161	-.135
	6	.185*	.002	.000	.177	.193
	1	-.826*	.009	.000	-.858	-.794
	2	-.631*	.003	.000	-.641	-.621
6	3	-.500*	.003	.000	-.509	-.491
	4	-.333*	.004	.000	-.346	-.320
	5	-.185*	.002	.000	-.193	-.177

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

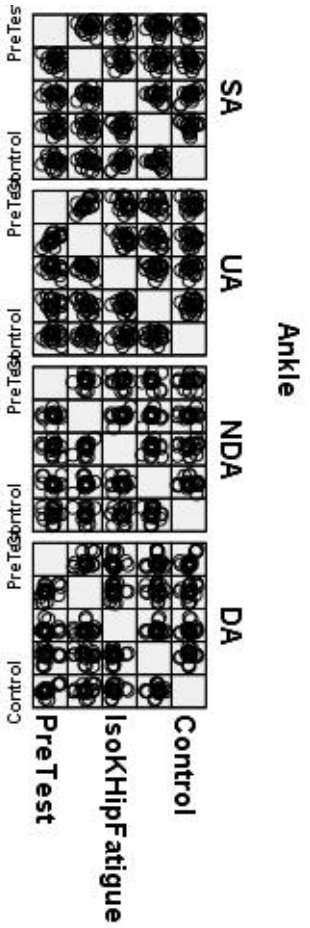
Tests of Normality

	SubjectCondition	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
PreTest	DA	.261	20	.061	.781	20	.100
	NDA	.223	20	.061	.890	20	.057
	UA	.152	20	.200*	.931	20	.159
	SA	.127	20	.200*	.935	20	.189
IsoKAnkleFatigue	DA	.147	20	.200*	.920	20	.100
	NDA	.229	20	.077	.803	20	.061
	UA	.099	20	.200*	.976	20	.876
	SA	.116	20	.200*	.955	20	.448
IsoKHipFatigue	DA	.270	20	.061	.731	20	.070
	NDA	.158	20	.200*	.912	20	.070
	UA	.186	20	.069	.914	20	.077
	SA	.184	20	.074	.908	20	.057
FootballFatigue	DA	.209	20	.052	.883	20	.060
	NDA	.207	20	.055	.881	20	.068
	UA	.103	20	.200*	.949	20	.352
	SA	.131	20	.200*	.958	20	.503
Control	DA	.178	20	.096	.908	20	.059
	NDA	.128	20	.200*	.920	20	.100
	UA	.122	20	.200*	.962	20	.583
	SA	.141	20	.200*	.970	20	.761

Residuals Statistics[a]

	Minimum	Maximum	Mean	Std. Deviation	N
Predicted Value	1.2907	3.7472	2.5000	1.00863	80
Std. Predicted Value	-1.199	1.237	.000	1.000	80
Standard Error of Predicted Value	.085	.233	.137	.034	80
Adjusted Predicted Value	1.2701	3.9392	2.4995	1.01238	80
Residual	-.74719	.71794	.00000	.49849	80
Std. Residual	-1.451	1.394	.000	.968	80
Stud. Residual	-1.626	1.463	.000	1.007	80
Deleted Residual	-.93923	.79088	.00048	.53988	80
Stud. Deleted Residual	-1.645	1.475	.000	1.009	80
Mahal. Distance	1.143	15.166	4.938	3.039	80
Cook's Distance	.003	.113	.014	.015	80
Centered Leverage Value	.014	.192	.063	.038	80

a. Dependent Variable: Ankle



Between-Subjects Factors

		Value Label	N
Ankle	1.00	DA	20
	2.00	NDA	20
	3.00	UA	20
	4.00	SA	20

Descriptive Statistics

	Ankle	Mean	Std. Deviation	N
PreTest	DA	48.4690	.31114	20
	NDA	48.1825	.42232	20
	UA	54.5760	.59687	20
	SA	54.7385	.53274	20
	Total	51.4915	3.22222	80
IsoKAnkleFatigue	DA	48.3235	.52868	20
	NDA	48.3385	.29068	20
	UA	54.6426	.67214	20
	SA	54.6631	.55223	20
	Total	51.4919	3.22297	80
IsoKHipFatigue	DA	48.3610	.22342	20
	NDA	48.1360	.44144	20
	UA	54.7880	.51693	20
	SA	54.7491	.59174	20
	Total	51.5085	3.31304	80
FootballFatigue	DA	48.2190	.43192	20
	NDA	48.3655	.36635	20
	UA	54.6316	.73140	20
	SA	54.6496	.53316	20
	Total	51.4664	3.23726	80
Control	DA	48.3425	.45928	20
	NDA	48.3105	.52844	20
	UA	54.5441	.48163	20
	SA	54.6716	.62139	20
	Total	51.4672	3.20272	80

