The Effects of Foot Structure and Athletic Taping on Lower Limb Biomechanics

By

Joanna Denyer

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

October 2012

Acknowledgements

There are a number of people to whom I would like to express my appreciation. First and foremost, my sincere thanks to Mitch, for ‘planting the seed’ in the final year of my undergraduate degree and encouraging me to continue my research. Until your suggestion, it had never occurred to me to stay on - so thank you for seeing my potential at the beginning, and for guiding me through. Many thanks also to Naomi Hewitt for your advice with methodology and your uncanny attention to detail.

To Pete and Ash, for your on-going support through the highs and lows….especially Pete for the endless phone calls, emails and advice…I’m so glad we started this together, and I look forward to graduating with you both!

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

And finally, to Tom, Mum, Dad, Jack, Luce and Julia; not a chance any of you will read this, but thank you all so much for the support, whether it’s been financial, emotional, motivational, inspirational, accommodational, transportational, procrastinational…… the list is endless and I will always be grateful.

Abstract

Context: Despite an association between foot structure and the incidence of lower limb injury in sport, few studies have measured the effects of neutral, pronated and supinated foot structures during dynamic activity. Furthermore, despite its widespread use as an injury prevention method, the effects of athletic taping on individuals with pronated and supinated foot structures are unclear.

Objectives: To explore whether individuals with pronated and supinated foot structures have poorer lower limb neuromuscular control as measured by postural stability and muscle reaction time in comparison to those with neutral feet. Additionally, the effects of athletic taping on individuals with neutral, pronated and supinated foot structures on aspects of lower limb neuromuscular control are also examined.

Subjects: All subjects used in this thesis were aged from 18 – 30 years and took part in at least two hours of exercise each week. Subjects were categorised in to groups according to navicular drop height measures; neutral 5 – 9 mm; pronated ≥ 10 mm; supinated ≤ 4 mm.

Methods: Neuromuscular control was analysed in subjects with neutral, pronated and supinated feet using dynamic postural stability and muscular reaction time measures. These measures were then repeated with four athletic taping conditions (arch tape, ankle tape, proprioceptive tape and no-tape) both before and after a period of exercise.

Results: Individuals with pronated and supinated foot structures were shown to have reduced postural stability in comparison to those with neutral foot structures during some dynamic tasks. Pronated and supinated foot structures also resulted in slower muscle reaction times in comparison to those with neutral feet during a tilt platform perturbation. No differences were identified between dominant and non-dominant limbs on subjects with neutral, pronated or supinated foot structures; however the high incidence of foot structure asymmetry did appear to result in differences between contralateral limbs in both postural stability and reaction time parameters. Arch and ankle taping resulted in increased neuromuscular control after application, yet these effects diminished after a period of exercise.

Conclusions: The results of this thesis provide evidence to suggest that foot structure does affect lower limb neuromuscular control as measured by postural stability and muscle reaction time. In addition athletic taping has been shown to affect neuromuscular control on subjects with neutral, pronated and supinated foot structures both before and after exercise. These findings may have wide implications in sport where individuals with pronated and supinated feet may be more susceptible to injury in comparison to those with neutral feet.

Contents

Accompanying Material

CD attached to inside back cover containing statistical outputs for all studies.

List of Figures

List of Tables

Abbreviations

AMTI	Advanced Mechanical Technology Inc.
BMI	Body Mass Index
COP	Centre of Pressure
DPSI	Dynamic Postural Stability Index
EMG	Electromyography
GRF	Ground Reaction Force
ICC	Intraclass Correlation Coefficient
MVIC	Maximal Voluntary Isometric Contraction
RMS	Root Mean Squared
SD	Standard Deviation
SEM	Standard Error of Measure
SENIAM	Surface EMG for a Non-Invasive Assessment of Muscles

Chapter One

Introduction

Seventy five percent of injuries during sport occur in the lower extremities (Chan, Yuan, Chien & Tsang, 1993); over 50% of which are associated with the closed kinetic chain whereby the foot is fixed on the ground when the injury occurs (Giza, Fuller, Junge & Dvorak, 2003). As the foot is the only base of support during single limb stance, it is reasonable that even slight alterations in the structure of the foot may affect lower limb biomechanics during sport (Cote, Brunet, Gansneder & Shultz, 2005). Alterations to the medial longitudinal arch of the foot are frequently associated with lower limb injuries (Kaufman, Brodine, Shaffer, Johnson & Cullison, 1999; Mei-Dan et al., 2005; Williams, McClay & Hamill, 2001) the most prevalent of which are lateral ankle sprains, with both low arches (Mei-Dan et al., 2005) and high arches (Williams et al., 2001) shown to be a risk factor. The prevention of ankle sprains is a serious challenge both to sports clinicians and to the wider society; as many as 302,000 patients attend emergency departments with ankle sprains each year the UK (Bridgman et al., 2003). Ankle sprains are reported to be by far the most common soft tissue injury in sport, accounting for 40% of all sporting injuries (Maffulli, Longo, Petrillo & Denaro, 2012). There is a clear demand for further research in this area both in terms of treatment, and in providing suggestions to reduce the incidence of ankle sprains.

Foot structure is related to foot function; a high arch is typically rigid and a characteristic of excessive supination, whereas a low arch is usually hypermobile and is related to excessive pronation (Franco, 1987). Despite an association between foot structure and lower limb injury, few studies have measured the effects of foot structure during methods applicable to sport. Previous studies have associated excessively pronated and supinated foot structures with deficits in some aspects of lower limb biomechanics, specifically neuromuscular control (Cote et al., 2005; Tsai, Yu, Mercer & Gross, 2006). Neuromuscular control is defined by Riemann and Lephart (2002) as “the unconscious activation of dynamic restraints occurring in preparation for, and in response to, joint motion and loading for the purpose of maintaining and restoring functional joint stability” (p. 73). Components of neuromuscular control include postural stability and muscular reaction time (Richie, 2001). Both Cote et al. (2005) and Tsai et al. (2006) found evidence to suggest that static postural stability is affected by foot structure; however these studies cannot be applied to the dynamic nature of sport where the majority of acute lower limb injuries occur indicating a major gap in current literature. As yet, no previous study has assessed the effects of pronated and supinated foot structures on muscular reaction time indicating a further need for research in this area. If differences are identified between neutral, pronated and supinated foot structures, interventions such as the use of athletic taping may then be required in order to prevent injury during sport.

In terms of injury prevention, athletic taping is widely used to help support structures of the foot and ankle during sport. In addition to the widely observed restrictive effects of athletic taping, it has been reported that taping may also result in neuromuscular effects including changes in muscle activity and changes in postural stability (Leanderson, Ekstam & Salomonsen, 1996; Shima, Maeda & Hiroashi, 2005). Some research has been carried out on muscle activity and postural stability with ankle taping but arch taping is less researched. Furthermore, a limitation of previous studies measuring the effects of athletic taping on lower limb biomechanics is that measurements are typically taken immediately after the application of tape. Whilst this provides an insight to the effects of newly applied tape, it has been shown that through exercise, taping loosens therefore reducing the mechanical restriction of the tape (Leanderson et al., 1996; Martin & Harter, 1993).This indicates that the results of newly applied tape cannot be generalised to sport. No studies have yet been found to have considered the effects of foot structure on athletic taping either before or after a period of exercise indicating a further gap in the literature.

1.1 Aims

The original aims of this thesis were as follows:

- To examine the effects of foot structure on lower limb biomechanics as measured by aspects of neuromuscular control, specifically postural stability and muscle reaction time.

- To examine the effects of foot structure and athletic taping on lower limb biomechanics as measured by aspects of neuromuscular control, specifically postural stability and muscle reaction time.

Through the natural progression and development of this thesis, a further area of interest was researched involving the effects of limb dominance and asymmetric foot structures on neuromuscular control. Whilst this was not originally intended to be included, it was observed during Chapters Three and Four that 31% of subjects had asymmetrical feet. Unexpectedly affecting such a large proportion of subjects' further exploration was required. The following aim was subsequently incorporated into this thesis in Chapter Five:

- To examine the effects of asymmetrical foot structure on lower limb biomechanics as measured by aspects of neuromuscular control.

1.2 Objectives

- Carry out a review of literature on current methods of identifying foot structure and measuring lower limb neuromuscular control.

- Measure neuromuscular control using tests of postural stability and muscle reaction time on individuals with pronated and supinated foot structures in comparison to those with neutral feet.

- Determine the effects of limb dominance and asymmetrical foot structures on measures of neuromuscular control.

- Carry out a review of literature on lower limb athletic taping techniques.

- Compare the effects of lower limb athletic taping techniques on individuals with neutral, pronated and supinated foot structures on aspects of lower limb neuromuscular control both before and after exercise.

1.3 Hypotheses

Below are general hypotheses referring to the thesis as a whole; specific hypotheses are identified within each study.

H_1 – Individuals with pronated and supinated foot structures will have reduced neuromuscular control on measures of dynamic postural stability (poorer) and muscular reaction time (slower) in comparison to those with neutral feet.

H_2 – Athletic taping will affect lower limb neuromuscular control on measures of dynamic postural stability and muscular reaction time in comparison to a no-tape control.

1.4 Contributions to Literature

A study in this thesis titled 'The Effects of Foot Structure on Muscular Reaction Time' has already been accepted for publication in the Journal of Athletic Training and is awaiting press (Study Two, Section 4.7). A number of other papers will also be submitted for publication in the near future.

Chapter Two

Review of Literature

2.1 Introduction

This review of literature examines the anatomy of the medial longitudinal arch and explores how variation from a ‘neutral’ foot structure may be associated with injury. In addition, it analyses different ways to quantify the arch, and investigates existing methods of measuring lower limb neuromuscular control. A number of search engines were used to acquire the relevant literature in this review including Scopus, Web of Knowledge and Google Scholar. Typical terms used in the search included *foot structure*, *navicular drop height*, *neuromuscular control* and *postural stability*.

2.2 Anatomy of the Medial Longitudinal Arch

In order to fully understand the function of the medial longitudinal arch, it is essential to have a thorough understanding of the anatomy of the area. As shown in Figure 2.1 the medial longitudinal arch is formed by the calcaneus, the talus, the navicular, the three cuneiforms, and the first, second and third metatarsals (Martini, 2006).

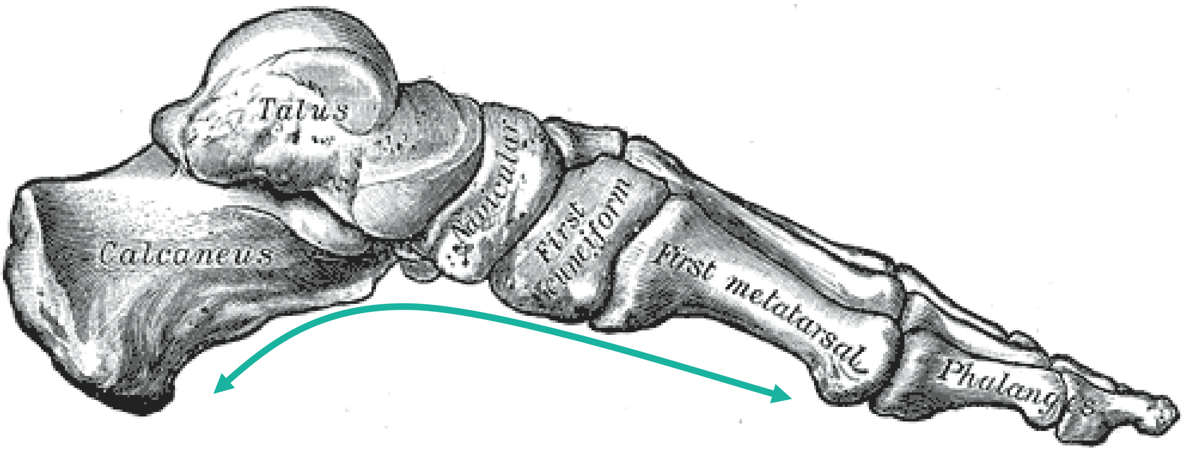


Figure 2.1 Medial view of the foot showing the medial longitudinal arch (Gray, 1918)

The arch is passively maintained by ligaments and tendons, in particular the plantar calcaneonavicular ligament (also known as the spring ligament) and the tibialis posterior tendon (Kaye & Jahss, 1991) (Figure 2.2). The calcaneonavicular ligament is of particular importance as when stretched it causes the calcaneus and navicular to separate and so the talus (the highest point of the arch), is lowered between the two bones (Sinnatamby, 1999) therefore reducing the overall height of the arch. Kaufman et al. (1999) noted that plantar fascia also plays a significant role in maintaining arch structure by accounting for approximately 25% of arch stiffness.

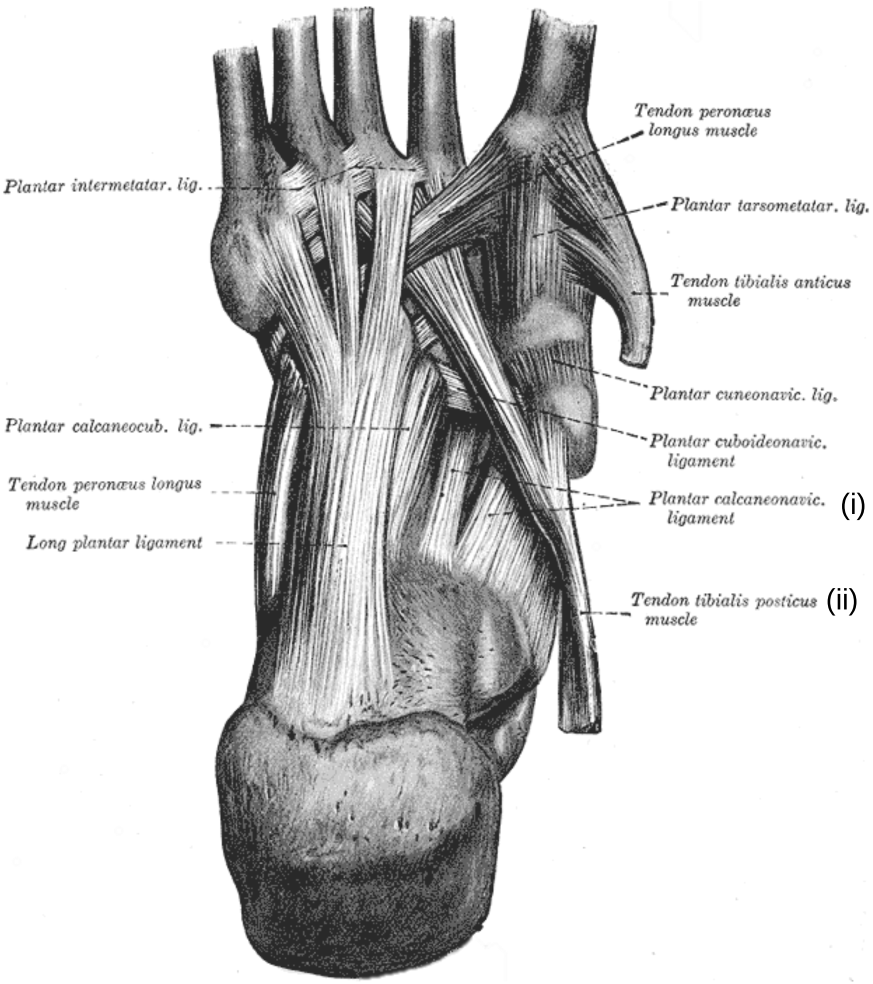


Figure 2.2 Plantar surface of the foot indicating the location of (i) the plantar calcaneonavicular ligament and (ii) the tibialis posterior tendon (Gray, 1918).

2.2.1 Functional Anatomy of the Medial Longitudinal Arch

The medial plantar surface of the foot is elevated to ensure that the muscles, nerves and blood vessels that supply the inferior surface are not in contact with the ground; the arch also acts as a shock absorber and allows weight transfer during gait (Martini, 2006). The arch of the foot must therefore maintain a balance between being rigid and inflexible for efficient weight transfer, yet flexible and adaptable to cope with varying surfaces (Franco, 1987).This function is best described through the gait cycle, where both pronation and supination of the subtalar joint occur; this is highlighted in Figure 2.3.

Starting at heel strike (i), the foot is in supination, combining subtalar inversion, plantar flexion, and forefoot adduction with the tibia in external rotation; at this point, the subtalar joint is in a locked and stable position (Nishikawa, Kurosaka, Yoshiva, Lundin & Grabiner, 2002). Prior to the mid-stance phase (ii), the lateral surface of the foot contacts the ground first, this initiates pronation requiring the subtalar joint to unlock, therefore revealing a less stable loose-packed position (Langdon, Brukner & Baker, 1991). This enables shock absorption, and adaptation to the walking surface (Bolgla & Malone, 2004). Supination reoccurs during mid-stance to toe off (iii), tightening the plantar fascia which provides structure and rigidity. This is known as the close-packed position which is required for push off and propulsion in to the swing phase of gait (Bolgla & Malone, 2004; Langdon et al., 1991; Nishikawa et al., 2002).

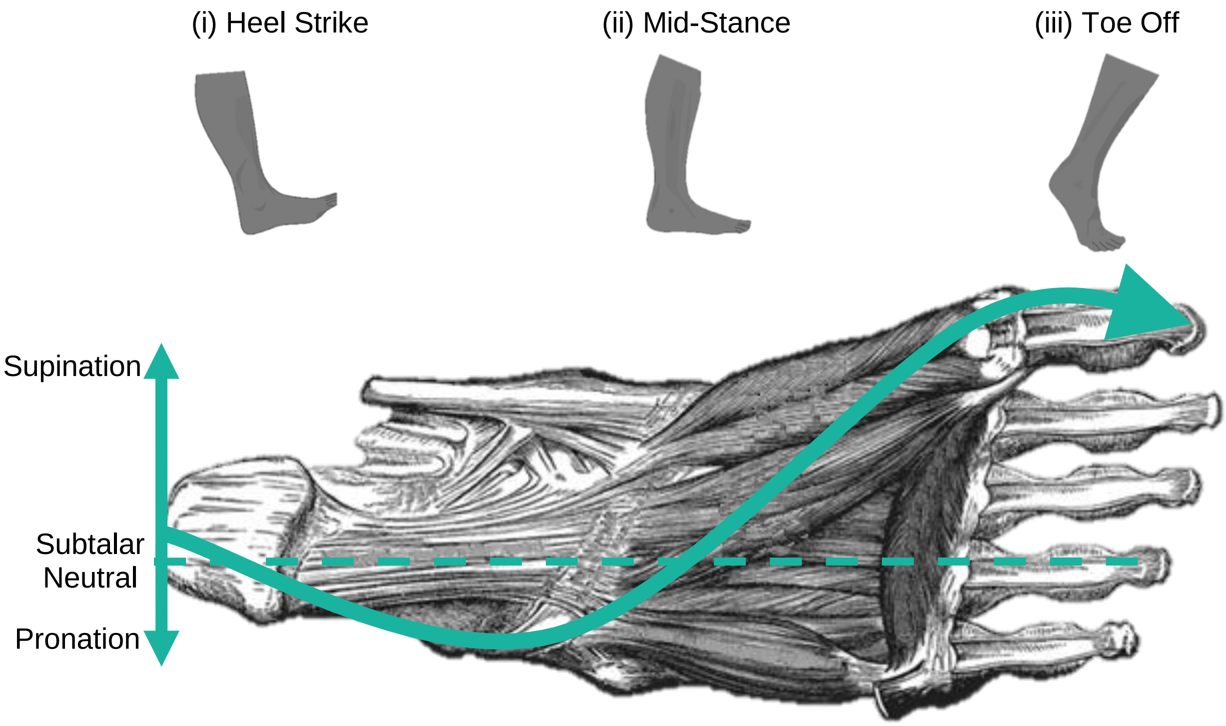


Figure 2.3 Plantar surface of the foot highlighting the transfer between supination and pronation during the stance phase of gait. Adapted from Gray (1918).