Box's Test of Equality of Covariance Matrices[a]

Box's M	91.561
F	1.790
df1	45
df2	14297.290
Sig.	.003

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
PreTest	2.620	3	76	.057
IsoKAnkleFatigue	5.090	3	76	.073
IsoKHipFatigue	3.594	3	76	.067
FootballFatigue	4.161	3	76	.059
Control	.589	3	76	.624

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Intercept	Pillai's Trace	1.000	834891.903[b]	5.000	72.000	.000	1.000
	Wilks' Lambda	.000	834891.903[b]	5.000	72.000	.000	1.000
	Hotelling's Trace	57978.604	834891.903[b]	5.000	72.000	.000	1.000
	Roy's Largest Root	57978.604	834891.903[b]	5.000	72.000	.000	1.000
Ankle	Pillai's Trace	1.106	8.643	15.000	222.000	.000	.369
	Wilks' Lambda	.004	84.976	15.000	199.162	.000	.841
	Hotelling's Trace	222.750	1049.399	15.000	212.000	.000	.987
	Roy's Largest Root	222.630	3294.917[c]	5.000	74.000	.000	.996

a. Design: Intercept + Ankle

b. Exact statistic

c. The statistic is an upper bound on F that yields a lower bound on the significance level.

Tests of Between-Subjects Effects

Source	Dependent Variable	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Corrected Model	PreTest	802.843[a]	3	267.614	1169.596	.000	.979
	IsoKAnkleFatigue	799.322[b]	3	266.441	950.952	.000	.974
	IsoKHipFatigue	850.742[c]	3	283.581	1315.688	.000	.981
	FootballFatigue	806.249[d]	3	268.750	942.992	.000	.974
	Control	789.280[e]	3	263.093	949.552	.000	.974
Intercept	PreTest	212109.966	1	212109.966	927017.262	.000	1.000
	IsoKAnkleFatigue	212113.467	1	212113.467	757052.505	.000	1.000
	IsoKHipFatigue	212250.252	1	212250.252	984746.330	.000	1.000
	FootballFatigue	211903.432	1	211903.432	743529.445	.000	1.000
	Control	211909.608	1	211909.608	764820.795	.000	1.000
Ankle	PreTest	802.843	3	267.614	1169.596	.000	.979
	IsoKAnkleFatigue	799.322	3	266.441	950.952	.000	.974
	IsoKHipFatigue	850.742	3	283.581	1315.688	.000	.981
	FootballFatigue	806.249	3	268.750	942.992	.000	.974
	Control	789.280	3	263.093	949.552	.000	.974
Error	PreTest	17.389	76	.229			
	IsoKAnkleFatigue	21.294	76	.280			
	IsoKHipFatigue	16.381	76	.216			
	FootballFatigue	21.660	76	.285			
	Control	21.057	76	.277			
Total	PreTest	212930.198	80				
	IsoKAnkleFatigue	212934.083	80				
	IsoKHipFatigue	213117.375	80				
	FootballFatigue	212731.341	80				
	Control	212719.946	80				
Corrected Total	PreTest	820.232	79				
	IsoKAnkleFatigue	820.616	79				
	IsoKHipFatigue	867.123	79				
	FootballFatigue	827.909	79				
	Control	810.337	79				