The plantar fascia helps to maintain the arch throughout the gait cycle via the windlass mechanism whereby the plantar fascia shortens due to flexion of the great toe. This pulls the calcaneus closer to the metatarsal heads and therefore raises the arch (Bolgla & Malone, 2004). In doing so, the plantar fascia contributes to the appropriate amount and timing of pronation and supination during the gait cycle (Bolgla & Malone, 2004).

2.2.2 Variances in the Anatomy of the Medial Longitudinal Arch

The structure of the arch has been shown to relate to the function of the foot (Kaufman et al., 1999; Razeghi & Batt, 2002). Franco (1987) found that deviations in the normal structure of the medial longitudinal arch produce unbalanced and unstable conditions of the foot such as pes cavus or pes planus. Pes planus is identified when the calcaneus is everted and the medial longitudinal arch is low or absent (Ledoux & Hillstrom, 2002). It is normally due to biological variability, however a low arch foot has traditionally been seen as undesirable both functionally and cosmetically (Staheli, Chew & Corbett, 1987). Several studies have associated the presence of pes planus with prolonged pronation or hyper-pronation (Bennell, Matheson, Meeuwisse & Brukner, 1999; Del Rossi, Fiolkowski, Horodyski, Bishop & Trimble, 2004; Franco, 1987). The main cause of excessive pronation is joint hypermobility and tightness of the posterior lower leg muscle groups (Nicolopoulos, Scott & Giannoudis, 2000). Some increased pronation of the foot is often functional, but excessive pronation may lead to injury (Hintermann & Nigg, 1998) (see Section 2.3). Pes planus often results in elongation of the plantar fascia, making the propulsion phase of gait less effective (Bolgla & Malone, 2004).

Pes cavus is identified when the calcaneus is inverted and the arch is high (Ledoux & Hillstrom, 2002); it is associated with excessive supination and is also seen as an undesirable foot trait relating to inflexible rigid arches which may not adapt to the underlying surface (Franco, 1987). Franco (1987) stated that this often results in increased stress on the surrounding structures in order to maintain postural stability; furthermore pes cavus results in uneven weight distribution which leads to increased pressure on the lateral side of the foot. Bolgla and Malone (2004) highlighted that people with pes cavus feet usually have decreased flexibility of the gastrocnemius, soleus, and Achilles tendon.

2.2.3 Incidence of Pronated and Supinated Foot Structures

Franco (1987) reported arch abnormalities such as pes planus and pes cavus to be among the most common lower extremity disorders seen by physical therapists working in sports medicine. Approximate foot structure distribution was not found for a healthy sporting population, however a study using Indian adults found that of the 1846 participants screened, 87% had neutral arches, 3% had low arches and 10% had high arches (Sachithanandam & Joseph, 1995). In a similar study of 2047 Western diabetic patients, Ledoux et al. (2003) found that 57% of patients had neutral feet, 19% had low arches and 24% had high arches. Korpelainen, Orava, Karpakka, Siira and Hulkko (2001) observed that the incidence of pronated and supinated foot structures may be largely underestimated due to inaccuracies in assessment of foot structures (see Section 2.5).

Race and ethnicity are thought to be associated with differences in foot structure, although few studies were found to have observed this within a sporting population. Using footprint ratio (also known as arch index; a measure of the ratio of total midfoot area to toeless foot area), Igbigbi and Msamati (2002) showed that the incidence of pes planus was significantly higher in Malawians than Caucasian American and European teenagers. In addition, a study of 784 American adults over 65 years old found that 19% had pes planus feet; most commonly among African Americans, followed by Non-Hispanic Caucasians and Puerto Ricans (Dunn et al., 2004). Only 5% of adults in this study had pes cavus feet, which was not found to differ between race. A recent study by Gurney, Kuch, Rosenbaum and Kersting (2012) found that Maori adults had significantly higher arch index values than New Zealand Caucasians suggesting that healthy Maori adults are more likely to have pronated feet in comparison to healthy New Zealand Caucasian adults. Interestingly this difference was not identified in a sporting population when comparing elite rugby league Maori and New Zealand Caucasians (Gurney, Kersting & Rosenbaum, 2009).

Gender is also speculated to have an effect on foot structure, however current research is inconclusive. In the United States, Zifchock, Davis, Hillstrom and Song (2006) found no differences between the arch height index of 145 adults (68 men and 77 women). Similarly, Nguyen and Shultz (2009) found no differences between 218 adults (102 men and 116 women) through measures of navicular drop height (a measure of excessive subtalar joint pronation; see Section 2.5.2). In contrast, Dunn et al. (2004) found that the prevalence of pes cavus in an elderly American population (≥ 65 years) was more common in women than in men. In Malawi, Igbigbi and Msamati (2002) showed that males had a higher incidence of pes planus than females.

Body mass index (BMI) is also thought to have an effect on foot structure. Again, few studies have measured this among healthy populations however Hills, Hennig, McDonald and Bar-Or (2001) found that obese men and women (BMI ≥ 30 kg/m^2) had significantly higher plantar pressures under the medial longitudinal arch suggesting excessive pronation, in comparison to non-obese groups during both walking and standing. In contrast, using navicular drop height, Nielsen, Rathleff, Simonsen and Langberg (2009) found no evidence to suggest that BMI affected arch height, however a major flaw in this claim is that the study excluded all over weight participants (BMI ≥ 30.5 kg/m^2). Nielsen et al. (2009) also compared foot length and navicular drop height and found that foot length had a significant effect; for every 10 mm increase in foot length, the navicular drop increased by .4 mm for males and .3 mm for females. This indicates that people with larger feet are more likely to have pronated feet however; this was the only study found to have measured the effects of foot size on arch structure and so requires further verification.

2.3 Epidemiology of Injuries Associated With Foot Structure

Past literature is contrasting as to whether pronated or supinated foot structures are associated with injury; Michelson, Durant and McFarland (2002) and Hargrave, Carcia, Gansneder and Shultz (2003) found no evidence to show that differences in foot structure predispose lower limb injury, however numerous studies have found otherwise.

2.3.1 Foot and Ankle Injuries Associated with Foot Structure

Mei-Dan et al. (2005) observed the incidence of lateral ankle sprains among Israeli female military recruits and found that those with lower medial longitudinal arches had significantly more acute and recurrent lateral ankle sprains than those with neutral arches, therefore indicating that a low medial longitudinal arch is a possible risk factor for lateral ankle sprains. In addition, in a study of 40 runners, Williams et al. (2001) found that lateral ankle sprains were the most common injury in the high arched group. The findings of both Mei-Dan et al. (2005) and Williams et al. (2001) are important as several studies have shown that lateral ankle sprains are the most common of all injuries in a variety of sports (Hootman, Dick & Agel, 2007; Kumai, Takakura, Rufai, Milz & Benjamin, 2002), with around 5000 ankle sprains being reported each day in the UK (Brooks, Potter & Rainey, 1981). A study by Dahle, Mueller, Delitto and Diamond (1991) found no relationship between foot structure and the incidence of ankle sprains, however this study has a major flaw in that foot structure was categorised according to visual inspection rather than an objective measure such as navicular drop height (this is further discussed in Section 2.5).

Manoli and Graham (2005) stated that cavus foot structures commonly result in recurrent inversion sprains. Recurrent ankle sprains can lead to neuromuscular deficits which in turn can result in functional ankle instability if left unaddressed (Freeman, Dean & Hanham, 1965). Functional ankle instability is defined as “the subjective feeling of ankle instability” or, “recurrent symptomatic ankle sprains due to proprioceptive and neuromuscular deficits” (Tropp, 2002, p. 512). It is a serious medical concern as it has been reported to affect around 40% of patients with ankle sprains (Safran, Benedetti, Bartolozzi & Mandelbaum, 1999). It has been estimated that as many as 302,000 patients attend accident and emergency departments with ankle sprains each year in the UK alone, with approximately 42,000 being classed as severe (defined as patients who could not weight-bear, had lateral ankle tenderness, and did not have a fracture) (Bridgman et al., 2003). Additionally, in the United States ankle sprains have an estimated incidence rate of 2.15 per 1000 person-years (Waterman, Owens, Davey, Zacchilli & Belmont, 2010), or more than 23,000 per day

(Kannus & Renström 1991) with estimates of an annual healthcare cost of $2 billion (Soboroff, Pappius & Komaroff, 1984). There is a clear demand for further research in this area both in terms of treatment, and in providing suggestions to reduce the rate of ankle sprains.

Hunt, Sneed, Hamann and Chisam (2004) indicated that those with excessive pronation during gait are more prone to plantar fasciitis, which is common in both athletic and sedentary populations (Radford, Landorf, Buchbinder & Cook, 2006). McKenzie, Clement and Taunton (1985) stated that plantar fasciitis is also common among runners with excessive supination. In Williams et al. (2001) study of 40 runners, it was found that those with high arches appear to be more prone to bony injuries, injuries on the lateral side of the lower extremities, and injuries at the foot. Common injuries of the high arched group included tibial stress fractures and lateral ankle sprains, whereas those with low arches appear to be more at risk of soft tissue injuries on the medial side of the lower extremity and at the knee.

2.3.2 Lower Leg and Knee Injuries Associated with Foot Structure

A study by Kaufman et al. (1999) found that U.S Navy recruits with either pes planus or pes cavus feet had nearly twice the incidence of stress fractures during basic training compared with subjects with neutral arch height. Kaufman et al. (1999) added that the most common site for stress fractures were in the lower leg (49%), the foot (39%) and the femur (12%). Bennell et al. (1999) suggested that both pes planus and pes cavus may cause shin pain encompassing bone, periosteum and muscle compartment pain. Bennell et al. (1999) advised that pes planus may cause shin pain due to irregular contraction of lower limb muscles caused by excessive pronation during gait, whereas pes cavus has limited shock absorption capacities therefore increasing impact pressure on bone. In children and adolescents, pes planus has been linked to several conditions of the lower extremities including peroneal tendinitis, tibialis posterior tendinitis, Achilles tendinitis, and calcaneal apophysitis (Micheli & Fehlandt, 1992).

Lutter (1980) found that 77% of knee injuries were related to biomechanical dysfunction of the foot in a study of 264 runners. Several studies have noted a strong link between excessive pronation and anterior cruciate ligament (ACL) tears (Allen & Glasoe, 2000; Carcia, Drouin & Houglum, 2006). Beckett, Massie, Bowers and Stoll (1992), and Stergiou and Bates (1997) suggest this may be caused by excessive internal rotation of the tibia during prolonged pronation which produces a preloading effect on the ACL. Additionally, Williams et al. (2001) found that runners with low arches were prone to patella tendonitis.

McKenzie et al. (1985) stated that excessively pronated feet predispose an athlete to injuries on the medial side including tibial stress syndrome, patellofemoral pain syndrome, and posterior tibialis tendinitis. In contrast, those with excessively supinated feet are more prone to injuries on the lateral side including iliotibial band friction syndrome, peroneus tendinitis, stress fractures, and trochanteric bursitis (McKenzie et al., 1985).

2.4 Validity and Reliability

Before reviewing literature on current methods used to classify foot structures, validity and reliability must first be understood in order to make comparisons between different techniques.

2.4.1 Defining Validity

Validity refers to the quality of a research study (Vincent & Weir, 2012); high quality research must demonstrate both external and internal validity. According to the Oxford Dictionary of Sport Science and Medicine (Kent, 2006), external validity is defined as “the extent to which the results of an investigation may be generalised to the population as a whole” (p. 200); in this case, the sporting population. This is related to ecological validity, which indicates how closely the conditions under which the measurements are taken reflect the actual conditions of real sporting environment (Williams & Wragg, 2003). Internal validity on the other hand refers to “the extent to which the outcome of a specific research study is a function of the variables that are measured, controlled or manipulated” (Kent, 2006, p. 289); or in simple terms, the control of anything which could potentially influence the results. Sport science research must therefore work to achieve a balance between internal and external validity where research is conducted in such a way that it is directly applicable to sport and sports injury whilst at the same time maintaining strong experimental control in order to avoid sources of error. Sources of error include intervening variables (factors which are not controlled), instrument error (data collected using faulty equipment) and investigator error (bias in recording data) (Vincent & Weir, 2012).

2.4.2 Defining Reliability

Reliability refers to the consistency of a test or measurement (Weir, 2005), and is an integral part of most quantitative research. Reliability testing attempts to identify the amount of error present within a measurement, which can be defined as “all sources of variability not explained by the independent variable”, this can be calculated as the difference between the true value and the observed value of a measurement (Bruton, Conway & Holgate, 2000, p. 95). Potential errors may be random or systematic; random errors may occur due to inattention, tiredness or mechanical inaccuracy, which may cause either underestimation or overestimation of the true quantity (Bruton et al., 2000; McDowell, 2006). Systematic errors are on the other hand, predictable errors, for example a variation of the true quantity caused by the learning effect. Such errors would therefore affect validity rather than reliability, as the test values would not represent the true quantity being measured (Bruton et al., 2000; McDowell, 2006).

There are two types of reliability; relative, and absolute. Relative reliability is defined as the degree to which individuals maintain their position in a sample over repeated measurements which is usually measured using a correlation coefficient test, giving details about the association between two variables, not necessarily their proximity (Bruton et al., 2000; McDowell, 2006; Weir, 2005). Absolute reliability on the other hand can be defined as the degree to which repeated measurements vary for individuals, focusing on the precision of the measurement, which is often quantified using the standard error of measurement (SEM) (Bruton et al., 2000; McDowell, 2006; Weir, 2005).

2.4.3 Calculating Reliability

Until recently, the Pearson's correlation coefficient was the preferred test to establish relative reliability, however it could be criticised as it did not take in to account systematic differences within data, therefore only indicated whether or not sets of data were proportional to each other, rather than highlighting the extent of agreement between them (Bruton et al., 2000). Intraclass correlation coefficient (ICC) is now the preferred measure of reliability (McDowell, 2006), as it overcomes some of the problems associated with Pearson's correlation coefficient and shows whether the actual values among sets of data are the same, rather than merely identifying the similarity of relative standings between the data (Bruton et al., 2000; McDowell, 2006). As with the Pearson's correlation coefficient, the ICC ranges from 0 to 1, with 0 indicating no reliability, and 1 indicating strong reliability (Bruton et al., 2000). Interpretation of the ICC is straightforward in that a score of .95 indicates that 95% of the observed variance is due to true score variance, where as 5% is attributable to error (Weir, 2005). Munro (2000) gave advisory guidelines to determine the strength of ICC scores; .00 - .25 indicates little, if any correlation; .26 - .49 indicates low correlation; .50 - .69 indicates moderate correlation; .70 - .89 indicates high correlation, and .90 - 1.00 indicates a very high correlation.

The standard error of measurement (SEM) is often used alongside ICC tests giving a measure of absolute reliability, which has the same units as the measurement of interest, and refers to the precision of individual scores on a test (Weir, 2005). The SEM can be calculated as follows: SEM $= SD\sqrt{1 - ICC}$, where SD is the standard deviation of the scores from all subjects, and ICC is the reliability coefficient (Weir, 2005). The SEM can be described as a range of estimated measurement error around the observed score (Eechaute, Vaes, Duquet & van Gheluwe, 2007), the smaller the SEM value, the greater the absolute reliability (Bruton et al., 2000). Denegar and Ball (1993) and Weir (2005) recommend that ICC and SEM scores should be reported alongside each other following a repeated measures ANOVA, to provide representative analysis regarding the reliability and precision of a particular measurement.

2.5 Classification of Foot Structure

There are a number of ways to classify foot structure with reference to the medial longitudinal arch. In most routine clinical assessments of the foot, this is first done subjectively by visual non-quantitative inspection where the therapist examines the foot in a weight bearing position to observe absence or presence of foot arches (Razeghi & Batt, 2002). Although this method may be quick, simple and therefore more clinically applicable than some of the more scientific techniques such as the use of radiography, it allows for high inter and intra tester variability (Cowan, Robinson, Jones, Polly & Hudson Berrey, 1994). Even with intensive training and operational definitions associated with foot structure, Dahle et al. (1991) only reached 73.3% agreement when identifying foot structure by visual inspection of 77 feet. A study by Dunn et al. (2004) required a fast and simple way to quantify foot structure in a sample of 784 patients and identified a flat foot as one where the examiner was unable to insert his/her fingers under the arch of the foot with the respondent in a standing position. A high arch was identified if the examiner could insert his/her fingers all the way underneath the arch to the lateral edge of the foot. This study reported no evidence of reliability testing, and did not identify whether one examiner, or a team of examiners carried out the testing and so can therefore be criticised for using a subjective and vague method for identifying foot structure.

2.5.1 Objective Measures of Classifying Foot Structure

There are several more objective ways of classifying foot structure; foot print indices involve taking measurements from an imprint of the foot either using a simple ink pad, or using a foot scanner (Razeghi & Batt, 2002). Such measurements produce foot prints similar to those shown in Figure 2.4, where foot structure can be subjectively examined. Similar measurements can also be obtained from X-rays and MRI scans.

(i) (ii) (iii)

Figure 2.4 Images to indicate foot prints of different foot structures; (i) low arch, (ii) neutral, (iii) high arch

Video analysis and plantar pressure measures can also be used as an indirect indicator of subtalar pronation and supination during gait (Ator, Gunn, McPoil & Knecht, 1991; Lange, Chipchase & Evans, 2004; Russo & Chipchase, 2001), however each of these measures are time consuming, and are dependent on expensive equipment, therefore not very practical in a clinical environment.

Anthropometric measures such as navicular drop height or arch index are also used to classify foot structure, and are more clinically applicable than some of the previously mentioned techniques as they are easier to perform and require less equipment (Razeghi & Batt, 2002). Most anthropometric measures are based on the assumption that the dynamic behaviour of the foot can be anticipated from static characteristics (Razeghi & Batt, 2002). The navicular drop height measurement is therefore favoured by many researchers in comparison to other anthropometric measures such as direct measure of arch height, rearfoot angle, arch index or valgus index as it takes into account the dynamic nature of the subtalar joint.

2.5.2 Navicular Drop Height

Navicular drop height was first identified by Brody (1982) as an indicator of excessive subtalar joint pronation, measuring the displacement of the navicular bone between the subtalar neutral position and a relaxed stance. Razeghi and Batt (2002) described navicular drop height as the simplest method of obtaining quantifiable information about foot structure.

As described by Brody (1982), the subtalar neutral position can be found by placing a thumb and index finger either side of the talus joint (as shown in Figure 2.5) then by inverting and everting the ankle, the therapist can detect when both depressions on the thumb and index finger feel equal. This point is known as the subtalar neutral position where the ankle is neither in pronation or supination (Root, 1973). The most medial point of the navicular tuberosity (also shown in Figure 2.5) is then marked, and the distance from the floor is measured using a height gauge. After finding the subtalar neutral position, the most medial point of the navicular tuberosity is again measured in a relaxed bilateral stance. Navicular drop height is calculated by subtracting the relaxed stance measure, from the neutral stance measure (Brody, 1982).

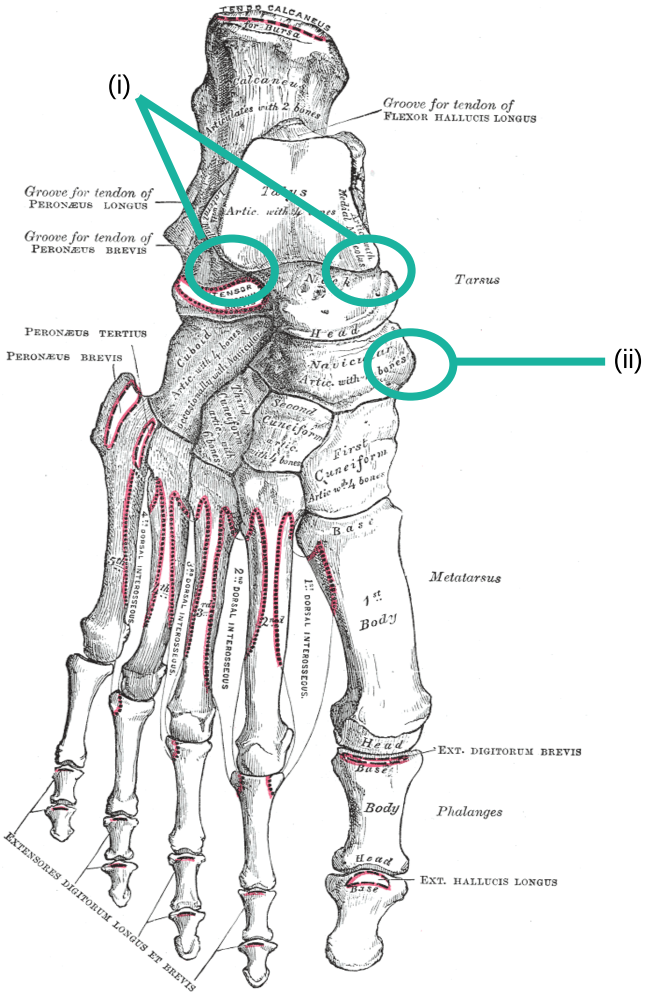


Figure 2.5 Dorsal view of right ankle joint. (i) Indicates positioning for palpation of subtalar neutral position, (ii) indicates the navicular tuberosity (Gray, 1918).