Multiple Comparisons

Tukey HSD

Dependent Variable	(I) Ankle	(J) Ankle	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
PreTest	DA	NDA	.2865	.15126	.239	-.1108	.6838
		UA	-6.1070*	.15126	.000	-6.5043	-5.7097
		SA	-6.2695*	.15126	.000	-6.6668	-5.8722
	NDA	DA	-.2865	.15126	.239	-.6838	.1108
		UA	-6.3935*	.15126	.000	-6.7908	-5.9962
		SA	-6.5560*	.15126	.000	-6.9533	-6.1587
	UA	DA	6.1070*	.15126	.000	5.7097	6.5043
		NDA	6.3935*	.15126	.000	5.9962	6.7908
		SA	-.1625	.15126	.706	-.5598	.2348
	SA	DA	6.2695*	.15126	.000	5.8722	6.6668
		NDA	6.5560*	.15126	.000	6.1587	6.9533
		UA	.1625	.15126	.706	-.2348	.5598
IsoKAnkleFatigue	DA	NDA	-.0150	.16739	1.000	-.4547	.4247
		UA	-6.3191*	.16739	.000	-6.7588	-5.8794
		SA	-6.3396*	.16739	.000	-6.7793	-5.8999
	NDA	DA	.0150	.16739	1.000	-.4247	.4547
		UA	-6.3041*	.16739	.000	-6.7438	-5.8644
		SA	-6.3246*	.16739	.000	-6.7643	-5.8849
	UA	DA	6.3191*	.16739	.000	5.8794	6.7588
		NDA	6.3041*	.16739	.000	5.8644	6.7438
		SA	-.0205	.16739	.999	-.4602	.4192
	SA	DA	6.3396*	.16739	.000	5.8999	6.7793
		NDA	6.3246*	.16739	.000	5.8849	6.7643
		UA	.0205	.16739	.999	-.4192	.4602
IsoKHipFatigue	DA	NDA	.2250	.14681	.423	-.1606	.6106
		UA	-6.4270*	.14681	.000	-6.8126	-6.0414
		SA	-6.3881*	.14681	.000	-6.7737	-6.0025
	NDA	DA	-.2250	.14681	.423	-.6106	.1606
		UA	-6.6520*	.14681	.000	-7.0376	-6.2664
		SA	-6.6131*	.14681	.000	-6.9987	-6.2275
	UA	DA	6.4270*	.14681	.000	6.0414	6.8126
		NDA	6.6520*	.14681	.000	6.2664	7.0376
		SA	.0389	.14681	.993	-.3467	.4245
	SA	DA	6.3881*	.14681	.000	6.0025	6.7737
		NDA	6.6131*	.14681	.000	6.2275	6.9987
		UA	-.0389	.14681	.993	-.4245	.3467
FootballFatigue	DA	NDA	-.1465	.16882	.821	-.5900	.2970
		UA	-6.4126*	.16882	.000	-6.8561	-5.9691

		SA	-6.4306*	.16882	.000	-6.8741	-5.9871
		DA	.1465	.16882	.821	-.2970	.5900
	NDA	UA	-6.2661*	.16882	.000	-6.7096	-5.8226
		SA	-6.2841*	.16882	.000	-6.7276	-5.8406
		DA	6.4126*	.16882	.000	5.9691	6.8561
	UA	NDA	6.2661*	.16882	.000	5.8226	6.7096
		SA	-.0180	.16882	1.000	-.4615	.4255
		DA	6.4306*	.16882	.000	5.9871	6.8741
	SA	NDA	6.2841*	.16882	.000	5.8406	6.7276
		UA	.0180	.16882	1.000	-.4255	.4615
		NDA	.0320	.16645	.997	-.4052	.4692
	DA	UA	-6.2016*	.16645	.000	-6.6388	-5.7644
		SA	-6.3291*	.16645	.000	-6.7663	-5.8919
		DA	-.0320	.16645	.997	-.4692	.4052
	NDA	UA	-6.2336*	.16645	.000	-6.6708	-5.7964
		SA	-6.3611*	.16645	.000	-6.7983	-5.9239
Control		DA	6.2016*	.16645	.000	5.7644	6.6388
	UA	NDA	6.2336*	.16645	.000	5.7964	6.6708
		SA	-.1275	.16645	.869	-.5647	.3097
		DA	6.3291*	.16645	.000	5.8919	6.7663
	SA	NDA	6.3611*	.16645	.000	5.9239	6.7983
		UA	.1275	.16645	.869	-.3097	.5647

Based on observed means.

The error term is Mean Square(Error) = .277.

*. The mean difference is significant at the .05 level.