There is some discrepancy in the literature over whether the subtalar neutral measurement should be collected in a weight bearing, partial weight bearing, or non-weight bearing position. Brody (1982) first describes taking the subtalar neutral measurement in a bilateral stance, however since then other studies have measured participants in a single limb stance (McPoil & Cornwall, 1996; Vinicombe, Raspovic & Menz, 2001) and when seated (McPoil et al., 2008). Vinicombe et al. (2001) suggested that single limb stance is more representative of the midstance phase of gait, therefore a more valid measure, however, this could be criticised for low reproducibility. Most studies use a bilateral stance as it is easier to reproduce and has been shown to have higher inter-tester reliability than non-weight bearing open kinetic chain measures (Picciano, Rowland & Worrell, 1993).

2.5.3 Reliability of Navicular Drop Height Measures

In studies using participants with different foot structures, it is imperative that participants are assigned to the correct group as any discrepancies may confound results. Many studies have therefore questioned the reliability of the navicular drop height measurements with contradicting conclusions (Mueller, Host & Norton, 1993; Picciano et al., 1993; Sell, Verity, Worrell, Pease & Wigglesworth, 1994; Vinicombe et al., 2001). After measuring navicular drop height on 29 healthy subjects, Mueller et al. (1993) concluded that it is an acceptable method of measuring foot pronation with ICC values at ≥ .67. Whereas Vinicombe et al. (2001) found that with a standard error range from ± 1.5 mm to ± 3.5 mm it was only a moderately reliable measure. In a study of 15 healthy subjects, Picciano et al. (1993) found poor inter-tester reliability when comparing two inexperienced examiners, and moderate to poor intra-tester reliability in navicular drop height measurements, concluding that it is not a reliable measure when performed by inexperienced testers. Conversely, Sell et al. (1994) found acceptable levels of reliability when comparing two testers with no experience in navicular drop height measures with ICC results ranging from .73 - .96 and SEM ranging from 1.5 to 1.9 mm suggesting that navicular drop height is a reliable measure, and is not dependent on experience. Findings by Saltzman, Nawoczenski and Talbot (1995) were consistent with previous literature showing higher intra-tester values of reliability than inter-tester values for navicular measurements.

Differences in measurements may be explained by the slight variances in technique, and the ambiguity and subjectivity in finding subtalar neutral (Menz, 1998). Maslen and Ackland (1994) identified that the displacement of skin over the navicular tuberosity during the transition between weight bearing and non-weight bearing postures when taking measurements may affect results. Some authors have attributed inaccuracies of navicular measurements to anatomical variability in bony and soft tissue landmarks making palpation difficult (Saltzman et al., 1995; Vinicombe et al., 2001). As it is difficult to overcome inter-tester variability, most studies only allow one tester to carry out the measurements of navicular drop height (Ator et al., 1991; Vicenzino, Feilding, Howard, Moore & Smith, 1997). This makes it difficult to compare similar studies, but also has negative practical implications for example in multi-practitioner clinics, measurements taken by different health professionals may be incomparable (McPoil et al., 2008). Carcia et al. (2006) attempted to control reliability of testing by allowing testing to commence only when the tester achieved intra-tester reliability of .90 ± .5 mm. This could be tested by observing the day to day consistency of blind measurements over three attempts.

2.5.4 Identifying Pronated and Supinated Feet with Navicular Drop Height Measures

A further limitation to using navicular drop height as an indicator of foot structure is that there is no consensus in current literature as to how much movement at the navicular is perceived as 'normal'. It is well established that some movement of the subtalar joint is functional, particularly during gait (Nishikawa et al., 2002); however, there is no agreement in current literature on the amount of movement which is thought to be excessive. A navicular drop measurement of 10 mm or more is generally accepted as an indicator of excessive pronation (Lange et al., 2004; Vicenzino, McPoil, Russell & Peisker, 2006; Wall, Swankik & Swankik, 2005), however Brody (1982) suggested 10 mm is normal and 15 mm indicates excessive pronation. More recently, in a study by Cote et al. (2005), subjects were grouped according to foot structure where a navicular drop height of ≥ 10 mm indicated a pronated foot, 5 – 9 mm indicated those with neutral arches and ≤ 4 mm indicated a supinated foot. Hargrave et al. (2003) also used similar categories. There seems to be little evidence to support the varying classifications of foot structures, however it is advisable to use the same categories as used in other studies in order to make comparisons valid.

Although there are several criticisms to the navicular drop height measurement, it remains the preferred method for differentiating foot structure in current literature, perhaps because it is inexpensive, relatively easy to perform, and practical for a clinic setting (Razeghi & Batt, 2002).

2.6 Foot Structure and Lower Limb Neuromuscular Control

Previous studies have associated excessively pronated and supinated foot structures with deficits in some aspects of lower limb neuromuscular control in comparison to those with neutral feet (Cote et al., 2005; Tsai et al., 2006). Neuromuscular control is defined by Riemann and Lephart (2002) as "the unconscious activation of dynamic restraints occurring in preparation for, and in response to, joint motion and loading for the purpose of maintaining and restoring functional joint stability" (p. 73). Components of lower limb neuromuscular control include proprioception, muscle strength, muscle reaction time and postural stability (Richie, 2001).

2.6.1 Postural Stability as a Measure of Neuromuscular Control

Postural stability has been defined by Chen, Yeung, Wang, Chu and Yeh (1999) as the ability to keep the centre of mass within the base of support when maintaining static postures. Chen et al. (1999) further identified dynamic postural stability as the ability to move the centre of mass in relation to the base of support in a controlled manner when engaged in movement. Postural stability is maintained by balance, which can be defined as the process of maintaining the centre of gravity within the body's base of support (Guskiewicz & Perrin, 1996), this involves both central and peripheral components of the nervous system (Chen et al., 1999; Guskiewicz & Perrin, 1996). Peripheral components of the nervous system include the somatosensory, visual, and vestibular systems (Cote et al., 2005); these systems work continuously to influence the central nervous system which in turn causes appropriate muscular responses to maintain postural stability (Nashner, Block & Wall, 1982). The somatosensory system is of particular importance as it incorporates information from mechanoreceptors (mechanical stress), thermoreceptors (temperature) and nociceptors (pain) (Riemann and Lephart (2002). When standing, the mechanoreceptors in the feet will detect perturbations and provide afferent feedback to the central nervous system which will in turn respond by initiating actions to either widen the base of support, or shift centre of gravity (Kell & van Deursen, 2005). Stevens and Tomlinson (1971) stated that even in the most comfortable position, it is not possible for man to stand absolutely still; the apparently static posture is in fact composed of continuous small movements.

Balance is dependent on the closed kinetic chain, the foot which is the most distal part of the chain, is the only base of support in normal standing (Guskiewicz & Perrin, 1996). Balance is therefore easily influenced by irregularity or injury to any part of the closed kinetic chain which may inhibit appropriate feedback required to maintain postural stability (Guskiewicz & Perrin, 1996). Cote et al. (2005) suggested that with such a relatively small base of support, even minor biomechanical alterations of the foot may influence postural stability.

Four studies were found to have measured postural stability on healthy subjects with different foot structures during static single leg stance (Cobb, Tis, Johnson & Higbie, 2004; Cote et al., 2005; Hertel, Gay & Denegar, 2002; Tsai et al., 2006). Hertel et al. (2002) found that those with cavus feet had significantly larger centre of pressure excursion areas than those with neutral feet. No differences were found when comparing the neutral group to the planus group. Cobb et al. (2004) found that subjects with > 7° forefoot varus (associated with excessive pronation) had significantly impaired postural stability in the anterior-posterior plane. Tsai et al. (2006) found that both those with pronated and supinated feet had significantly poorer postural stability than those with neutral feet, whereas Cote et al. (2005) found differences between the pronated and supinated groups, but neither groups were significantly different to the neutral group. Cote et al. (2005) suggested that static tests may not be demanding enough on the postural control system to identify deficits caused by altered feedback or structural mal-alignments, however this is questionable as Hertel et al. (2002), Cobb et al. (2004), and Tsai et al. (2006) did identify differences during the same task. Static tests are further discussed in Section 2.7.2. A vast difference between the four studies was the technique for classifying foot structure which could account for the different findings; Hertel et al. (2002), and Cobb et al. (2004) used measures of forefoot alignment, Tsai et al. (2006) measured the angle of the medial longitudinal arch, and Cote et al. (2005) used navicular drop height. Cote et al. (2005) also measured dynamic postural stability on healthy subjects with different foot structures and found that both pronated and supinated groups showed differences during a star excursion balance test. The star excursion balance test integrates single leg stance on one leg, with maximum reach on the opposite leg (Cote et al., 2005). Dynamic tests of postural stability are discussed in Section 2.7.3.

There are two main theories to explain why different foot structures may affect postural stability; researchers are inconclusive as to whether differences may be due to mechanical reasons, or due to differences in peripheral input. Hertel et al. (2002) suggested that those with cavus foot structures have no anatomical block in pronation between the medial aspect of the foot and the ground when compared with neutral or planus feet; instead the movement is restricted by the physiological limits of the subtalar and midtarsal joints, therefore allowing more movement than other foot structures. Hertel et al. (2002) also suggested an alternative theory that differences in pes cavus subjects may be simply due to having reduced contact area between the plantar surface of the foot and the force plate therefore reducing cutaneous stimulation. Tsai et al. (2006) also theorised that those with supinated feet had reduced plantar contact leading to reduced sensory input. However reduced stability in the pronated group was thought to be due to the structure of the pronated foot, which is typically hypermobile therefore leading to increased instability (Tsai et al., 2006). This is consistent with Cobb et al. (2004) who suggested that those with hypermobile feet are reliant on soft tissue structures for stability. Cote et al. (2005) also suggested that increased foot mobility may explain differences in pronated subjects.

Despite the evidence to suggest that variances in foot structure may lead to differences in postural stability, neither Hargrave et al. (2003) or Nachbauer and Nigg (1992) found differences between subjects with different foot structures during dynamic tasks (landing from a 30 cm single leg drop, and running respectively). More research is required in this area to determine if foot structure has an effect on dynamic postural stability. As there are potential differences caused by foot structure, Hertel et al. (2002) recommends that when measuring postural stability, subjects should be matched for foot structure, as well as age, gender and activity level. This recommendation is supported in literature as it has been identified that postural stability becomes worse with age (Hageman, Leibowitz, & Blanke, 1995), and also is worse among those who are not involved with regular activity (Skelton, 2001). Wikstrom, Tillman, Kline and Borsa (2006) found that females may have different postural control strategies to males as females had higher dynamic postural stability scores in the vertical direction, as well as having higher overall scores (indicating reduced postural stability), therefore supporting the use of matched subjects in postural stability research.

2.6.2 Other Measures of Neuromuscular Control

As some studies have indicated that foot structure may affect postural stability, it is logical to review other aspects of lower limb neuromuscular control to see if any other differences have been identified in previous literature.

Measures of muscular reaction time are frequently used in literature to identify neuromuscular deficits among subjects with lower limb abnormalities such as hypermobile ankles (Beckman & Buchanan, 1995; Shima et al., 2005), or functional ankle instability (Fernandes, Allison & Hooper, 2000; Mitchell, Dyson, Hale & Abraham, 2008a; Vaes, Duquet & van Gheluwe, 2002). Either using a static (standing) platform (Karlsson & Andreasson, 1992) or more recently, a dynamic (walking) platform (Hopkins, Hunter & McLoda, 2006; Hopkins, Brown, Christensen & Palmieri-Smith, 2009), they are designed to simulate the mechanics of an inversion ankle sprain, therefore stressing the dynamic defence mechanism (Mitchell et al., 2008a). The dynamic defence mechanism coined by Konradsen, Voight and Hojsgaard (1997) consists of "both a peripheral reflex reaction of the peroneal muscles that can counteract the inverting moment, and also a centrally mediated strategy that results in a uniform reaction pattern of the muscles and joints of both the unilateral and contralateral limb" (p. 57). Muscular reaction time is measured using electromyography (EMG) and typically includes the peroneus longus and the tibialis anterior (Mitchell et al., 2008a) as these muscles have a direct role in the dynamic defence mechanism. The gluteus medius has also been included to identify the use of hip strategy during perturbation (Beckman & Buchanan, 1995). As yet, no studies have been found to use subjects with pronated or supinated foot structures. This is surprising considering the association between foot structure and lateral ankle sprains; it is clear that further research is required in this area.

Proprioception is another well documented aspect of neuromuscular control, it is a broad concept which can be defined as “a specialised variation of the sensory modality of touch that encompasses the sensation of joint movement (kinesthesia), and joint position (joint position sense)” (Lephart, Pincivero, Giraido & Fu 1997, p. 130). Kinesthesia at the ankle is typically measured by passively moving the joint slowly within the normal range of movement to test whether or not the patient is able to detect the movement (Konradsen, 2002). Joint position sense is typically assessed by measuring reproduction of active and passive joint movements (Lephart et al., 1997). This is measured on either the ipsilateral or contralateral ankle after a practice trial where the examiner passively positions the foot, or instructs the patient to hold a position when moving actively (Konradsen, 2002). Whilst measures of kinesthesia and joint position sense are able to target specific joints and measure them in isolation (Boyle & Negus, 1998), the main criticism of these tests is that they are usually conducted in a non-weight bearing position, therefore offering limited ecological validity. As with muscular reaction time, no studies have been found to measure joint position sense on subjects with different foot structures.

2.7 Measuring Postural Stability

In order to measure the effects of foot structure on aspects of lower limb neuromuscular control, one must first review equipment, techniques and parameters used in current literature.

2.7.1 Equipment Used to Measure Postural Stability

Postural stability is commonly measured using a force plate which usually consists of a flat top plate supported by force transducers underneath; the force transducers produce an electrical signal proportional to any pressure applied to the top plate (Browne & O’Hare, 2000). Currently there are two types of force plates available; those with strain gauges (e.g. AMTI, Watertown, MA), or those with piezoelectric crystals (e.g. Kristler AG, Winterthur, Switzerland) (Robertson, Caldwell, Hamill, Kamen & Whittlesey, 2004). Modern force plates usually have four strain elements, located near the corners of the force plate enabling accurate measures across the entire surface of the force plate (Robertson et al., 2004). To improve ecological validity, Bartlett (2007) recommended covering the force plate with a surface relevant to the sport being studied, as the aluminium top plate of most force platforms may be unrepresentative of many sport surfaces. Force plates enable force measurements in three dimensions; vertical (Z), along the length of the plate (Y), and across the width (X) (Robertson et al., 2004). As the subject stands on the force plate, ground reaction forces (GRF) are measured which enables the centre of pressure (COP) to be calculated. Centre of pressure was defined by Winter (1995) as the point location of the vertical ground reaction force vector which represents a weighted average of all the pressures over the surface of the area in contact with the ground. The movement of the centre of pressure over a given time period provides an indirect measure of postural stability (Guskiewicz & Perrin, 1996; Hertel et al., 2002; Palmieri, Ingersoll, Stone & Krause, 2002). For example, an increase in plantar flexion moves the centre of pressure anteriorly and an increase in inverter activity moves the centre of pressure laterally (Winter, 1995).

As previously mentioned other methods such as the star excursion balance test have been used to measure postural stability (Cote et al., 2005), although this method may be more clinically applicable, it can be criticised for being unreliable. In addition it could be argued that the star excursion balance test is in fact a measure of lower limb flexibility rather than postural stability as it is quantified according to how far the subject can reach in a particular direction. In order to identify potential differences between subject groups with different foot structures, a more accurate and precise measure of postural stability is required and so the use of a more objective tool such as a force plate is advisable.

Bartlett (2007) advised that valid and reliable force plate measures rely on good sensitivity, low force detection threshold, high linearity, low hysteresis, low cross talk, and the elimination of cable interference, electrical inductance and temperature and humidity variations. Although most force plates are automatically calibrated according to manufacturer's guidelines, a lack of research has been conducted into the reliability of collected measurements. Russo and Chipchase (2001) conducted the only study found to have included reliability testing of a force plate before carrying out main testing. Plantar pressure measurements were taken over two consecutive days, the measurements were averaged, and then tested for differences; no significant differences were found between the two days. Browne and O'Hare (2000) recommended testing repeatability of force plate measurements by applying a load at least 20 times in the same position and recording the centre of pressure measurement each time. Measures should be precise to within 1 mm in order to detect changes in postural stability.

2.7.2 Static tests of Postural Stability

Many studies have used static measures when assessing postural stability (Eils & Rosenbaum, 2001; Hertel et al., 2002; Mitchell, Dyson, Hale & Abraham, 2008b) whereby the subjects are typically required to stand on a force plate either unilaterally or bilaterally for a specified duration and centre of pressure parameters are calculated. There is much variation with regards to how long the subjects are required to stand on the force plate to provide adequate data. Durations have ranged from five seconds (Goldie, Evans & Bach, 1992), 10 seconds (Hertel et al., 2002), 15 seconds (Eils & Rosenbaum, 2001), 25 seconds (Feuerbach & Grabiner, 1993) to 35 seconds (Mitchell et al., 2008b). Some studies use a buffer at either end of the trial, where the data is not recorded (Mitchell et al., 2008b); this is to remove excessive measures that may occur at the beginning or end of a trial. Le Clair and Riach (1996) indicated that shorter time periods produce less reliable results, with 20-30 seconds producing the most reliable results. In terms of application to sport, each of the previously mentioned time periods can be criticised for low ecological validity as rarely do athletes stand on a single limb for a prolonged period of time. Le Clair and Riach (1996) highlighted that only trials of the same time duration, using the same outcome parameters should be directly compared.

Movement in the medial-lateral direction is controlled by the subtalar joint, whereas the movement in the anterior-posterior direction is controlled by the talocrural joint (Eils & Rosenbaum, 2001). In static tests, several studies have shown medial-lateral postural sway to be significantly higher than anterior-posterior sway (Feuerbach & Grabiner, 1993; Mitchell et al., 2008b); Feuerbach and Grabiner (1993) suggested this is due to having a smaller base of support in the medial-lateral direction. Mitchell et al. (2008b) also noticed this trend, suggesting the majority of sway occurs at the subtalar joint, rather than the talocrural joint.

In order to place more challenging demands on the postural control system, several studies have attempted to reduce visual sensory input by measuring static postural stability when a subject has their eyes closed (Cote et al., 2005; Mitchell et al., 2008b). Whilst this does increase demands on neuromuscular control, it can be criticised for low ecological validity in terms of application to sport as there are few sports in which athletes are required to close their eyes. Other studies have attempted to challenge visual sensory input by showing simulations; Laurens et al. (2010) found that a stationary stimulus had a stabilising effect on postural stability; however a moving stimulus had a negative effect. Buckley, Amand, Scally and Elliott (2005) attempted to manipulate vestibular input by measuring postural stability in different head tilt positions; this showed that disruption to the vestibular system decreases postural stability.