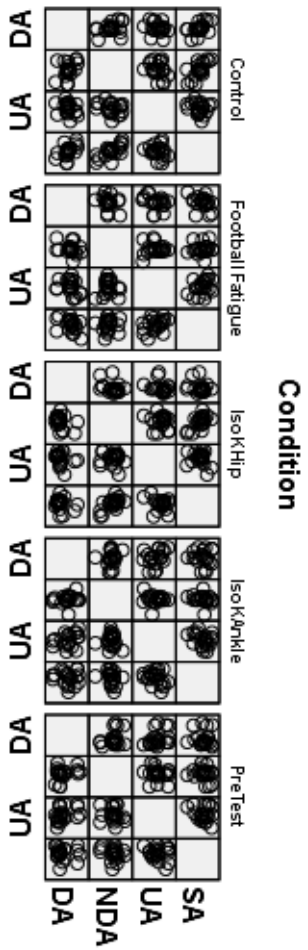
Tests of Normality

	Condition	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
DA	PreTest	.261	20	.081	.781	20	.090
	IsoKAnkle	.147	20	.200*	.920	20	.100
	IsoKHip	.270	20	.091	.731	20	.080
	FootballFatigue	.209	20	.072	.883	20	.098
	Control	.178	20	.096	.908	20	.059
NDA	PreTest	.223	20	.061	.890	20	.067
	IsoKAnkle	.229	20	.067	.803	20	.061
	IsoKHip	.158	20	.200*	.912	20	.070
	FootballFatigue	.207	20	.075	.881	20	.078
	Control	.128	20	.200*	.920	20	.100
UA	PreTest	.152	20	.200*	.931	20	.159
	IsoKAnkle	.099	20	.200*	.976	20	.876
	IsoKHip	.186	20	.069	.914	20	.077
	FootballFatigue	.103	20	.200*	.949	20	.352
	Control	.122	20	.200*	.962	20	.583
SA	PreTest	.127	20	.200*	.935	20	.189
	IsoKAnkle	.116	20	.200*	.955	20	.448
	IsoKHip	.184	20	.074	.908	20	.057
	FootballFatigue	.131	20	.200*	.958	20	.503
	Control	.141	20	.200*	.970	20	.761

Residuals Statistics[a]

	Minimum	Maximum	Mean	Std. Deviation	N
Predicted Value	2.4870	3.7274	3.0000	.24796	100
Std. Predicted Value	-2.069	2.934	.000	1.000	100
Standard Error of Predicted Value	.149	.502	.304	.098	100
Adjusted Predicted Value	2.4354	3.6891	2.9979	.25902	100
Residual	-2.35616	2.31080	.00000	1.39954	100
Std. Residual	-1.649	1.617	.000	.980	100
Stud. Residual	-1.692	1.671	.001	1.005	100
Deleted Residual	-2.47914	2.47810	.00209	1.47388	100
Stud. Deleted Residual	-1.709	1.687	.001	1.009	100
Mahal. Distance	.080	11.231	3.960	3.029	100
Cook's Distance	.000	.063	.011	.013	100
Centered Leverage Value	.001	.113	.040	.031	100

a. Dependent Variable: Condition


Control
Football Fatigue
IsoK Hip
IsoK Ankle
PreTest
UA
DA
SA
NDA
Condition

Descriptive Statistics

	Condition	Mean	Std. Deviation	N
DA	PreTest	48.4690	.31114	20
	IsoKAnkle	48.3235	.52868	20
	IsoKHip	48.3610	.22342	20
	FootballFatigue	48.2190	.43192	20
	Control	48.3425	.45928	20
	Total	48.3430	.40564	100
NDA	PreTest	48.1825	.42232	20
	IsoKAnkle	48.3385	.29068	20
	IsoKHip	48.1360	.44144	20
	FootballFatigue	48.3655	.36635	20
	Control	48.3105	.52844	20
	Total	48.2666	.41890	100
UA	PreTest	54.5760	.59687	20
	IsoKAnkle	54.6426	.67214	20
	IsoKHip	54.7880	.51693	20
	FootballFatigue	54.6316	.73140	20
	Control	54.5441	.48163	20
	Total	54.6365	.60055	100
SA	PreTest	54.7385	.53274	20
	IsoKAnkle	54.6631	.55223	20
	IsoKHip	54.7491	.59174	20
	FootballFatigue	54.6496	.53316	20
	Control	54.6716	.62139	20
	Total	54.6944	.55728	100

Box's Test of Equality of Covariance Matrices[a]

Box's M	80.474
F	1.826
df1	40
df2	19908.088
Sig.	.011

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

a. Design: Intercept + Condition

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
DA	2.197	4	95	.075
NDA	1.881	4	95	.120
UA	1.280	4	95	.283
SA	.062	4	95	.993