Generally, static measures of postural stability can be criticised for low ecological validity as few sports require athletes to balance on one leg for a prolonged period of time. In addition, static tasks do not replicate the dynamic conditions in which sports injuries such as ankle sprains typically occur, therefore the results from such tests have limited use. Brunt et al. (1991) argued that performance on static postural sway tests may not be applicable to individuals experiencing instability during activity. It has been suggested that static tests only stimulate slow-adapting mechanoreceptors, whereas dynamic tests are more functional and stimulate fast-adapting mechanoreceptors (Wikstrom, Tillman, Chmielewski & Borsa, 2006). Therefore using static tests may not always be relevant, or demanding enough to identify postural stability deficits, particularly when testing subjects with functional ankle instability who may present with deficiencies when performing dynamic tasks (Wikstrom, Tillman, Chmielewski et al., 2006). Wikstrom, Tillman, Chmielewski et al. (2006) therefore supported the use of dynamic postural stability tests, as they are thought to be more challenging, and consequently may highlight potential deficiencies that static tests may be unable to detect.

2.7.3 Dynamic tests of Postural Stability

Rather than having subjects start on a force plate, several studies have attempted to make postural stability tasks more demanding by incorporating a jumping or hopping element whereby subjects have to maintain stability on landing (Wikstrom, Tillman, Schenker & Borsa 2008; Ross & Guskiewicz, 2004; Sell, 2012). Recently, Sell (2012) compared a static single leg stance to a dynamic forward hop and lateral hop over a small hurdle and found no correlation between results indicating that different responses are used to maintain postural stability in different situations; therefore it could be suggested that in order to elicit differences in healthy populations a more advanced task is required.

Wikstrom et al. (2008) used dynamic tests and calculated the postural stability indices for anterior-posterior, medial-lateral, and vertical planes during forward, diagonal, and lateral jumps, each starting 70 cm from the centre of the force plate. This study showed that jump direction significantly affects dynamic postural stability in frontal and vertical planes. In attempt to control for potential differences caused by differences in jump technique, Wikstrom et al. (2008) controlled the height of the jump by ensuring each subject jumped 50% of their maximum vertical jump height. In a similar study, Ross & Guskiewicz (2004) measured postural stability on landing from a 70 cm forward hop and ensured subjects jumped within 50 - 55% of maximum jump height, trials were excluded if subjects failed to jump within this range. It could be argued that ensuring subjects hit a marker during a jump will affect the jump technique by making the subjects look upwards which may affect their balance on landing, however this speculation is yet to be researched.

With dynamic tests which emphasise landing on a force plate, the duration for which subjects are required to maintain balance varies in the literature. Wikstrom et al. (2008) asked subjects to maintain balance for three seconds after landing from multi-directional jumps, whereas Ross & Guskiewicz (2004) asked subjects to maintain single leg stance for 20 seconds. Wikstrom, Tillman, Smith & Borsa (2005) suggested that a three second sampling interval should be used as it is more functional, imitating athletic activity. Whilst this is more representative of sport than the longer durations previously used, it could be argued that a shorter time period would be more applicable. There is some existing research into dynamic stability within the first 200 ms of landing which is of interest to researchers as it is beneath the level of conscious control (Madigan & Pidcoe, 2003; Suda, Amorium & Sacco, 2009). It has been suggested that a time frame of 150 – 200 ms is required in order to complete the stages of information processing and obtain sensorimotor feedback (Peterka, 2002; Schmidt & Lee, 2005). Peterka (2002) highlighted that quiet stance is regulated by high order control, which requires time for the nervous system to extract and combine information from various sensory resources to then generate a motor command. This is supported by Schmidt and Lee (2005) who stated that 150 – 200 ms is required to generate an error, detect the error, determine the correction, initiate the correction, and correct the movement. This time frame is also of interest to researchers as it has been suggested that impact absorption lasts approximately 150 – 200 ms, after which downwards momentum is substantially reduced (Lee, 1981). In addition, as most acute injuries such as ankle sprains occur during landing in a time frame beneath conscious control (Fong, Chan, Mok, Yung & Chan, 2009), it could be argued that postural stability during the first 200 ms after landing is of most importance.

In attempt to replicate the mechanism of several lower limb injuries including ankle sprains and anterior cruciate ligament sprains some studies have measured postural stability following landing from a drop jump (Hargrave et al., 2003; Riemann, Schmitz, Gale & McCaw, 2002; Yi, Brunt, Kim & Fiolkowski, 2003). The height of the jump has varied from 30 cm (Hargrave et al., 2003), to 40 cm (Yi et al., 2003), and 59 cm (Riemann et al., 2002). Whilst this method of testing postural stability is clearly much more demanding for healthy active populations in comparison to static postural control, it could be criticised that jumping from a height is not applicable to a wide range of sports during match play as jumps typically originate from ground level rather than from a height. It could be argued that it is directly relevant to aspects of training for example plyometrics as commonly used in a variety of sports; however a study of the injury rate in women's football indicated that athletes are 3.5 times less likely to be injured in training than in match play (Engström, Johansson & Tornkvist, 1991). This is likely to be because of the controlled and less competitive nature of training in comparison to match play. It is therefore thought that activities which directly replicate the demands of match play are most worthy of postural stability research.

Some studies have used more sport specific measures in attempt to replicate the demands of sport, and therefore the situations in which lower limb injuries are most likely to occur. Dayakidis and Boudolos (2006) found that subjects with functional ankle instability had greater first vertical force peak, and lower relative time to peak during a dynamic v-cut when compared with the contra-lateral stable joint. This cutting manoeuvre is directly relevant to the dynamic nature of many sports, in particular football where Bloomfield, Polman and O'Donoghue (2007) identified that the majority of turns performed were in a forwards direction either left or right between 0° to 90° during 90 minutes of match play. This summated to approximately 700 forward turns per match by defenders, 500 by midfielders and 600 by strikers.

Postural stability when walking or running forwards across a force plate has been well researched (Grundy, Tosh, McLeish & Smidt, 1974; Nachbauer & Nigg, 1992; Rodgers 1988). Rosenbaum and Becker (1997) highlighted several points to consider when measuring gait on a force platform; velocity, walkway length, acclimatisation to task, and selection of participants. Firstly, velocity is an important consideration as it has been shown that an increase in walking cadence results in decreased foot to floor contact time and increases in plantar pressures (Zhu, Wertsch, Harris & Alba, 1995). Rosenbaum and Becker (1997) suggested that controlling velocity with a metronome or with light gates is not always beneficial as it may distract the participant from normal gait causing them to over or under step in order to keep within time restrictions. Secondly, walkway length should also be considered to encourage normal gait for a number of steps both before and after the force plate (Rosenbaum & Becker, 1997). Thirdly, subjects should have adequate time to acclimatise to the laboratory settings in order to avoid slow and cautious steps across the force plate which are not representative of the participant's normal gait (Rosenbaum & Becker, 1997). Finally, Rosenbaum and Becker (1997) suggested that due to naturally high variation in gait pattern, subjects should be matched for age and foot structure.

In the Bloomfield et al. (2007) performance analysis of football match play, it was identified that defenders spend approximately 10% of total match play moving directly backwards. In a separate study, football referees were also found to spend a considerable amount of time backwards running; Caterall, Reilly, Atkinson and Coldwells (1993) calculated that an average of 18.2% of total distance covered during match play is spent moving backwards. The only study found to have linked backpedalling with injury is that of Wilson, Byrne and Gissane (2012) where it was suggested that the high incidence of Achilles tendinopathy among referees maybe due to the amount of time spent moving backwards due to excessive loading on the lower leg. In light of these interesting findings, it is surprising that very few studies have included back pedalling in postural stability analysis despite it being highly prevalent in some sports.

2.7.4 Parameters Used to Measures Postural Stability

There are a number of measurements that can be obtained from force plates, with no apparent consensus among the literature regarding which is most sensitive; or most suitable (Palmieri et al., 2002; Pavol, 2005). Some researchers prefer the use of ground reaction forces measures (Hargrave et al., 2003; Wikstrom et al., 2005), some use plantar pressure measures (Russo & Chipchase, 2001; D'Août, Pataky, DeClercq & Aert, 2010), and others have used centre of pressure measures (Eils & Rosenbaum, 2001; Hertel et al., 2002; Ross & Guskiewicz, 2004). Different measures make comparisons between studies difficult however it has been suggested that each of the different parameters may provide different information on the strategies used to maintain postural stability (Hunter & Hoffman, 2001; Le Clair & Riach, 1996).

Rosenbaum and Becker (1997) distinguish force as the interaction between two bodies or the body and its environment, whereas pressure is the measure that analyses the distribution of force across a surface area. A criticism of centre of pressure and ground reaction force calculations is that they do not identify specific anatomical locations within the foot. For example, as highlighted by Rodgers (1988), the forces recorded may occur underneath both the forefoot and rearfoot simultaneously causing the centre of pressure to fall at an intermediate point which may not actually be loaded. In order to gain more specific information about the loading of pressure distribution as it occurs, peak plantar pressure measures may be recorded using either a pressure plate or insole. Whilst the information does not directly measure stability, it gives an indication as to the distribution of pressure throughout the plantar surface of the foot which has wide applications and has been used when measuring footwear (D'Août et al., 2010), orthotics (Janin & Dupui, 2009), and taping (Russo & Chipchase, 2001).

Time to stabilisation is a common method of ground reaction force analysis which is defined as the time required to minimise ground reaction force on landing to within a baseline range (Wikstrom et al., 2005); a prolonged duration indicates reduced postural stability. Time to stabilisation has been criticised by Wikstrom et al. (2005) for giving three outcome measures of postural stability, vertical, medial-lateral and anterior-posterior, rather than one overall global measure of postural stability. Wikstrom et al. (2005) devised a new measurement known as the dynamic postural stability index (DPSI) which measures how well balance is maintained during the transition from a dynamic to static stance using ground reaction force measures. Similarly to time to stabilisation measures, the DPSI assesses the subject's ability to control ground reaction forces and maintain centre of gravity within the base of support, however, DPSI allows for multi-planar analysis of dynamic postural stability (Wikstrom et al., 2005). In a comparison of time to stabilisation and DPSI, Wikstrom et al. (2005) concluded that the DPSI was a more reliable and more precise measure of dynamic postural stability.

In terms of centre of pressure measures, sway amplitudes are commonly used as indicators of postural stability; these are one dimensional and so can be used to identify extreme centre of pressure movements in a particular direction (Palmieri et al., 2002). Mitchell et al. (2008b) suggested that individual postural sway components should be identified, so that rehabilitation can target specific deficits in a particular direction. Increased sway amplitude is thought to be indicative of reduced postural stability; Palmieri et al. (2002) suggested that maximum, minimum, and peak to peak amplitudes show high variability between trials and therefore advise the use of mean sway amplitude over other sway amplitude measures as it reduces error by taking in to account the average of multiple trials rather than considering only one data point. Sway amplitudes can be potentially affected by foot positioning on the force plate, however some force plates have settings which can account for different foot positioning and use centre of pressure average data point as the origin of the data rather than basing calculations on the force plate coordinates (for example AMTI, Watertown, MA).

Total distance, also referred to as total excursion or sway path, is an alternative way of measuring postural stability using centre of pressure. As shown in Figure 2.6 it can be calculated by summating the actual distance between successive centre of pressure data points (Hufschmidt, Dichgans, Mauritz & Hufschmidt, 1980). In current literature it is thought that an increase in total distance indicates reduced postural stability, however Palmieri et al. (2002) dispute this and claim that a large total excursion can be seen during stable stance, and a small total excursion can be seen during an instable stance. It seems likely that only valid comparisons can be made with trials of the same task and duration.

Figure 2.6 Total distance of centre of pressure adapted from Palmieri et al. (2002). (i) represents a single centre of pressure trial, (ii) represents the total distance of centre of pressure which is the same length as (i) if it was pulled apart.

Total velocity (or sway velocity), is another commonly used measure of postural stability. It takes into account centre of pressure measures in both the anterior-posterior direction and the medial-lateral direction over time, therefore giving an overall profile of postural control (Hunter & Hoffman, 2001; Palmieri et al., 2002). An increase in centre of pressure velocity indicates decreased postural stability, whereas a decrease in centre of pressure velocity represents an increase in postural stability (Le Clair & Riach, 1996; Palmieri et al., 2002). Both Hunter and Hoffman (2001) and Ross and Guskiewicz (2004) suggested that measures of maximum excursion and maximum velocity may not be as accurate as other postural stability measures as they may be influenced by a minority of excessive movements in a particular direction, therefore lacking in sensitivity and being unrepresentative of the entire test duration.

More recently, 95% confidence ellipse area measures have been used to measure postural stability using centre of pressure data; however a study by Rocchi, Sisti, Ditroilo, Calavalle and Panebianco (2005) criticised the use of this measure as it is often misinterpreted as “the area that contains 95% of all data points” (p.169). The correct definition of a 95% confidence ellipse area, according to Rocchi et al. (2005) is “the ellipse that, with 95% of probability contains the centre of the points of sway” (p. 169); therefore a smaller ellipse area represents better postural stability than a larger ellipse area. As yet, there is limited research on the reliability of this measure however Doyle, Hsiao-Wecksler, Ragan and Rosengren (2005) found evidence to suggest that a 95% confidence ellipse area has acceptable levels of reliability during measures of quiet stance at 90 seconds and 60 seconds, but not at 30 seconds. Among a population of injured subjects, Salavati et al. (2009) found that the 95% confidence ellipse area was not a reliable measure, and instead found mean total velocity to be most reliable. Clearly further research is required to establish the use of this measure in dynamic postural stability assessments.

Perhaps due to the ambiguity in selecting postural stability parameters, many researchers do not indicate why specific measures were chosen. Hertel et al. (2002) however stated that measures of centre of pressure velocity were used as it was most prevalent in other literature, and overall measures of centre of pressure excursion length were chosen as they were thought to provide a more robust measure of postural stability when compared with directional excursions.

2.8 Summary

Variations from a neutral structure of the medial longitudinal arch of the foot can lead to functional problems such as excessive pronation, or excessive supination. Although such foot structures may be asymptomatic, they have been shown to be associated with a range of lower limb injuries including ankle sprains, stress fractures, and knee pain. There are a number of ways to classify foot structure according to arch height; navicular drop height appears to be the most objective method and is commonly used in current literature enabling comparisons to previous research.

Several studies have associated differences in foot structure with deficits in some measures of lower limb neuromuscular control. Of these measures, postural stability has been most researched; however, current research has contradicting results leaving researchers undecided about the consequences of pronated or supinated feet on postural control. In addition, most studies have used static measures which are unrepresentative of sporting situations in which the majority of injuries associated with foot structure are most likely to occur. There is a clear demand for further research in this area to determine whether foot structure affects dynamic postural stability during tasks directly relevant to sport. If differences are found between pronated, supinated and neutral foot structures, recommendations can then be made to prevent potential injuries from occurring during sport.

Chapter Three

The Effects of Foot Structure on Dynamic Postural Stability

3.1 Introduction to Chapter

This chapter includes two pilot studies and Study One; The Effects of Foot Structure on Dynamic Postural Stability. In light of the review of literature (Chapter 2), it was intended that subjects in Study One would be split into three groups; neutral, pronated and supinated, using navicular drop height measures. Each subject would then undergo a series of postural stability tasks which mimic typical sporting movements in order to determine potential differences between both the pronated and supinated groups in comparison to the neutral group. Pilot testing therefore had to be carried out in order to determine the reliability of navicular drop height measures (Section 3.2), and of the postural stability tasks which were hoped to be included in Study One (Section 3.3).

3.2 Pilot Study One: Test Retest Reliability of Navicular Drop Height Measures

3.2.1 Abstract

Context: Navicular drop height is a well-established method of measuring neutral, pronated and supinated foot structures; however previous literature has highlighted concern over the reliability of this method.
Objective: To determine intra-tester reliability of navicular drop height measures on physically active subjects.
Subjects: Fifteen (six males, nine females); age = 20 ± 2.0 years, height = 171.1 ± 9.8 cm, mass = 66.9 ± 14.1 kg.
Methods: Three navicular drop height measures were taken by the same tester on two separate days using a height gauge. Intraclass correlation coefficient and standard error of measurement of navicular drop height measures were then calculated.
Results: Reliability was found to be very high over two days (ICC = .98; SEM = .5mm).
Conclusions: Due to very high measures of reliability which are comparable to those found in previous literature, it can be concluded that navicular drop height is a suitable method for classifying foot structure, and will therefore be used in Study One.

3.2.2 Introduction

As discussed in Section 2.5, navicular drop height is widely used to classify neutral, pronated and supinated foot structures; however there is some concern in previous literature about the reliability of the navicular drop height procedure (Mueller et al., 1993; Picciano et al., 1993;

Sell et al., 1994; Vinicombe et al, 2001). It was therefore decided that before commencing with Study One, navicular drop height must first be tested for reliability in order to determine whether it is a suitable measure for use.

3.2.3 Methods

Subjects:

Fifteen volunteers participated in this study; prior to testing, ethical approval was granted from the institutional ethics committee, and all subjects read a subject briefing document (Appendix One), provided written informed consent (Appendix Two) and completed a heath screen document (Appendix Three).

In order for the results of this study to be directly applicable to Study One, the same inclusion and exclusion criteria were used in this Pilot Study. Subjects were included in this study if they were aged between 18 - 30 years, and took part in at least two hours of exercise each week. Subjects were excluded if any of the following applied; recent lower limb injury (within six months), history of lower limb surgery, history of ankle sprains, myositis ossificans, poor circulation, general illness, acute trauma to lower limb, soft tissue inflammation, skin infection, allergy to alcohol wipes, were under the influence of alcohol or any other psycho-active substance, or had regular use of orthotics, taping or bracing.

Following recommendations by Tsai et al. (2006), to improve internal validity, subjects were also excluded if they took part in any activity involving regular balance training (e.g. ballet, gymnastics, Tai Chi) during the one year prior to testing, or for a total period of more than one year, during the 10 years prior to testing. Subject characteristics are shown in Table 3.1.

Table 3.1 Subject Characteristics

Sex	Age (years)	Height (cm)	Mass (kg)
6 Males 9 Females	20 (2.0)	171.1 (9.8)	66.9 (14.1)

Values are mean (SD).

Procedures:

Navicular drop height measurements were tested for reliability by analysing measurements by the same tester over two days. On each day, three measures were collected from each subject on the dominant foot (right foot for all subjects). In accordance with Hoffman, Schrader, Applegate and Koceja (1998) the dominant side was defined as the foot used to kick a ball. If subjects were unsure, or stated that either foot could be used to kick a ball, the dominant leg was further defined as the leg the subject would prefer to recover balance on if pushed.

All measures were taken using the same height gauge (Axminster HG-1, United Kingdom) and were collected following the same procedure as described by Brody (1982). Firstly, with the subject standing with feet shoulder width apart in a relaxed position, the most prominent point of the navicular tuberosity was marked with a pen. Secondly, the subtalar neutral position was found by pinching gently with the thumb and index finger either side of the talus, and asking the subject to slowly invert and evert their ankle until there was a feeling of equal pressure on both fingers. At this point the height of the marked navicular tuberosity from the ground was measured. Next, the subject was asked to return to a relaxed stance, and the height of the navicular tuberosity to the ground was measured again. The navicular drop height was then calculated by subtracting the relaxed stance measure, from the subtalar neutral measure.

Data Reduction and Statistical Analysis:

The average of the three measurements taken on each day was used for analysis. As data were normally distributed (Shapiro-Wilk P > .05) significant differences were tested using a paired t-test with the a priori alpha level set at .05. Correlations were then calculated using ICC and SEM. All statistical analyses were calculated using Excel (Microsoft Office, 2010).

3.2.4 Results

Results of test retest reliability measures are shown in Table 3.2. No significant differences were found. The mean navicular drop height measures were 6 mm on each day, resulting in a high ICC value.

Table 3.2 Test Retest Reliability of Navicular Drop Height Measurements

	Test	Retest	*P*	ICC	SEM
Navicular Drop Height (mm)	6.0 (3.0)	6.0 (3.0)	.55	.98	.5

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement. ICC > .70 indicating high correlation highlighted.