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

a. Design: Intercept + Condition

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Intercept	Pillai's Trace	1.000	1023993.217[b]	4.000	92.000	.000	1.000
	Wilks' Lambda	.000	1023993.217[b]	4.000	92.000	.000	1.000
	Hotelling's Trace	44521.444	1023993.217[b]	4.000	92.000	.000	1.000
	Roy's Largest Root	44521.444	1023993.217[b]	4.000	92.000	.000	1.000
Condition	Pillai's Trace	.125	.768	16.000	380.000	.721	.031
	Wilks' Lambda	.877	.770	16.000	281.702	.720	.032
	Hotelling's Trace	.136	.772	16.000	362.000	.718	.033
	Roy's Largest Root	.107	2.534[c]	4.000	95.000	.045	.096

a. Design: Intercept + Condition

b. Exact statistic

c. The statistic is an upper bound on F that yields a lower bound on the significance level.

Tests of Between-Subjects Effects

Source	Dependent Variable	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Corrected Model	DA	.639[a]	4	.160	.970	.428	.039
	NDA	.820[b]	4	.205	1.177	.326	.047
	UA	.704[c]	4	.176	.478	.752	.020
	SA	.169[d]	4	.042	.131	.971	.005
Intercept	DA	233704.565	1	233704.565	1418602.240	.000	1.000
	NDA	232966.468	1	232966.468	1337067.779	.000	1.000
	UA	298514.276	1	298514.276	810225.038	.000	1.000
	SA	299147.520	1	299147.520	929419.436	.000	1.000
Condition	DA	.639	4	.160	.970	.428	.039
	NDA	.820	4	.205	1.177	.326	.047
	UA	.704	4	.176	.478	.752	.020
	SA	.169	4	.042	.131	.971	.005
Error	DA	15.651	95	.165			
	NDA	16.553	95	.174			
	UA	35.001	95	.368			
	SA	30.577	95	.322			
Total	DA	233720.855	100				
	NDA	232983.840	100				
	UA	298549.982	100				
	SA	299178.266	100				
Corrected Total	DA	16.290	99				
	NDA	17.373	99				
	UA	35.705	99				
	SA	30.746	99				

a. R Squared = .039 (Adjusted R Squared = -.001)

b. R Squared = .047 (Adjusted R Squared = .007)

c. R Squared = .020 (Adjusted R Squared = -.022)

d. R Squared = .005 (Adjusted R Squared = -.036)

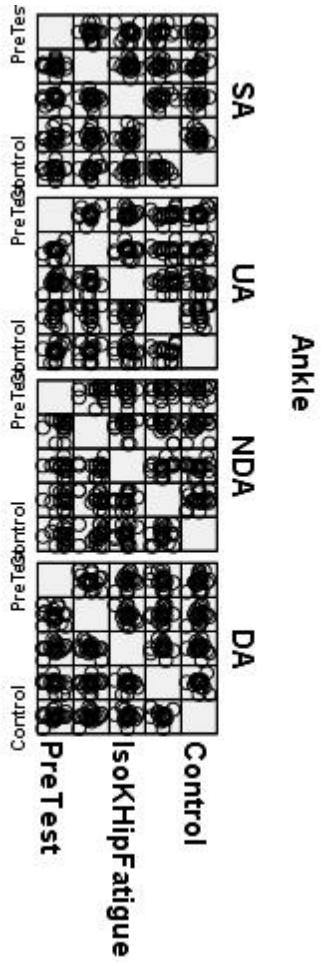
Tests of Normality

	Ankle	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
PreTest	DA	.177	20	.100	.928	20	.142
	NDA	.315	20	.070	.798	20	.061
	UA	.211	20	.060	.903	20	.066
	SA	.243	20	.053	.879	20	.067
IsoKAnkleFatigue	DA	.173	20	.117	.948	20	.345
	NDA	.262	20	.051	.752	20	.070
	UA	.261	20	.051	.838	20	.063
	SA	.180	20	.088	.928	20	.140
IsoKHipFatigue	DA	.211	20	.060	.903	20	.056
	NDA	.291	20	.100	.840	20	.064
	UA	.210	20	.061	.895	20	.063
	SA	.218	20	.063	.890	20	.066
FootballFatigue	DA	.330	20	.070	.803	20	.061
	NDA	.210	20	.061	.907	20	.056
	UA	.263	20	.061	.887	20	.064
	SA	.247	20	.052	.898	20	.078
Control	DA	.179	20	.094	.915	20	.079
	NDA	.133	20	.200*	.932	20	.169
	UA	.198	20	.068	.904	20	.059
	SA	.280	20	.070	.888	20	.064

Residuals Statistics[a]

	Minimum	Maximum	Mean	Std. Deviation	N
Predicted Value	2.0005	3.0474	2.5000	.19823	80
Std. Predicted Value	-2.520	2.761	.000	1.000	80
Standard Error of Predicted Value	.142	.518	.300	.091	80
Adjusted Predicted Value	1.9897	3.1122	2.5053	.21814	80
Residual	-1.75690	1.79550	.00000	1.10749	80
Std. Residual	-1.535	1.569	.000	.968	80
Stud. Residual	-1.681	1.636	-.002	1.005	80
Deleted Residual	-2.11225	1.95205	-.00532	1.19438	80
Stud. Deleted Residual	-1.702	1.655	-.003	1.009	80
Mahal. Distance	.221	15.200	4.938	3.400	80
Cook's Distance	.000	.097	.013	.015	80
Centered Leverage Value	.003	.192	.063	.043	80

a. Dependent Variable: Ankle



Between-Subjects Factors

		Value Label	N
Ankle	1.00	DA	20
	2.00	NDA	20
	3.00	UA	20
	4.00	SA	20

Descriptive Statistics

	Ankle	Mean	Std. Deviation	N
PreTest	DA	3.3450	.02724	20
	NDA	3.3455	.01538	20
	UA	3.3505	.01849	20
	SA	3.3440	.02563	20
	Total	3.3463	.02195	80
IsoKAnkleFatigue	DA	3.3450	.02585	20
	NDA	3.3450	.01762	20
	UA	3.3420	.02042	20
	SA	3.3510	.01889	20
	Total	3.3458	.02079	80
IsoKHipFatigue	DA	3.3505	.01849	20
	NDA	3.3420	.02215	20
	UA	3.3515	.02323	20
	SA	3.3455	.01638	20
	Total	3.3474	.02024	80
FootballFatigue	DA	3.3460	.01984	20
	NDA	3.3495	.02645	20
	UA	3.3400	.01777	20
	SA	3.3535	.01424	20
	Total	3.3473	.02031	80
Control	DA	3.3510	.01861	20
	NDA	3.3450	.02965	20
	UA	3.3470	.02716	20
	SA	3.3430	.02105	20
	Total	3.3465	.02424	80

Box's Test of Equality of Covariance Matrices[a]

Box's M	71.545
F	1.399
df1	45
df2	14297.290
Sig.	.040

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

a. Design: Intercept + Ankle

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
PreTest	2.049	3	76	.114
IsoKAnkleFatigue	1.026	3	76	.386
IsoKHipFatigue	.686	3	76	.563
FootballFatigue	1.293	3	76	.283
Control	1.911	3	76	.135

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

a. Design: Intercept + Ankle

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Intercept	Pillai's Trace	1.000	1847765.491[b]	5.000	72.000	.000	1.000
	Wilks' Lambda	.000	1847765.491[b]	5.000	72.000	.000	1.000
	Hotelling's Trace	128317.048	1847765.491[b]	5.000	72.000	.000	1.000
	Roy's Largest Root	128317.048	1847765.491[b]	5.000	72.000	.000	1.000
Ankle	Pillai's Trace	.168	.876	15.000	222.000	.592	.056
	Wilks' Lambda	.838	.880	15.000	199.162	.587	.057
	Hotelling's Trace	.188	.884	15.000	212.000	.583	.059
	Roy's Largest Root	.148	2.189[c]	5.000	74.000	.064	.129

a. Design: Intercept + Ankle

b. Exact statistic

c. The statistic is an upper bound on F that yields a lower bound on the significance level.