3.2.5 Discussion

The results indicate that navicular drop height measures are very highly correlated (Munro, 2000) when measured over different days by the same tester. The standard deviation of 3 mm on both days is seemingly high in comparison to the mean; the reason for this is due to the different foot structures among the fifteen subjects used in this pilot study, however as the standard deviations are the same over the two days it indicates that the same foot structures were identified over two days adding to the strength of the results of this study. The SEM of .5 mm is comparable to that of Carcia et al. (2006), who allowed testing to commence when the primary investigator had established a day to day intra-tester reliability of .90 ± .5 mm. Therefore it was concluded that navicular drop height was suitable to be used as a measure for classifying foot structure in Study One.

3.3 Pilot Study Two: Test Retest Reliability of Postural Stability Measurements

3.3.1 Abstract

Context: In previous literature, few studies have used dynamic movements which are directly applicable to sport. In addition, many previous studies have analysed postural stability over time periods which are too long to be representative of the dynamic nature of sporting movements.

Objective: To test reliability of dynamic movements over two separate days at two time frames; 200 ms to exclude voluntary responses and three seconds to be comparable to previous literature.

Subjects: Eight (five males, three females); age = 21 ± 1.7 years, height = 174.1 ± 10.9 cm, mass = 72.9 ± 16.1 kg.

Methods: Centre of pressure measures were taken over two days on seven dynamic movements; forward, diagonal and lateral hop, and drop jump, forward run, 45° cut and backpedal. Each movement was performed three times on each day. Centre of pressure parameters included peak, mean and total directional measures, total distance, average velocity, and 95% ellipse area; each were recorded at 200 ms and three seconds. Intraclass correlation coefficient and standard error of measurement of each parameter were then calculated.

Results: Test retest reliability of the seven dynamic movements showed that most ICC values were high (> .70) or moderate (> .50). Reliability was generally higher at 200 ms than

at three seconds, and was also generally higher during the hopping tasks rather than the drop jump, forward run, 45° cut and backpedal.

Conclusions: Despite low reliability on some measures, all tasks, time frames and parameters were decided to be included in Study One in order to provide the most comprehensive overview of postural stability during dynamic movements, particularly movements which have gained little attention in existing literature yet are highly prevalent in sport.

3.3.2 Introduction

As discussed in Section 2.7 there is a need for postural stability research in dynamic measurements which are directly applicable to sport. Seven tasks were identified for potential use in Study One; three hopping tasks as used by Wikstrom et al. (2008) (forward, diagonal and lateral), and four other dynamic movements including a 45° cut, a 30 cm drop jump, a backpedal as used by Wall et al., (2005) and a forward run. Pilot testing was conducted on these tasks in order to determine whether they were reliable measures and therefore suitable for use in Study One. This testing followed a procedure by Moghadam et al. (2011) who measured test retest reliability of different postural tasks in healthy adults over a variety of centre of pressure parameters.

3.3.3 Methods

<u>*Subjects:*</u>

Reliability was tested over two days using the same eight subjects on each day; subject characteristics are identified in Table 3.3. The inclusion and exclusion criteria were the same as those to be used in Study One, as identified in Section 3.2.3. Prior to testing, ethical approval was granted from the institutional ethics committee, and all subjects read a subject briefing document (Appendix One), provided written informed consent (Appendix Two) and completed a heath screen document (Appendix Three).

Table 3.3 Subject Characteristics

Sex	Age (years)	Height (cm)	Mass (kg)
5 Males 3 Females	21 (1.7)	174.1 (10.9)	72.9 (16.1)

Values are mean (SD).

Procedures:

All tasks were demonstrated to the subjects before testing, subjects then performed three practice trials of each task to ensure that they were performed correctly, and to ensure that each subject felt comfortable with the movements. All tasks were performed on the subject's dominant leg as defined in Section 3.2.3. The sequence of testing was randomised to reduce a possible fatigue effect.

Postural stability measures were collected using the same Advanced Mechanical Technology Inc. (AMTI) OR6-7 (Watertown, MA) force plate to be used in Study One. For the three hopping movements, markers were placed on the ground in front of the force platform prior to testing at exactly 70 cm from the centre (Figure 3.1). The hopping movements required the subjects to start with both feet on the marker, and jump with both legs, landing on only one in the centre of the force platform. Subjects were instructed to stabilise as quickly as possible and maintain balance for four seconds until they were instructed to stop. Arm position was not controlled for to encourage natural movement.

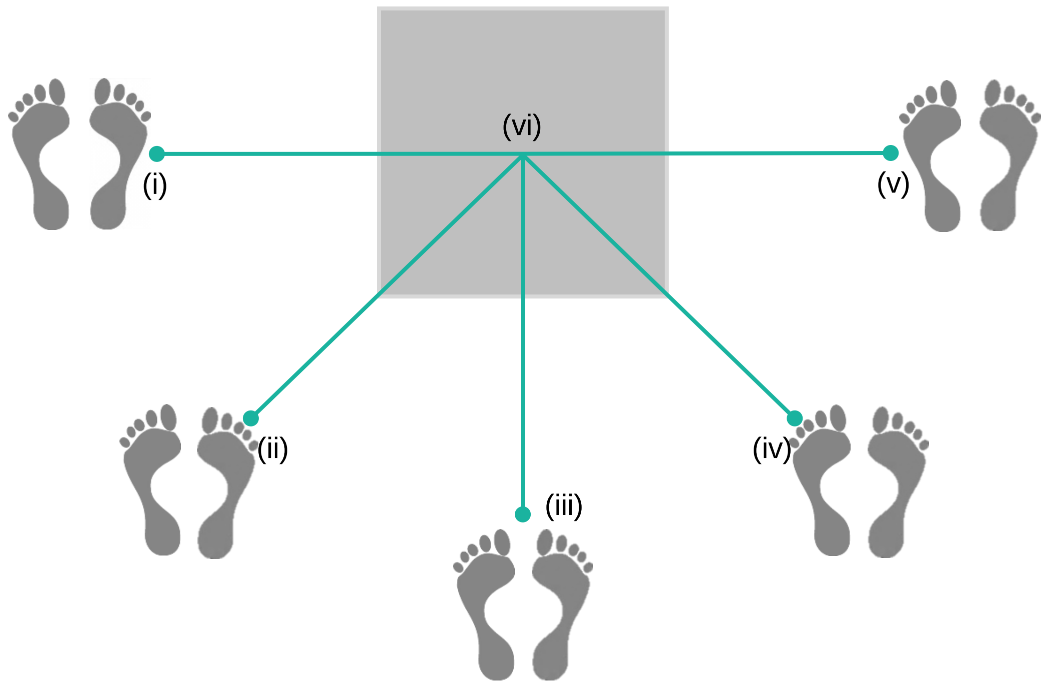


Figure 3.1 Set up for hopping movements, measured 70 cm from the centre of the force plate. (i) Starting position for a lateral hop landing on the right foot; (ii) starting position for a diagonal hop landing on the right foot; (iii) starting position for both left and right forward hops; (iv) starting position for diagonal hop landing on the left foot; (v) starting position for a lateral hop landing on the left foot; (vi) landing position for all tasks.

For the drop jump task, a 30 cm bench was positioned directly in front of the force plate and as with the hopping tasks, subjects were instructed to jump from the bench with two feet but land on only the dominant foot in the centre of the force plate and maintain balance until instructed to stop. For the forward run, 45° cut and backpedal; markers were placed on the floor to indicate starting locations as shown in Figure 3.2.

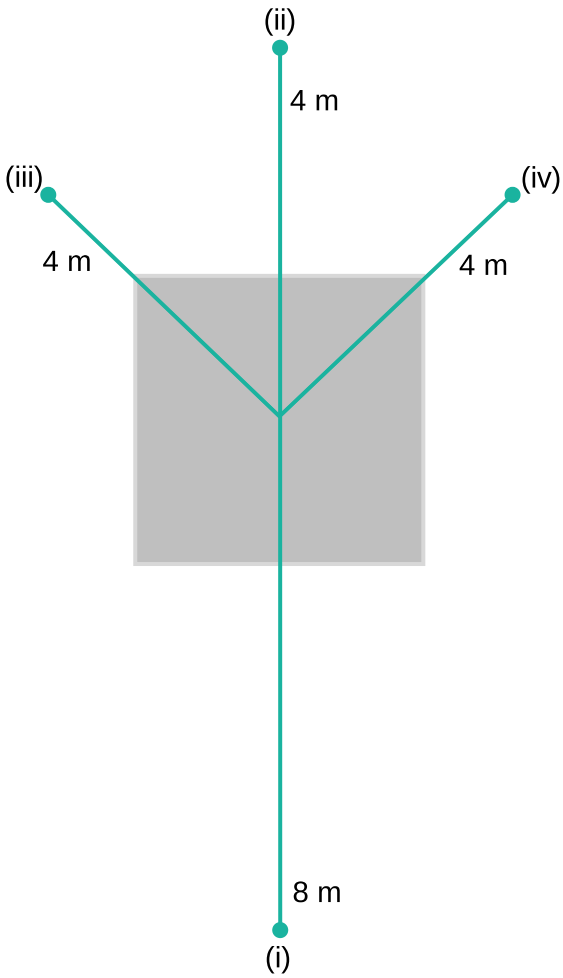


Figure 3.2 Set up for forward run, 45° cut and backpedal; (i) starting position for forward run and 45° cut, (ii) starting position for backpedal, (iii) direction of right footed 45° cut, (iv) direction of left footed 45° cut.

As each subject had varied stride length, the starting location for these tasks varied slightly between subjects to ensure that each subject landed on the dominant foot in the centre of the force platform on each trial; this was determined during practice trials. For the forward run the approximate starting location was 8 m posteriorly from the centre of the force plate and subjects were instructed to run directly over it and continue running for a further 4 m. The approximate starting location for the backpedal was 4 m anteriorly, and again subjects were instructed to continue posteriorly for approximately 4 m after crossing the force plate. The starting position for the 45° cut was the same as that for the forward run, however subjects were instructed to plant the dominant foot on the centre of the force plate, then push off in the opposite direction; for example if subjects planted the right foot on the force plate, they would cut off to the left. Again subjects were instructed to continue for 4 m after crossing the force plate towards a marker on the ground positioned at 45° from the centre of the force plate.

Data Reduction and Statistical Analysis:

Recording began as soon as the pressure on the force platform exceeded 50 N, and data were collected at 1000 HZ. Centre of pressure data were collected, and six variables were calculated using a custom made Excel template (Microsoft Office, 2010) (Appendix Four). Measurements included peak COP, mean COP, total COP, total distance, average velocity, and 95% ellipse area. Definitions of each COP variable are shown in Appendix Five.

For the hopping tasks and the drop jump, subjects were required to maintain balance for four seconds, however unknown to the subjects, only the first three seconds were recorded; this was to improve validity as some subjects may alter posture in anticipation of the end of the trial. For these tasks, each variable was calculated at the first 200 ms after landing to exclude voluntary responses, and at three seconds after landing to enable comparisons to previous literature.

For the forward run, 45° cut and backpedal; the duration of the trial was determined by the amount of time spent in contact with the force plate. This was not controlled using light gates which were thought to distract subjects from a natural gait pattern during each task, therefore compromising external validity (Bartlett, 1997). However as part of this reliability testing, it was analysed retrospectively according to the duration of the stance phase of gait as calculated automatically on BioAnalysis software (AMTI, Watertown, MA), this is shown in Figure 3.3.

As data were normally distributed (Shapiro-Wilk $P > .05$), paired t-tests were performed to test for significant differences between test and retest data with level of significance set at $P < .05$. No significant differences were found between test retest scores for all COP measures over all tasks, or during the retrospective analysis of the duration of stance during the dynamic movements. Intraclass correlation coefficients and SEM were then calculated to estimate both relative and absolute reliability. All statistical analyses were calculated using Excel (Microsoft Office, 2010).

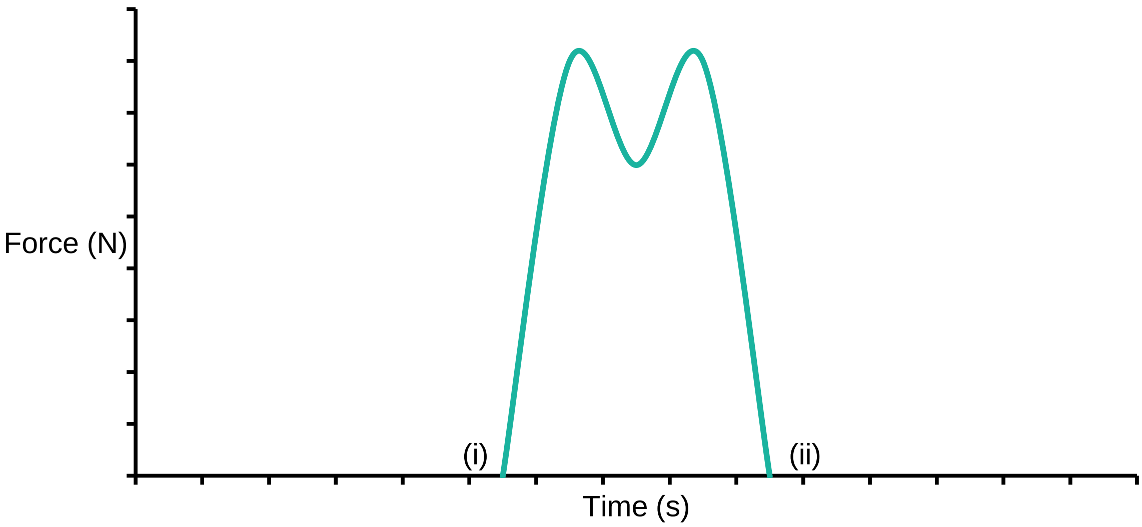


Figure 3.3 Representation of BioAnalysis software (AMTI, Watertown, MA) indicating automatic calculation of stance phase of gait as calculated by analysing time between heel strike (i) and toe off (ii).

3.3.4 Results

Results of the retrospective analysis of the duration of stance during the forward run, 45° cut and backpedal are shown in Table 3.4; high reliability is indicated for each task. Absolute and relative reliability of the seven dynamic movements are shown in Tables 3.5 and 3.6 showing the initial 200 ms and three seconds trial respectively. Raw data are shown in Appendix Six. Reliability of the seven tasks indicated that almost 50% of the ICC values are high (> .70), and a further 26% indicated moderate correlation (> .50) (Munro, 2000).

Table 3.4 Duration of Stance Phase for Forward Run 45° Cut and Backpedal

Task	Test	Retest	*P*	ICC	SEM
Forward Run (ms)	217 (25)	219 (20)	.68	.94	5.26
45° Cut (ms)	175 (32)	187 (33)	.13	.79	14.66
Backpedal (ms)	227 (27)	228 (26)	.88	.74	15.10

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement. ICC > .70 indicating high correlation highlighted.

Table 3.5 Test Retest Reliability of the Hopping and Drop Jump Tasks at 200 ms and Forward Run, 45° Cut and Backpedal

		Forward Hop		Diagonal Hop		Lateral Hop		Drop Jump		Forward Run		45° Cut		Backpedal	
		ICC	SEM	ICC	SEM	ICC	SEM	ICC	SEM	ICC	SEM	ICC	SEM	ICC	SEM
Peak (cm)	M	.77	0.41	.22	1.27	.33	0.60	.58	0.32	.53	0.59	.66	3.43	.02	5.79
	L	.19	0.40	.74	1.01	.86	0.96	.43	0.63	.54	0.79	.89	4.08	.16	4.71
	A	.75	1.01	.82	1.32	.67	0.62	.84	0.98	.21	4.09	.20	7.54	.28	7.55
	P	.76	1.68	.94	0.96	.72	0.70	.46	1.10	.21	4.61	.84	4.19	.21	11.19
Mean (cm)	M	.85	0.15	.66	0.22	.88	0.18	.18	0.19	.55	0.50	.52	0.00	.53	0.57
	L	.35	0.12	.85	0.25	.78	0.34	.25	0.23	.74	1.25	.30	7.53	.54	0.00
	A	.90	0.38	.89	0.35	.72	0.52	.82	0.47	.56	0.88	.15	1.94	.01	3.20
	P	.84	0.77	.81	0.69	.93	0.17	.42	0.57	.70	0.91	.53	2.25	.16	9.87
Total (cm)	M	.67	0.78	.34	2.11	.38	2.18	.55	2.14	.80	2.44	.99	2.71	.55	0.41
	L	.83	1.04	.67	1.38	.33	3.45	.35	2.49	.82	2.81	.98	3.86	.13	3.53
	A	.19	3.42	.73	2.24	.84	1.60	.55	2.28	.59	3.17	.96	0.97	.78	3.86
	P	.96	1.56	.92	1.85	.78	1.67	.44	2.23	.47	5.38	.75	3.29	.08	4.22
T.Dist (cm)		.97	1.73	.79	4.01	.90	3.32	.47	5.40	.67	9.46	.99	6.44	.92	3.99
V.Avg (cm/s)		.97	8.76	.79	20.09	.91	16.57	.47	26.95	.85	50.59	.99	18.21	.62	22.87
95% EA (cm^2)		.92	4.82	.63	15.26	.58	14.69	.39	6.97	.87	11.59	.99	5.03	.58	4.71

ICC: intraclass correlation coefficient; SEM: standard error of measurement; M: medial; L: lateral; A: anterior; P: posterior; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). ICC > .70 indicating high correlation highlighted.

Table 3.6 Test Retest Reliability of the Hopping and Drop Jump Tasks at Three Seconds

		Forward Hop		Diagonal Hop		Lateral Hop		Drop Jump	
		ICC	SEM	ICC	SEM	ICC	SEM	ICC	SEM
Peak (cm)	M	.51	0.57	.29	0.37	.36	0.47	.24	0.52
	L	.51	0.19	.77	0.81	.16	1.56	.34	0.49
	A	.67	0.92	.74	1.23	.65	0.65	.81	1.04
	P	.97	0.79	.85	1.56	.75	0.96	.10	1.10
Mean (cm)	M	.63	0.15	.03	0.18	.52	0.07	.04	0.22
	L	.31	0.27	.46	0.15	.38	0.34	.65	0.11
	A	.72	0.35	.50	0.30	.57	0.22	.37	0.63
	P	.41	0.24	.54	0.32	.68	0.40	.74	0.13
Total (cm)	M	.62	1.74	.66	2.18	.42	2.07	.76	1.98
	L	.84	1.25	.89	1.07	.96	0.97	.81	1.63
	A	.70	2.58	.66	2.91	.82	2.15	.74	1.73
	P	.92	1.97	.96	1.39	.85	2.25	.60	1.49
T.Dist (cm)		.96	2.34	.57	8.51	.38	9.83	.50	5.45
V.Avg (cm/s)		.96	0.78	.89	1.34	.92	1.34	.50	1.82
95% EA (cm^2)		.67	9.19	.67	8.27	.53	16.9	.74	4.18

ICC: intraclass correlation coefficient; SEM: standard error of measurement; M: medial; L: lateral; A: anterior; P: posterior; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). ICC > .70 indicating high correlation highlighted.

3.3.5 Discussion

Retrospective analysis of the duration of the forward run, 45° cut, and backpedal resulted in high reliability for each task (ICC > .70); the results are similar to those of Cornwall and McPoil (1997), who also found standard deviations of < 40 ms when measuring the consistency of walking gait trials with ICC = .89. Despite having the same amount of practice trials, the backpedal has the lowest test retest reliability (ICC = .74, SEM = 15.10) perhaps due to less familiarity with this task in comparison to the forward run and 45° cut during everyday activity. In light of this pilot study, the use of light gates were disregarded for Study One as aside from distracting subjects from natural gait, it was thought that even with the use of light gates, it would be difficult for subjects to maintain a more consistent speed than the results shown.