Tests of Between-Subjects Effects

Source	Dependent Variable	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Corrected Model	PreTest	.001[a]	3	.000	.341	.796	.013
	IsoKAnkleFatigue	.001[b]	3	.000	.650	.585	.025
	IsoKHipFatigue	.001[c]	3	.000	.962	.415	.037
	FootballFatigue	.002[d]	3	.001	1.625	.191	.060
	Control	.001[e]	3	.000	.388	.762	.015
Intercept	PreTest	895.791	1	895.791	1812087.450	.000	1.000
	IsoKAnkleFatigue	895.523	1	895.523	2043837.292	.000	1.000
	IsoKHipFatigue	896.394	1	896.394	2185974.969	.000	1.000
	FootballFatigue	896.327	1	896.327	2223990.270	.000	1.000
	Control	895.925	1	895.925	1489289.118	.000	1.000
Ankle	PreTest	.001	3	.000	.341	.796	.013
	IsoKAnkleFatigue	.001	3	.000	.650	.585	.025
	IsoKHipFatigue	.001	3	.000	.962	.415	.037
	FootballFatigue	.002	3	.001	1.625	.191	.060
	Control	.001	3	.000	.388	.762	.015
Error	PreTest	.038	76	.000			
	IsoKAnkleFatigue	.033	76	.000			
	IsoKHipFatigue	.031	76	.000			
	FootballFatigue	.031	76	.000			
	Control	.046	76	.001			
Total	PreTest	895.829	80				
	IsoKAnkleFatigue	895.558	80				
	IsoKHipFatigue	896.426	80				
	FootballFatigue	896.359	80				
	Control	895.971	80				
Corrected Total	PreTest	.038	79				
	IsoKAnkleFatigue	.034	79				
	IsoKHipFatigue	.032	79				
	FootballFatigue	.033	79				
	Control	.046	79				

a. R Squared = .013 (Adjusted R Squared = -.026)

b. R Squared = .025 (Adjusted R Squared = -.013)

c. R Squared = .037 (Adjusted R Squared = -.001)

d. R Squared = .060 (Adjusted R Squared = .023)

e. R Squared = .015 (Adjusted R Squared = -.024)

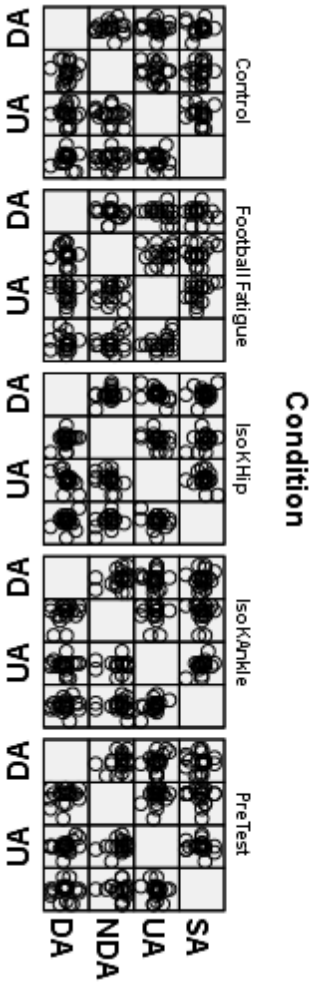
Tests of Normality

	Condition	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
DA	PreTest	.177	20	.100	.928	20	.142
	IsoKAnkle	.173	20	.117	.948	20	.345
	IsoKHip	.211	20	.060	.903	20	.076
	FootballFatigue	.330	20	.098	.803	20	.081
	Control	.179	20	.094	.915	20	.079
NDA	PreTest	.315	20	.070	.798	20	.091
	IsoKAnkle	.262	20	.061	.752	20	.120
	IsoKHip	.291	20	.070	.840	20	.064
	FootballFatigue	.210	20	.081	.907	20	.056
	Control	.133	20	.200*	.932	20	.169
UA	PreTest	.211	20	.090	.903	20	.076
	IsoKAnkle	.261	20	.051	.838	20	.083
	IsoKHip	.210	20	.061	.895	20	.073
	FootballFatigue	.263	20	.081	.887	20	.064
	Control	.198	20	.098	.904	20	.079
SA	PreTest	.243	20	.083	.879	20	.087
	IsoKAnkle	.180	20	.188	.928	20	.140
	IsoKHip	.218	20	.113	.890	20	.056
	FootballFatigue	.247	20	.122	.898	20	.068
	Control	.280	20	.090	.888	20	.074

Residuals Statistics[a]

	Minimum	Maximum	Mean	Std. Deviation	N
Predicted Value	2.6263	3.3705	3.0000	.14062	100
Std. Predicted Value	-2.657	2.635	.000	1.000	100
Standard Error of Predicted Value	.158	.509	.309	.094	100
Adjusted Predicted Value	2.5444	3.4172	2.9981	.17364	100
Residual	-2.21044	2.24531	.00000	1.41436	100
Std. Residual	-1.531	1.555	.000	.980	100
Stud. Residual	-1.573	1.626	.001	1.008	100
Deleted Residual	-2.33462	2.45561	.00189	1.49936	100
Stud. Deleted Residual	-1.586	1.641	.001	1.012	100
Mahal. Distance	.194	11.326	3.960	2.775	100
Cook's Distance	.000	.064	.012	.014	100
Centered Leverage Value	.002	.114	.040	.028	100

a. Dependent Variable: Condition


Control
Football Fatigue
IsoK Hip
IsoK Ankle
PreTest
UA
DA
SA
NDA
Condition

Descriptive Statistics

	Condition	Mean	Std. Deviation	N
DA	PreTest	3.3450	.02724	20
	IsoKAnkle	3.3450	.02585	20
	IsoKHip	3.3505	.01849	20
	FootballFatigue	3.3460	.01984	20
	Control	3.3510	.01861	20
	Total	3.3475	.02204	100
NDA	PreTest	3.3455	.01538	20
	IsoKAnkle	3.3450	.01762	20
	IsoKHip	3.3420	.02215	20
	FootballFatigue	3.3495	.02645	20
	Control	3.3450	.02965	20
	Total	3.3454	.02254	100
UA	PreTest	3.3505	.01849	20
	IsoKAnkle	3.3420	.02042	20
	IsoKHip	3.3515	.02323	20
	FootballFatigue	3.3400	.01777	20
	Control	3.3470	.02716	20
	Total	3.3462	.02173	100
SA	PreTest	3.3440	.02563	20
	IsoKAnkle	3.3510	.01889	20
	IsoKHip	3.3455	.01638	20
	FootballFatigue	3.3535	.01424	20
	Control	3.3430	.02105	20
	Total	3.3474	.01968	100

Box's Test of Equality of Covariance Matrices[a]

Box's M	58.031
F	1.317
df1	40
df2	19908.088
Sig.	.087

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.

a. Design: Intercept + Condition

Levene's Test of Equality of Error Variances[a]

	F	df1	df2	Sig.
DA	1.538	4	95	.197
NDA	2.215	4	95	.073
UA	.841	4	95	.503
SA	1.382	4	95	.246

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.

a. Design: Intercept + Condition

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Intercept	Pillai's Trace	1.000	2343066.878[b]	4.000	92.000	.000	1.000
	Wilks' Lambda	.000	2343066.878[b]	4.000	92.000	.000	1.000
	Hotelling's Trace	101872.473	2343066.878[b]	4.000	92.000	.000	1.000
	Roy's Largest Root	101872.473	2343066.878[b]	4.000	92.000	.000	1.000
Condition	Pillai's Trace	.111	.678	16.000	380.000	.816	.028
	Wilks' Lambda	.891	.678	16.000	281.702	.815	.028
	Hotelling's Trace	.120	.679	16.000	362.000	.815	.029
	Roy's Largest Root	.098	2.336[c]	4.000	95.000	.061	.090

a. Design: Intercept + Condition

b. Exact statistic

c. The statistic is an upper bound on F that yields a lower bound on the significance level.

Tests of Between-Subjects Effects

Source	Dependent Variable	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Corrected Model	DA	.001[a]	4	.000	.361	.836	.015
	NDA	.001[b]	4	.000	.274	.894	.011
	UA	.002[c]	4	.001	1.098	.362	.044
	SA	.002[d]	4	.000	1.098	.362	.044
Intercept	DA	1120.576	1	1120.576	2248013.607	.000	1.000
	NDA	1119.170	1	1119.170	2138828.425	.000	1.000
	UA	1119.705	1	1119.705	2380219.673	.000	1.000
	SA	1120.509	1	1120.509	2906042.157	.000	1.000
Condition	DA	.001	4	.000	.361	.836	.015
	NDA	.001	4	.000	.274	.894	.011
	UA	.002	4	.001	1.098	.362	.044
	SA	.002	4	.000	1.098	.362	.044
Error	DA	.047	95	.000			
	NDA	.050	95	.001			
	UA	.045	95	.000			
	SA	.037	95	.000			
Total	DA	1120.624	100				
	NDA	1119.220	100				
	UA	1119.752	100				
	SA	1120.547	100				
Corrected Total	DA	.048	99				
	NDA	.050	99				
	UA	.047	99				
	SA	.038	99				

a. R Squared = .015 (Adjusted R Squared = -.026)

b. R Squared = .011 (Adjusted R Squared = -.030)

c. R Squared = .044 (Adjusted R Squared = .004)

d. R Squared = .044 (Adjusted R Squared = .004)