Analysis of the seven dynamic tasks indicated that the lowest values were generally during the peak, mean and total directional measures, whereas the highest were usually within the total distance and average velocity measures. This is similar to the findings by Wikstrom et al. (2006), who whilst using different measures of postural stability, found lower reliability scores for directional components, and higher reliability scores for the DPSI which includes all directional components. Using a non-sporting population, Moghadam (2011), found that test retest reliability of average velocity was consistently high across all tasks, and interestingly medial and lateral directional measures were consistently higher than anterior and posterior measures. In comparison to the current study, average velocity is high (ICC > .70) in all hopping tasks at both time frames, and is also high for the forward run and 45° cut. In the current study there is no obvious pattern between medial and lateral, and anterior and posterior measures.

The results of the forward hop task are comparable to those of Birmingham (2000), who also found high correlations (ICC = .85) using total distance as a measure of COP. The ICC of total distance measured in this pilot study was .97 at 200 ms and .96 at three seconds. Intraclass correlation coefficient values of the hopping tasks appear to be higher during the initial 200 ms rather than at three seconds for each task. As no other studies were found to have analysed postural stability within this time frame it is difficult to make comparisons, however as 200 ms is beneath the level of conscious control this is an interesting finding suggesting that the subconscious response to maintaining postural stability is similar to a conscious response.

Average velocity and 95% ellipse area measures indicated very high correlation (> .85) for the forward run and 45° cut, and indicated moderate correlation (> .50) for the drop jump and backpedal tasks. The backpedal tasks showed least overall consistency over two days which was to be expected due to subjects having less familiarity of this task in comparison to the other tasks. Despite low reliability on some measures, this task was still included in Study One as a matter of interest, however in light of this pilot testing; results should be treated with caution. In addition, as a result of this pilot testing, in attempt to increase the reliability during Study One, subjects were to be given extra time to practice to help overcome any caution they may have felt when performing the backpedal task.

In general, it appears that the more dynamic the movement, the less consistent the results. It is perhaps due to this reason that many researchers tend to use static measures of postural stability rather than observing dynamic movements. Due to the lack of research in this area with regards to foot structure, and considering that sports injuries such as ankle sprains occur during dynamic movement rather than when maintaining a static posture, it was still thought to be worthwhile to pursue the research into dynamic postural stability during all tasks. In light of this pilot testing, it seems that a combination of dynamic measures should be used to provide a more accurate overview of postural stability.

3.4 Study One: The Effects of Foot Structure on Dynamic Postural Stability

3.4.1 Abstract

Context: Despite an association between pronated and supinated foot structures and the incidence of lower limb injury, few studies have measured dynamic postural stability on subjects with neutral, pronated and supinated feet.

Objective: To determine whether pronated and supinated foot structures contribute to neuromuscular deficits as measured by dynamic postural stability.

Subjects: Thirty two volunteers were categorised into three groups according to navicular drop height measures. Thirteen subjects had neutral feet (navicular drop height 5 – 9 mm), ten subjects had pronated feet (navicular drop height ≥ 10 mm), and nine subjects had supinated feet (navicular drop height ≤ 4 mm).

Methods: Three hopping tasks; forward, diagonal and lateral, and four other dynamic tasks; forward run, 45° cut, drop jump and backpedal were performed by each subject three times across a force platform. Centre of pressure parameters including peak, mean and total directional measures, total distance, average velocity, and 95% ellipse area were calculated. Each were analysed at 200 ms and three seconds.

Results: On a number of tasks, those with pronated and supinated foot structures had reduced postural stability in comparison to those with neutral feet. Notably at 200 ms, during the diagonal hop the supinated group had increased average velocity (P = .01); during the lateral hop the supinated group had significantly higher peak medial COP (P = .01) and the pronated group had increased peak posterior COP (P = .04). During the 45° cut, the pronated group had increased mean anterior COP (P = .01); during the drop jump at 200 ms the pronated group had increased 95% ellipse area (P = .04), whereas at three seconds, the supinated group had increased 95% ellipse area (P = .01).

Conclusions: This study indicates that foot structure influences postural stability on a number of dynamic tasks. Future research should aim to explore this further in attempt to establish the consequences of pronated and supinated foot structures on dynamic postural control. In addition, researchers should be cautious of grouping together subjects with different foot structures during postural stability tasks as they may interfere with results.

3.4.2 Introduction

Medial longitudinal arch height is related to foot function; a high arch is typically rigid and a characteristic of excessive supination, whereas a low arch is usually hypermobile and is related to excessive pronation (Franco, 1987). Pronated and supinated foot structures are frequently associated with lower limb injuries; these include lateral ankle sprains (Mei-Dan et al., 2005; Williams et al., 2001) and anterior cruciate ligament tears (Lutter, 1980). Previous studies have associated excessively pronated and supinated foot structures with deficits in some aspects of neuromuscular control in comparison to those with neutral feet (Cote et al., 2005; Tsai et al., 2006). Components of neuromuscular control include proprioception, muscle strength, muscle reaction time and postural control (Richie, 2001).

The measurement of postural stability is an established method of analysing neuromuscular control. However most previous studies have used static measures (Cote et al., 2005; Hertel et al., 2002; Mitchell et al., 2008b) where typically, subjects are required to maintain balance in a single limb stance for between 10 and 35 seconds (Eils & Rosenbaum, 2001; Feuerbach & Grabiner, 1993; Hertel et al., 2002; Mitchell et al., 2008b). This task and time frame have limited external validity as during sport where the majority of acute lower limb injuries occur, movements are typically more dynamic and much faster. Some previous studies have measured the effects of foot structure on dynamic tasks such as a forward hop (Wikstrom et al., 2008; Ross & Guskiewicz, 2004), however it has previously been suggested that landing in the sagittal plane may not be demanding enough on neuromuscular control to elicit differences among different subjects groups (Wikstrom, Tillman, Chmielewski et al., 2006).

The purpose of this study was to determine whether pronated and supinated foot structures cause neuromuscular deficits as measured by dynamic postural stability. It was hypothesised that subjects with pronated and supinated foot structures would have significantly higher postural stability parameters indicating reduced postural stability in comparison to those with neutral foot structures. This was expected to occur on all seven tasks; three hopping movements (forward, diagonal and lateral), and a forward run, 45° cut, drop jump and backpedal.

3.4.3 Methods

<u>Subjects:</u>

Thirty two volunteers participated in this study and were categorised into three groups; neutral, pronated and supinated, which were dependant on navicular drop height measures. Navicular drop height was measured using the same procedure as shown to be reliable in Section 3.2. Prior to testing, ethical approval was granted from the institutional ethics committee, and all subjects read a subject briefing document (Appendix One), provided written informed consent (Appendix Two) and completed a heath screen form (Appendix

Three). The same inclusion and exclusion criteria were used as identified in Section 3.2.3. Subject characteristics are shown in Table 3.7.

Table 3.7 Subject Characteristics

Group	Sex	Age (years)	Height (cm)	Mass (kg)	Navicular Drop Height (mm)
Neutral	4 Males 9 Females	21 (4.0)	170.5 (9.1)	66.0 (10.3)	7.0 (1.4)
Pronated	6 Males 4 Females	19 (1.8)	172.2 (11.1)	72.8 (14.6)	11.5 (1.5)
Supinated	5 Males 4 Females	21 (3.1)	172.3 (9.3)	69.5 (12.4)	1.9 (1.1)

Values are mean (SD).

Procedures:

Using the same AMTI OR6-7 (Watertown, MA) force plate, the method as identified in Section 3.3.3 was also used in this study, however subjects did not repeat the testing on a second day. As discussed in Section 3.3.3, for the hopping tasks and the drop jump, subject were instructed to remain balance for four seconds, although only the first three were recorded. The duration of the forward run, 45° cut and backpedal were analysed retrospectively according to Bio-Analysis software as found to be reliable in Section 3.3.

Data Reduction and Statistical Analysis:

Recording began as soon as the pressure on the force platform exceeded 50 N, and data were collected at 1000 HZ. Centre of pressure data were collected, and six variables were calculated at two time frames (200 ms and three seconds) including peak COP, mean COP, total COP, total distance, average velocity, and 95% ellipse area using a custom made Excel template (Microsoft Office, 2010) (Appendix Four). Definitions of each COP variable are shown in Appendix Five.

As data were normally distributed (Shapiro-Wilk P > .05) for each task a separate one-way analysis of variance was performed for each dependant variable (peak COP, mean COP, total COP, total distance, average velocity, and 95% ellipse area) with foot structure (neutral, pronated and supinated) as the independent variable. Where a significant main effect was observed, a Dunnett's Post Hoc test was used to compare pronated and supinated foot structures against the neutral foot group. The a priori alpha level was set at .05. Effect size (η_p^2 values) was calculated where 0.01 - 0.06 indicated a small effect, 0.06 - 0.14 indicated a medium effect, and > 0.14 indicated a large effect. Observed power was also calculated. All statistical analyses were performed using IBM SPSS Statistics version 19.0 (SPSS Inc., Chicago, IL).

3.4.4 Results

At 200 ms, significant differences were present in the diagonal hop, the lateral hop and drop jump tasks. Differences were also present during the 45° cut and at three seconds significant differences were present in the drop jump task. No significant differences were present for the forward, diagonal or lateral hop at three seconds, or for the backpedal and forward run. Due to the number of results, only significant differences are shown here. Raw data for all tasks are shown in Appendix Seven.

As shown in Figure 3.4, analysis of the average velocity during the diagonal hop at 200 ms indicated a significant main effect of foot structure $F(2, 29) = 3.58$, $P = .04$, $\eta_p^2 = .19$, observed power = .61. Post hoc testing revealed a significant difference between the neutral group and the supinated group ($P = .01$) with the supinated group showing increased average velocity in comparison to the neutral group.

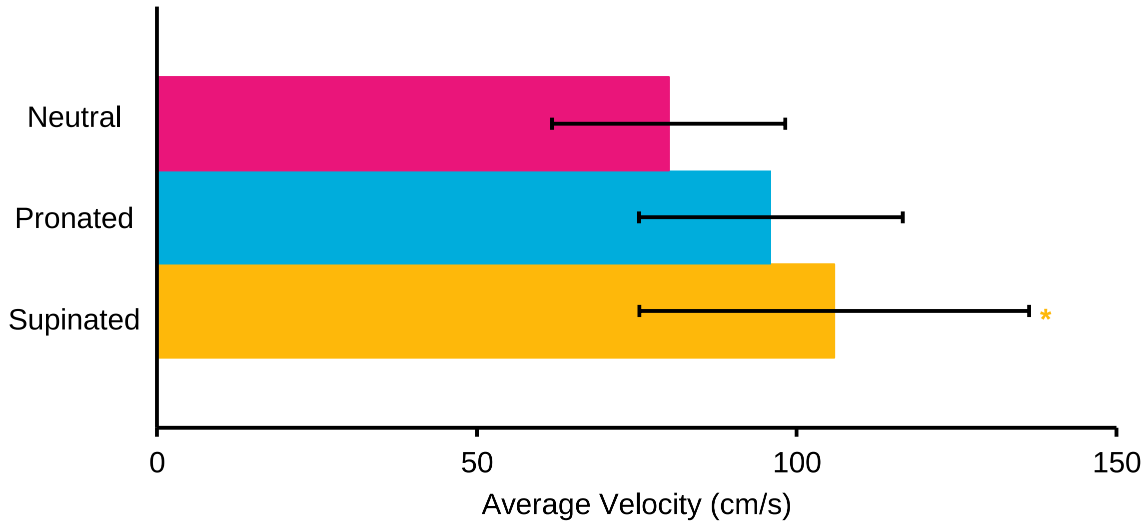


Figure 3.4 Average velocity of centre of pressure (cm/s) during diagonal hopping task at 200 ms; showing standard deviation. * Indicates significant increase in comparison to the neutral group ($P < .05$).

Analysis of the peak medial COP during the lateral hop at 200 ms indicated a significant main effect of foot structure $F(2, 29) = 4.19$, $P = .02$, $\eta_p^2 = .22$, observed power = .69. Post hoc testing revealed a significant difference between the neutral group and the supinated group (*P* = .01). As shown in Figure 3.5, the supinated group had increased peak medial COP in comparison to the neutral group. Additionally, during the same task, analysis of peak posterior COP indicated a significant main effect of foot structure *F*(2, 29) = 4.56, *P* = .01, η_p^2 = .23, observed power = .73. Post hoc testing revealed a significant difference between the neutral group and the pronated group (*P* = .04) with the pronated group showing increased peak posterior COP in comparison to the neutral group. This is also shown in Figure 3.5.

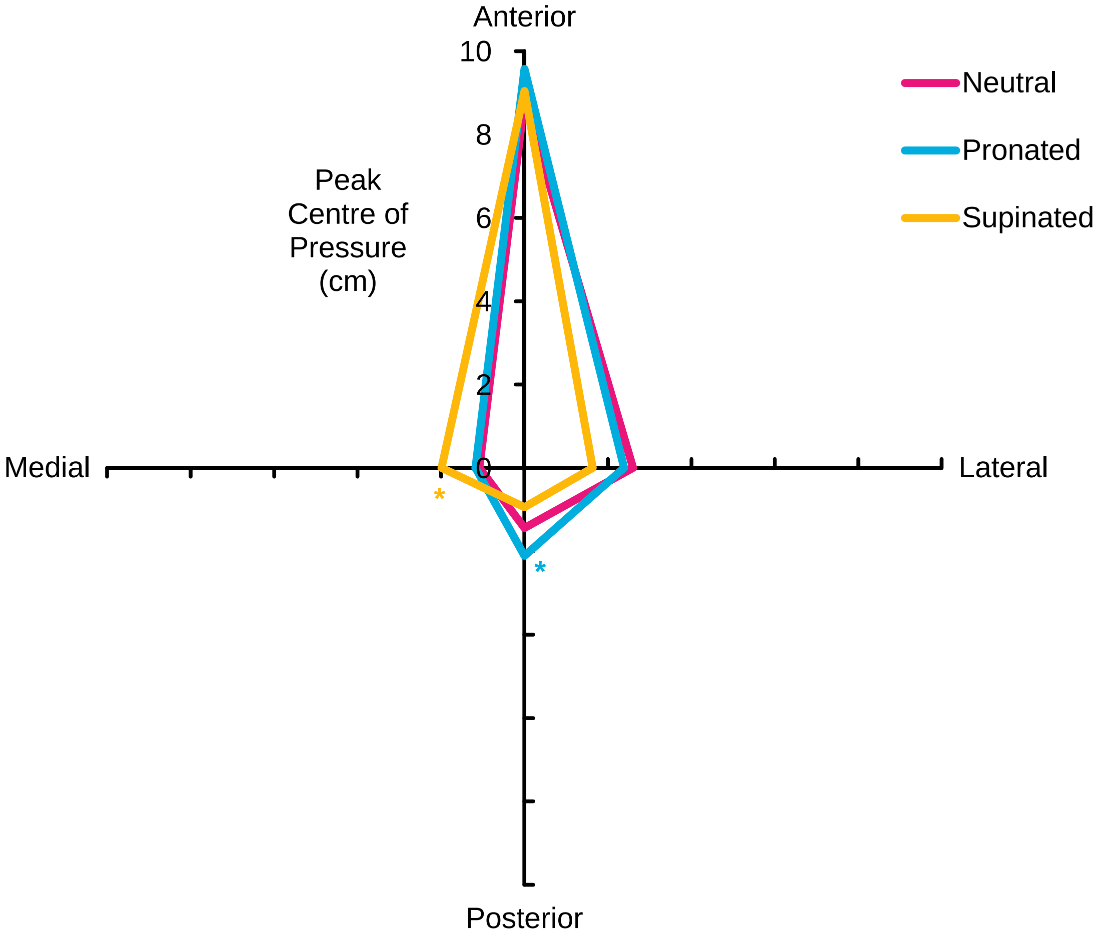


Figure 3.5 Peak anterior, lateral, posterior and medial centre of pressure (cm) during lateral hop at 200 ms. ** Indicates significant increase in comparison to neutral group (*P* < .05).

During the drop jump at 200 ms, the 95% ellipse area indicated a main effect of foot structure *F*(2, 29) = 3.16, *P* = .5, η_p^2 = .17, observed power = .56. Post hoc testing revealed a significant difference between the neutral group and the pronated group (*P* = .04). As shown in Figure 3.6, the pronated group had an increased ellipse area in comparison to the neutral group. During the 45° cut, analysis of the mean anterior COP indicated a significant main effect of foot structure *F*(2, 29) = 4.01, *P* = .02, η_p^2 = .22, observed power = .68. Post hoc testing revealed a significant difference between the neutral group and the pronated group (*P*

= .01). As shown in Figure 3.7, the pronated group had increased mean anterior COP in comparison to the neutral group.

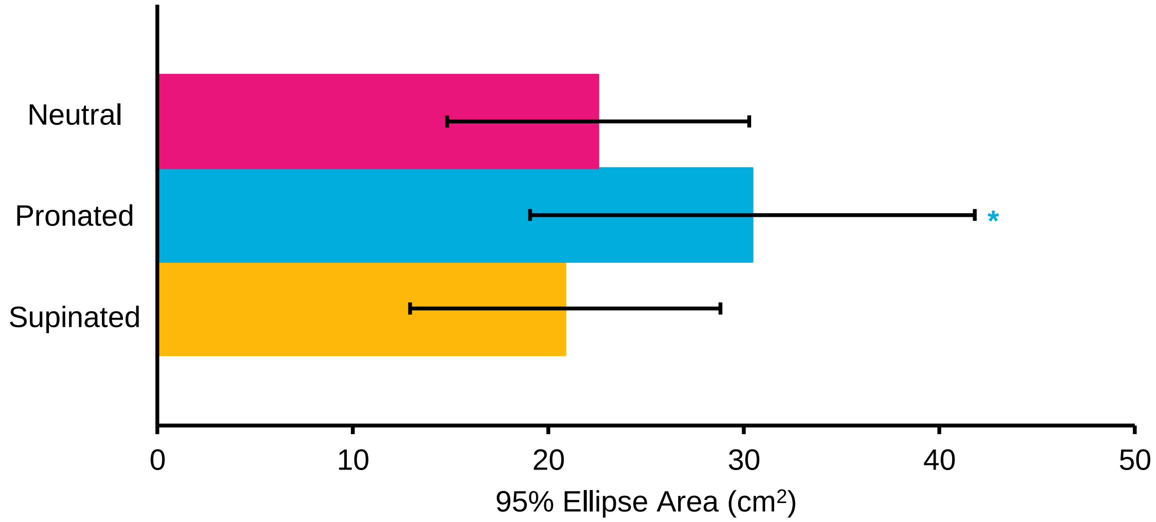


Figure 3.6 95% Ellipse area (cm^2) during drop jump at 200 ms; showing standard deviation. * Indicates significant increase in comparison to neutral group (*P* < .05).

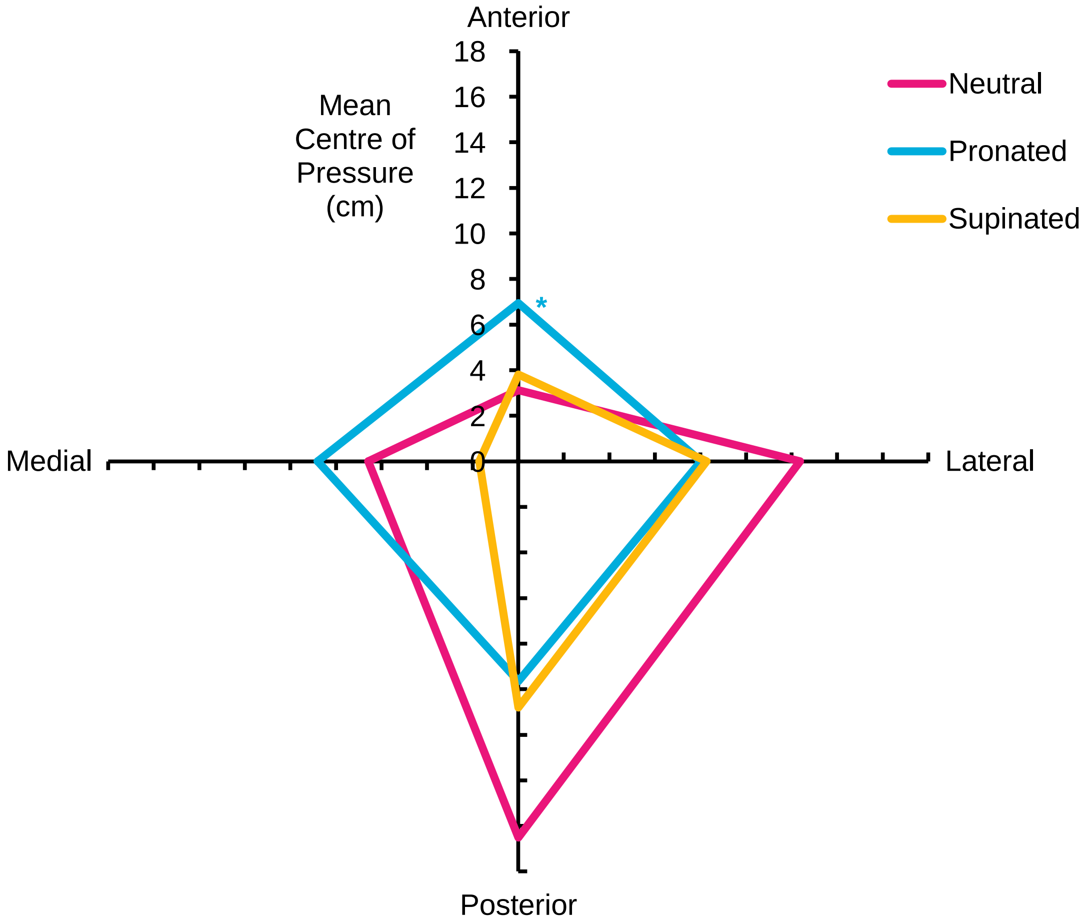


Figure 3.7 Mean anterior, lateral, posterior and medial centre of pressure (cm) during 45° cut. * Indicates significant increase in comparison to neutral group (*P* < .05).

During the drop jump at three seconds, analysis of the 95% ellipse area indicated a significant main effect of foot structure $F(2, 29) = 13.65$, $P = .01$, $\eta_p^2 = .48$, observed power = .96. Post hoc testing revealed a significant difference between the neutral group and the supinated group ($P = .01$) with the supinated group showing an increased ellipse area in comparison to the neutral group, this is shown in Figure 3.8.

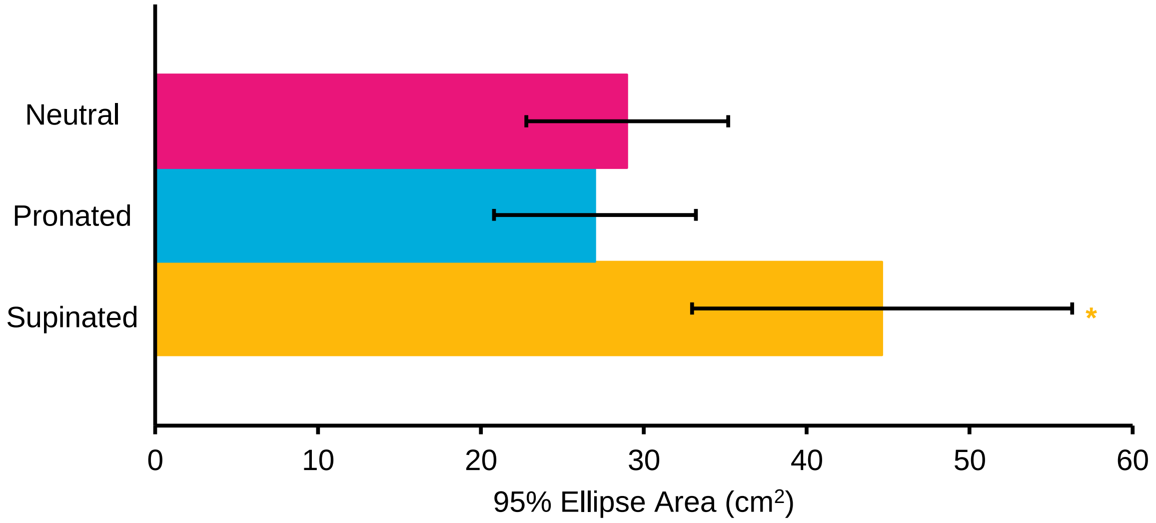


Figure 3.8 95% ellipse area during drop jump (cm^2); showing standard deviation. * Indicates significant increase in comparison to neutral group ($P < .05$).

3.4.5 Discussion

This is the first study to have measured dynamic postural stability during tasks representative of sport on subjects with pronated, supinated and neutral foot structures. It was hypothesised that subjects with pronated and supinated foot structures would have significantly higher postural stability parameters indicating reduced postural stability in comparison to those with neutral foot structures. This was expected to occur on all seven tasks; forward, diagonal and lateral hops, and forward run, 45° cut, drop jump and backpedal. The hypothesis is therefore partially accepted, as postural stability was shown to be reduced in the pronated and supinated groups on some dynamic tasks.

Aside from using dynamic tasks applicable to sport to measure postural stability, a further unique aspect of this study is that parameters were calculated at two time periods; 200 ms and three seconds. Parameters were analysed at three seconds in order to enable comparison of results to Wikstrom et al. (2008) where the same hopping tasks were also measured at three seconds. However despite having no studies available for comparison, it is thought that the 200 ms time frame may provide more useful information in determining postural stability deficits among subjects with different foot structures. The reason for interest in this time frame is that in addition being beneath the level of conscious control (Suda et al., 2009; Madigan & Pidcoe, 2003), it is also within the time frame that injuries such as ankle sprains occur (Fong et al., 2009); therefore in terms of sports injury prevention, it may be of more use to researchers.

Interestingly in this study, most differences were found at 200 ms, rather than at three seconds. For the diagonal and lateral hopping tasks, differences between groups were present at 200 ms but not at three seconds. This indicates that postural stability in those with pronated and supinated feet becomes equal to those with neutral feet between 200 ms and three seconds on these tasks. This cannot be said for all tasks, as during the drop jump task differences were present between the neutral and pronated groups at 200 ms and between the neutral and supinated groups at three seconds.

The findings of reduced postural stability in pronated and supinated foot structures at 200 ms is important as it indicates that these groups may be more vulnerable to injuries such as ankle sprains on landing in comparison to subjects with neutral feet, however as the incidence of ankle sprains were not measured in this study this is yet to be established. In addition, during this study subjects landed on a flat stable platform, so it is possible that the postural deficits indicated here may be amplified if measured on an uneven ground such as a football or rugby pitch; however again this speculation is yet to be researched.

Cote et al. (2005) conducted the only other study which claimed to have measured dynamic postural stability on subjects with pronated and supinated foot structures; however the star excursion balance test was used which can be criticised for not being a true dynamic test of stability. Cote et al. (2005) found that stability differed among groups but only in some directions. Comparison with this study is difficult due to vast differences in the star excursion balance test and the dynamic tasks used in this study; however the only consistency with the results of Cote et al. (2005) was that the pronated group showed increased mean anterior movement in the 45° cut. This was not found during any other task. The supinated group in the current study showed increased peak medial centre of pressure during the diagonal hop, whereas in the study by Cote et al. (2005), the supinated group showed more movement posteriorly and laterally. In addition, Cote et al. (2005) found that the pronated group had more movement anteriorly and medially, whereas the current study observed more movement in the posterior direction for the lateral hop.

Four other studies were found to have measured postural stability on subjects with pronated and supinated feet; however all of them used static tests which therefore mean that results cannot be applied to the dynamic nature of sport. During single leg stance, Hertel et al. (2002) found that those with cavus feet had reduced postural stability in comparison to those with neutral feet; however no differences were identified between the pronated and neutral groups. In contrast Cobb et al. (2004) found differences between the pronated group and the neutral group, but no differences between the supinated and neutral group. Tsai et al. (2006) found that both the pronated and supinated groups had poorer postural stability than the neutral group, and Cote et al. (2005) found differences between the pronated and supinated groups, but neither were significantly different to the neutral group. In the current study, pronated and supinated foot structures responded differently to each task; the supinated group indicated poorer postural stability than the neutral group in the diagonal hop, lateral hop and drop jump at three seconds, whereas the pronated group showed poorer postural control than the neutral group during the lateral hop, drop jump at 200 ms and during the 45° cut.

Whilst Ross and Guskiewicz (2004) and Wikstrom et al. (2008) used hopping tasks similar to that used in this study, neither study used subjects with different foot structures, and both studies used different parameters (time to stabilisation and DPSI respectively) to those used in this study which therefore makes comparison difficult. Ross and Guskiewicz (2004) used subjects with functional ankle instability and results indicated that differences in postural stability between different groups can be identified during a forward hop. In the current study, no significant differences were present in the forward hop task. Perhaps the reason for not finding any differences during the forward hop task in this study is that subjects with pronated or supinated foot structures have better postural control in comparison to subjects with functional ankle instability. This remains speculation as no study has been found to compare postural stability in subjects with functional ankle instability and subjects with pronated and supinated foot structures.

Wikstrom, Tillman, Chmielewski et al. (2006) previously suggested that a forward hop task is not demanding enough on the postural control system to elicit differences between different subject groups; this is supported by Sell et al. (2011), who also stated that more dynamic measures of postural control are a better tool for analysing risk factors for ankle and knee injuries as they pose greater challenge than static tasks. In the current study, in addition to no differences being found during the forward hop task, no differences were identified in the backpedal or forward run tasks either. Whilst the latter tasks are clearly more dynamic than the forward hop, all of these movements are in the anterior - posterior plane. In the current study, differences were only identified on tasks incorporating diagonal and lateral movements. This has interesting implications in terms of injury as ankle sprains often occur during landing from lateral and diagonal movements due to increased stress on the subtalar joint (Wright, Neptune, van den Bogert & Nigg, 2000). Increased instability at the subtalar joint suggests that individuals with pronated and supinated foot structures may be more at risk of obtaining a lateral ankle sprain in comparison to individuals with neutral feet.

When considering the structures of pronated and supinated feet, it is unsurprising that most differences were found in diagonal and lateral movements rather than anterior and posterior tasks. It has previously been suggested that rigid supinated feet are less able to adapt to unstable surfaces (Franco, 1987) therefore perhaps requiring a larger compensatory movement in order to maintain balance; this could be true for the lateral hop at 200 ms where the supinated group showed reduced postural stability as shown by increased peak medial centre of pressure. Perhaps due to the rigid structure of a supinated foot, a more emphasised medial movement was required in order to control balance to counteract the lateral movement of the hop in comparison to the neutral foot. Previous researchers have theorised that the reason those with pronated feet have difficulty in maintaining postural stability may be due to excessive mobility at the subtalar joint (Cobb et al., 2004). Alongside the clear structural differences between foot types, it has been suggested that differences in cutaneous input may also affect postural stability (Hertel et al., 2002). This may provide a reason for reduced postural stability in the supinated group across some tasks, however does not explain differences in the pronated group. Further research is required to determine the cause of reduced postural stability in pronated and supinated foot structures.

3.4.6 Limitations

Comparisons to other studies are difficult as few others have used dynamic tasks similar to those used in this study; a potential reason for this is that dynamic measures are inevitably harder to control, and therefore are likely to have more variability among subjects than more static tasks. This was identified during pilot testing where the backpedal task was found to be least reliable, but it was decided to be included in this study due to having a high occurrence in sport and also due to a vast lack of existing research. A further difficulty in postural stability research is the lack of consistent measures among studies. In the current study, a range of parameters were included in order to provide the most detailed overview of postural stability for each task.

3.4.7 Recommendations for Future Research

Future research should aim to establish definitive parameters across postural stability research in order to make more useful comparisons between studies. Additionally, as a result of this study, future research should aim to analyse postural stability on subjects with different foot structures within the first 200 ms of landing as it is within this time frame that injuries such as ankle sprains typically occur. In addition, it is beneath the level of conscious control, and was the time frame in which most differences were found between groups within this study.

3.4.8 Conclusion

The results of this study suggest that pronated and supinated foot structures affect postural stability in some tasks; specifically during lateral and diagonal movements. In light of this research, further research should be conducted into measuring the effects of foot structure on neuromuscular control in order to determine whether those with pronated or supinated foot structures are more vulnerable to injury during sport than those with neutral foot structures. In light of this study, it is advisable that researchers and clinicians alike are aware of grouping together subjects with different foot structures when taking postural stability measures as differences between foot structures may interfere with results.

3.5 Development of Research

The work in this chapter has shown that navicular drop height is a reliable tool which can be used to classify subjects into foot structure groups. In addition, in terms of postural stability, it has shown that more dynamic movements provide less consistent results, however due to the lack of research in current literature regarding dynamic movement and foot structure; it was still thought to be worthwhile to pursue. In addition, Study One showed that pronated and supinated foot structures do affect postural stability in a number of tasks. The results of this study lead to a further question: are other aspects of neuromuscular control affected by foot structure? As discussed briefly in Section 2.6.2., a further aspect of neuromuscular control that has received little attention in the literature with regards to foot structure is muscle reaction time. Considering the clear links between foot structure and lower limb injuries such as ankle sprains as discussed in Section 2.3, this is highly surprising. Chapter Four therefore aimed to examine the effects of foot structure on muscular reaction time to a tilt platform perturbation.

Chapter Four

The Effects of Foot Structure on Muscular Reaction Time

4.1 Introduction to Chapter

This chapter includes Study Two; The Effects of Foot Structure on Muscular Reaction Time. This involved dividing subjects in to three groups; neutral, pronated and supinated, using navicular drop height measures. Muscular reaction time to a tilt platform perturbation was then measured for each group. Before conducting Study Two, a thorough review of literature was undertaken in order to examine current measurement and analysis techniques, following this, pilot testing was undertaken in order to determine the reliability of proposed techniques to be used in Study Two.

4.2 Review of Literature on Muscular Reaction Time

A number of search engines were used to acquire the relevant literature in this review including Scopus, Web of Knowledge and Google Scholar. Typical terms used in the search included *muscular reaction time*, *neuromuscular control*, *onset detection* and *tilt platform*.

4.2.1 The Dynamic Defence Mechanism

The measurement of muscular reaction time to a tilt platform perturbation is a well-established method of analysing lower limb neuromuscular control (Benesch, Pütz, Rosenbaum & Becker, 2000; Eils & Rosenbaum, 2001; Lohrer, Alt & Gollhofer, 1999). Tilt platforms are designed to simulate an inversion ankle sprain, therefore stressing the dynamic defence mechanism. First coined by Konradsen et al. (1997), the dynamic defence mechanism consists of “both a peripheral reflex reaction of the peroneal muscles that can counteract the inverting moment and also a centrally mediated strategy that results in a uniform reaction pattern of the muscles and joints of both the unilateral and the contralateral limb” (p. 57). As shown by Mitchell et al. (2008a), the dynamic defence mechanism, is primarily an eversion movement, with a dorsi flexion component, which is initiated when the ankle is forced in to excessive inversion and plantar flexion. This is measured using EMG, where typically the reaction time is defined as the time from the first moment of inversion, to the first EMG response (Eils & Rosenbaum, 2001).

Previous studies have measured the peroneus longus (Beckman & Buchanan, 1995; Karlsson & Andreasson, 1992; Mitchell et al., 2008a; Shima et al., 2005) and tibialis anterior (Mitchell et al., 2008a) as these muscles are directly involved in the dynamic defence mechanism. There is evidence to suggest that the gluteus medius may also be affected by sudden inversion; Beckman and Buchanan (1995) found that subjects with hypermobile ankles had decreased latency of hip muscle activation during ankle inversion, implying that foot and ankle abnormalities can also influence joints proximal to the ankle. The anatomy of the peroneus longus, tibialis anterior and gluteus medius is explained in Appendix Eight.

4.2.2 Correction Strategies

Ankle and hip strategies have been used to describe the use of corrective movements when maintaining postural stability after perturbation (Gribble & Hertel, 2004; Horak, 1987). Ankle strategy involves shifting the centre of body mass by rotating the body about the ankle joint, whereas the hip strategy shifts the centre of body mass by flexing or extending the hips (Horak, 1987). Horak, (1987) identifies that ankle strategy is used for smaller perturbations, whereas hip strategy is typically used for larger perturbations where ankle strategy may be insufficient in maintaining balance. Hertel (2000) added that hip strategy is less effective than ankle strategy; ankle strategy is generally used by healthy adolescents and young adults, whereas hip strategy is used by the elderly and those with balance disorders who may be unable to maintain balance by ankle strategy.

4.2.3 Design of Tilt Platforms

In the literature, a variety of tilt platforms have been used, varying in terms of the direction of the tilt, as well as the degree to which the tilt occurs. Most existing platforms are uniplanar, causing movement only in the frontal plane (Beckman & Buchanan, 1995; Benesch et al., 2000; Cordova et al., 2009; Eils & Rosenbaum, 2001; Karlsson & Andreasson, 1992; Shima et al., 2005). However Ebig, Lephart, Burdett, Miller and Pincivero (1997), Lohrer et al. (1999), and Mitchell et al. (2008a) used tilt platforms with a combination of inversion and plantar flexion. The mechanism of an ankle sprain is a combination of inversion and plantar flexion, so tilt platforms with only uniplanar movement have limited ecological validity. The starting position of the tested foot is thought to have an effect on muscular reaction time, some researchers have used wedges to induce plantar flexion when standing on the tilt platform; Benesch et al. (2000) found that a slope of 15° significantly reduced reaction time of the peroneal muscle group. Similarly, Lynch, Eklund, Gottlieb, Renström and Beynnon (1996) used a wedge causing 20° of plantar flexion, which was also found to reduce reaction time of the peroneal muscle group, therefore indicating a loss of neuromuscular reflexes. The degree of forced inversion is inconsistent in the literature, which makes comparison difficult. Degrees of inversion range from 18° (Lynch et al., 1996) to 37° (Ricard, Sherwood, Schulthies & Knight, 2000), with most tilt platforms set at around 30° of inversion (Beckman

& Buchanan, 1995; Benesch et al., 2000; Eils & Rosenbaum, 2001; Lohrer et al.,1999; Karlsson & Andreasson, 1992; Mitchell et al., 2008a). Some studies have recorded the velocity at which the tilt platform inverts, Beckman and Buchanan (1995) reported that the ankle was inverted at 700° / second about an axis located near the subtalar joint and midtarsal joint axes. Lynch et al. (1996) measured latency changes at two different velocities, and found that, as expected, increased speed produced a shorter latency response of the peroneals.

Some tilt platforms have only one operational tilt (Cordova et al., 2009; Karlsson & Andreasson, 1992, Shima et al., 2005), meaning the subjects are aware of which foot will be tilted, whereas others have two independently tilting components, so either foot can be tested at any time, therefore reducing any subject anticipatory effect (Beckman & Buchanan, 1995; Ebig et al., 1997; Mitchell et al., 2008a; Vaes et al., 2002). In some tilt platform studies, subjects start in a bilateral stance, with equal weight distribution on both legs (Cordova et al., 2009; Ebig et al., 1997; Karlsson & Andreasson, 1992; Mitchell et al., 2008a; Shima et al., 2005), this could be criticised for lacking in ecological validity, as most ankle sprains occur during gait between heel strike and midstance (Hertel, Guskiewicz, Kahler & Perrin, 1996) rather than when both feet are fixed to the ground. Some researchers have attempted to overcome this by having subjects start with uneven weight distribution (Eils & Rosenbaum, 2001; Lohrer et al., 1999; Ricard, Sherwood et al., 2000; Vaes et al., 2002), however, by doing this, subjects anticipate the tilt occurring on that particular limb, potentially influencing results. Several researchers have emphasised the difficulty in reproducing an ankle sprain mechanism in a laboratory (Lynch et al., 1996; Mitchell et al., 2008a). Lynch et al. (1996) suggested that when standing on a tilt platform, bilateral stance with equal body weight distribution is the safest and most reproducible posture; therefore creating the most effective balance between internal and external validity.

More recently researchers have attempted to increase the ecological validity of tilt platforms by using a dynamic (walking) tilt platform as opposed to a static platform (Hopkins et al., 2006; Hopkins, Brown, Christensen & Palmieri-Smith, 2009). Whilst a dynamic platform may improve ecological validity by removing the static aspect, the major criticism is that at present, dynamic tilt platforms have no plantar flexion component. Until this is rectified, the use of a dynamic tilt platform can be questioned, as it is not directly applicable to the mechanism of an inversion and plantar flexion ankle sprain. A further study which has attempted to increase ecological validity in this area is that of Gutierrez and Kaminski (2010), who designed a tilt platform on which subjects land from a 30 cm drop jump. This tilt platform design relates directly to the typical mechanism of injury whereby athletes obtain an inversion ankle sprain when landing on an unstable surface (Garrick, 1977). At present, this is the only study to have used a tilt platform of this kind, and only six subjects of different populations were used in this study (two with unstable ankles, two with history of ankle sprain but no instability, and two uninjured controls). Until further research has been performed, no conclusions can be drawn with regards to how reliable this type of tilt platform is.

4.2.4 Reliability of Reaction Time Measures

Several studies have assessed the reliability of muscular reaction time to a tilt platform perturbation; Lynch et al. (1996) found that muscular reaction times of the peroneus longus and tibialis anterior muscles were reliable and repeatable across the two muscles over different days (ICC = .82 and .74 respectively). Benesch et al. (2000) also concluded that peroneal reaction time is a consistent measure, which is reliable in repeated measurements and independent of the time of measurement, however they noted that the results are specific to the tilt platform and methods used. Eechaute et al. (2007) reported high reliability on measures taken one week apart (ICC > .70). Although some studies have established good reliability during reaction time measures, there are a number of variables that must be considered before making comparisons between different studies. As previously mentioned it is important to consider differences in the design of the tilt platform, in addition to this, it is vital to consider the use of EMG.

4.2.5 Use of Electromyography

Electromyography is widely used in scientific research for the analysis of muscle function (Soderberg & Knutson, 2000). It is used to record changes in the action potential of a muscle when it is caused to contract by a motor impulse nerve (Bartlett, 2007). During voluntary muscle contractions, the motor impulse nerve reaches a depolarisation threshold which generates an electromagnetic field, as this occurs, the motor unit action potential, which is a collection of the individual action potentials for all the fibres within a specific motor unit, is detected by an electrode (Konrad, 2005; Rash, 2002). The motor unit action potentials, of all active motor units within the recording area, are electrically superposed, this creates the raw EMG signal showing positive and negative amplitudes which represent the recruitment and firing frequency of the motor unit action potentials (Konrad, 2005).

Until recently there was no consensus on the use of EMG equipment, or recording of the procedure and results in scientific reports. SENIAM (surface EMG for a non-invasive assessment of muscles) is a European based organisation attempting to standardise the use of EMG (Hermens, Freiks, Disselhorst-Klug & Rau, 2000). Its recommendations for use are comprised from peer-reviewed articles and expert opinions and are well regarded among current literature (Bartlett, 2007; Campanini et al., 2007; Konrad, 2005). Although it provides specific suggestions for electrode placement it has received some criticism for being unspecific by using generic terms such 'muscle belly' (Rainoldi, Melchiorri & Caruso, 2004; Sacco, Gomes, Otuzi, Pripas & Onodera, 2009).

Electromyography is the only way to objectively assess when a muscle is active (Bartlett, 2007), and can be analysed in both amplitude and frequency variables (Kollmitzer, Ebenbichler, & Kopf, 1999), however there are several factors which may influence results. Intrinsic factors which may affect EMG results include the type of muscle fibre, the diameter of the muscle fibres, and the thickness of the adipose tissue between the muscle fibres and the electrode (Bartlett, 2007; Rash, 2002). Intrinsic factors are out of the researchers control, however there are several extrinsic factors to consider when using EMG in research including equipment to be used, testing procedure, and the processing of data.

4.2.6 Electromyography Equipment

When selecting the equipment to be used for EMG research, the primary consideration is the type of electrode to be used. Most current studies in to human movement use surface electrodes for EMG analysis of superficial muscles; they take more of a global view of muscle activity due to having a larger surface area, whereas intra-muscular electrodes (needle or fine-wire) identify specific motor units (Soderberg & Knutson, 2000). There are several advantages to using surface electrodes; they are safer to use as they are less intrusive, they are easier to use and are often more accessible than intra-muscular electrodes (Bartlett, 2007). However, there are also several disadvantages; the larger surface area of surface electrodes allow for crosstalk from surrounding muscles, and only superficial muscles can be analysed (Bogey, Cerny & Mohammed, 2003).

Several studies have assessed the reliability of both surface and intra-muscular electrodes in EMG research. The use of surface electrodes has been shown to reduce error over different trials when compared with intra-muscular electrodes (Soderberg & Knutson, 2000). Pease and Elinski (2003) also found surface electrodes were significantly more reliable than intra-muscular electrodes when measuring muscular fatigue on the vastus lateralis and tibialis anterior in healthy subjects. In contrast, Bogey et al. (2003) found similar levels of repeatability for both surface electrodes and intra-muscular electrodes when using a test retest design, with both types of electrodes showing high levels of consistency. To standardise the use of EMG in research, SENIAM guidelines suggest the use of bipolar silver/silver chloride surface electrodes, with 20 mm between the centres of two pre-mounted electrodes.

Other equipment required for EMG analysis includes a recorder for analogue to digital conversion (Konrad, 2005); SENIAM recommends a 12-bit or 16-bit board. An amplifier is also used to optimise the resolution of the recording equipment (Gerdle, Karlsson, Day & Djupsjöbacka, 1999). Some surface electrodes have inbuilt pre-amplifiers, which greatly increase the skin impedance level (Gerdle et al., 1999); these allow the signal to be detected earlier, become amplified then transmit the signal at a lower level less sensitive to movement artefacts (Konrad, 2005). Band-pass filtering is important to reduce movement artefacts and other noise, typically frequency ranges are between 10 – 20 Hz (high pass filtering) to between 500 – 1000 Hz (low pass filtering) (Gerdle et al., 1999). This range reduces movement artefacts which are usually at a low frequency, and also reduces high frequency noise avoiding signal aliasing, where signal information becomes indistinguishable (Bartlett, 2007; Gerdle et al., 1999; Konrad, 2005). Konrad (2005) suggested that sampling frequency should be at least twice the value of the highest frequency in the signal in order to obtain adequate detail in EMG recordings. The common mode rejection ratio should also be considered when using EMG equipment; this is the ability of the amplifier to filter out common mode signals; the higher the ratio, the better the filtration of common signal (Rash, 2002).

4.2.7 Electromyography Testing Procedures

When planning the procedure for the use of EMG in research, there are several factors to consider dependant on the muscles to be tested. The primary consideration is the placement of the electrodes; Bartlett (2007) indicated that the location of electrodes in relation to motor end plates, and the orientation of electrodes with respect to muscle fibres can influence EMG recordings. With this in mind, there has been extensive research into the ‘correct’ placement of surface electrodes. SENIAM have collaborated much of the research into a guide identifying electrode placements for specific muscles, however since this guide was released in 2000 there has been some disagreement in more recent literature. SENIAM guidelines have been criticised for being too ambiguous by Rainoldi et al. (2004) and Sacco et al. (2009) found that on two of four muscles tested, the SENIAM guidelines were either incorrect or imprecise.

Gerdle et al. (1999) and Bartlett (2007) suggested that electrodes should be placed over the muscle belly, parallel to the direction of the muscle fibres or, for consistency, pointing towards the origin and insertion of the muscle. Some researchers suggest the avoidance of motor end plates and musculo-tendinous junctions due to increased signal instability (Gerdle et al., 1999; Konrad, 2005; Rainoldi et al., 2004; Rash, 2002). Hermens et al. (2000) also noted that placing surface electrodes over motor end plates results in uncharacteristic EMG signals; however dependant on the size of the muscle being analysed, and the shape of the electrodes, this is often unavoidable (Konrad, 2005).

For repeatability, Konrad (2005) suggested using palpable anatomical landmarks to ensure accurate positioning of surface electrodes. Rainoldi et al. (2004) stated that just 1 cm of electrode displacement can result in variations of 200% in amplitude measurements. To ensure accurate placement of electrodes, Rainoldi et al. (2004), identified the optimum locations between innervation zones (where nerve terminations and muscle fibres are connected) and tendons for specific muscles. For most muscles in this study, Rainoldi et al. (2004) found that it was possible to use bony landmarks as a basis for electrode placements without first finding the innervation zone, however for some muscles, for example peroneus longus and gluteus medius, the innervation zone varies between subjects and therefore standard definitions for electrode placement are not accurate. Mesin, Merletti and Rainoldi (2009) also found that innervation zones may vary between different muscles, and between the same muscles on different subjects. Rainoldi et al. (2004) therefore suggested that innervation zones for these muscles must be specifically identified for each subject prior to electrode placement to ensure accuracy.

A further consideration when positioning electrodes is the potential movement of the muscle beneath the skin during dynamic activity, particularly when testing smaller muscles, the electrodes may detect movement of surrounding muscles, or may slide over innervation zones (Campanini et al., 2007; Konrad, 2005; Mesin et al. 2009). Farina, Merletti, Nazzaro and Caruso (2001) found that innervation zones move during dynamic muscle contractions, which may wrongly appear as an increase in muscle activity if the innervation zone slides under the surface electrode; however this study was based on simulations. In agreement, Mesin et al. (2009) stated that electrodes should not cover the innervation zone of a muscle during any part of a dynamic movement as this can disrupt EMG measurements. Campanini et al. (2007) found high variability in results when measuring EMG amplitudes during gait, particularly on the peroneus longus and tibialis anterior muscles; this was thought to occur due to the long and thin structure of the muscles and their close proximity resulting in cross talk. A study by Sacco et al. (2009) found that during maximal isometric contractions, optimum electrode locations for some muscles varied slightly to those identified by SENIAM, specifically for the tibialis anterior muscle where the best site for electrode placement was at 47.5% of its length, rather than 33% as suggested by SENIAM. Sacco et al. (2009) recommend testing EMG signals in many positions along the muscle to take in to account the shift on innervation zones during dynamic movement; this however is not always practical. Correct skin preparation to reduce skin impedance is essential for accurate EMG results (Konrad, 2005). For optimum electrode to skin contact, it is recommended that the area is shaved and cleaned with an alcohol wipe, and left to dry before electrode attachment (Bartlett, 2007). A reference electrode should be placed on electronically inactive tissue close to the surface electrodes (Bartlett, 2007); this is usually a bony prominence such as the patella, or the radial styloid process. To avoid cable artefacts and disruption to data, cables should be held secure with elastic bands or tape (Bartlett, 2007), this is particularly important during dynamic activity, where loose cables may restrict performance (Konrad, 2005).

SENIAM recommend the use of clinical muscle tests, which can be performed on the subjects once the electrodes are in place, this enables the researcher to ensure the equipment is working correctly, and also ensures that the electrodes are positioned in the correct place. This technique is commonly used in scientific research (Rainoldi et al., 2004; Zazulak et al., 2005) although manual muscle tests are often criticised due to lack of specificity for some muscles. Soderberg and Knutson (2000) highlighted that the only way to isolate specific muscles when performing manual muscle tests would be to induce paralysis on surrounding muscle tissues, which is clearly impractical for most experimentation. Some studies, such as Zazulak et al. (2005) can be criticised for not identifying the manual muscles tests used, however for consistency, many studies use guidelines by SENIAM or Kendall, McCreary and Provance (1993).

4.2.8 Electromyography Data Processing

Before using EMG for research, it is important to consider the processing of the data. There are a number of ways to process data depending on the purpose of the study, specifically whether the researcher requires amplitude or frequency variables (Gerdle et al., 1999); most kinesiological studies require amplitude variables. The data must first be quantified, as the mean value without quantification is zero (Gerdle et al., 1999; Konrad, 2005). The only information that can be gained from non-quantified raw EMG is whether or not the muscle is active; this information can be gained by simple visual inspection of the raw EMG (Konrad, 2005), however can be criticised for being subjective. Figure 4.1 shows an example of a raw EMG trace.

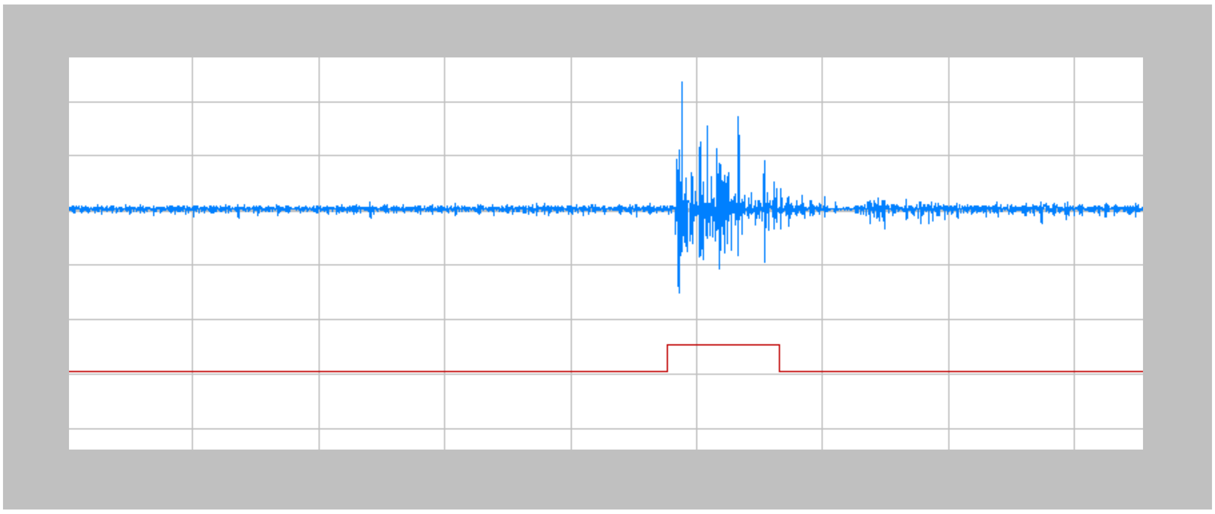


Figure 4.1 Raw EMG trace of peroneus longus. Red line indicates onset of tilt.

There are a number of ways to quantify EMG data; rectification is the simplest method; half wave rectification can be used where all values below the baseline are discarded, but more commonly full wave rectification is used where negative values are made positive (Bartlett, 2007; Burden, 2007; Soderberg & Knutson, 2000). Full wave rectification is often preferred as it retains all of the original signal energy (Gerdle et al., 1999). The average rectified value is an alternative way to quantify EMG data, this is the average value of full wave rectified EMG (Bartlett, 2007); this method suppresses fluctuations in the EMG signal, reducing the overall variability (Gerdle et al., 1999). This calculation can be made as a moving average, where the calculation is made several times at specific time intervals (Gerdle et al., 1999), this enables the researcher to smooth the data according to the type of activity being measured; shorter time frames are most suited for faster muscle contractions, resulting in more detailed data (Burden, 2007). Integrated EMG is similar to the average rectified value; however the value is not divided by time (Burden, 2007; Gerdle et al., 1999). It is not recommended by SENIAM however Morey-Klapsing, Arampatzis and Brüggemann (2004) used it as the main parameter in a study and claimed that there is little controversy regarding the use of integrated EMG in current literature.

Smoothed rectified EMG, or linear envelope, is low pass filtering of the full wave rectified EMG signal (Bartlett, 2007; Rash, 2003), this results in a smoothed curve representing the trends of the EMG signal (Burden, 2007). Winter and Scott (1991) claim this method has a strong physiological basis compared to other quantifying procedures as the output follows the trends of the muscle tension curve, SENIAM therefore recommend this for fast dynamic contractions. A cut off frequency must be predetermined for smoothed rectified EMG, similar to choosing the time windows for a moving average; a higher value will result in a very detailed curve, whereas a lower value will show a very smooth curve, lacking details of muscle activation (Burden, 2007). Generally, dynamic activities of short duration require a high cut off frequency (Robertson et al., 2004).

The root mean square (RMS) is one of the most common procedures to quantify data in current research and is defined as the square root of the mean square value (Merletti, 1996). This procedure does not require rectification as it squares each value in the signal, creates an average of a specific time interval, and then calculates the square root (Rash, 2003; Soderberg & Knutson 2000). The root mean square is often preferred to other quantification methods as it is thought to be based on more of a mathematical basis than other procedures, with more physical meaning (De Luca, 1997; Gerdle et al., 1999; Soderberg & Knutson, 2000). A moving average is more applicable than just one general calculation, as with the average rectified value, this is a series of RMS calculations made at specific durations (Burden, 2007; Gerdle et al., 1999). Depending on the movement being analysed, the specified time period is usually between 20 ms for more dynamic movements, up to 500 ms for more static activities (Konrad, 2005). The time periods can be adjusted according to the movement being analysed, with longer durations causing more smoothing of data, resulting in less variability of data. The Root Mean Square is recommended by SENIAM for non-dynamic amplitude contractions due to its ability to detect changes in signal which may be unidentifiable with other measures due to greater signal variability. De Luca (1997) also supports the use of RMS for voluntary muscle contractions as it represents signal power; it therefore provides more information than the alternative methods (Gerdle et al., 1999).

4.2.9 Onset Detection Methods

As previously mentioned, the on-off characteristic of EMG can be determined by observation of the raw EMG signal (Ebig et al., 1997; Konrad 2005); for example Shima et al. (2005) identified peroneal latency by “the time from the onset of the tilting of the trapdoor to the first big EMG response of peroneal longus muscle” (p. 477). Similarly Eils and Rosenbaum (2001) determined EMG onsets from “when the EMG response showed a steep increase followed by enduring activity” (p. 1993) however this can be criticised for being highly subjective and unscientific, particularly if the baseline is unclear. Threshold analysis provides a more objective measure to distinguish between baseline noise and muscle activity (Burden 2007; Soderberg & Knutson, 2000). This technique involves taking the mean baseline measure over a specific time period, then observing the point at which the EMG amplitude reaches a predefined number of standard deviations above the baseline value for a set duration. For example, in a study by Cordova et al. (2009), the initiation of the peroneus longus response was identified when the EMG signal reached five times the standard deviation of the baseline mean which was measured for 150 ms prior to the tilt platform perturbation. In current literature, the parameters used for threshold analysis vary greatly in terms of the length of the baseline measure, the duration for which the mean must exceed the threshold, and also the number of standard deviations above the baseline the amplitude must be. Hodges and Bui (1996) found that when a long sample is used, the chance of ignoring the onset is increased, however when a short sample is used, the researcher is more likely to mistake background activity for onset of muscle activity. Studies assessing muscle latency after tilt platform perturbations have ranged from using two standard deviations (Berg, Hart, Palmieri-Smith, Cross & Ingersoll, 2007; Konradsen, Ravn & Sorensen, 1993), three standard deviations (Ball & Scurr, 2009; Di Fabio, 1987) to as high as ten standard deviations above the baseline measure (Lynch et al., 1996). Hodges and Bui (1996) suggested that the standard deviation must be high enough to avoid a type I error, where the muscle is identified as active when it is not, yet low enough to avoid a type II error, where the researcher fails to identify the EMG onset when it occurs.

4.2.10 Normalisation of Electromyography Analysis

Normalisation is often used when analysing EMG to provide a standardised reference value, which enables the direct comparison of data across the same muscle on different occasions, as well as different muscles on different individuals (Burden, 2007; Soderberg & Knutson, 2000). This helps to reduce between-subject variability in EMG amplitudes, and converts EMG amplitude to an estimate of muscle activation (Staudenmann, Roeleveld, Stegeman & van Dieën, 2009). There are a number of methods used in current literature to normalise EMG data. One of the most common procedures is to express the data as a percentage of the maximum voluntary isometric contraction (MVIC) of the specific muscle being analysed (Rash, 2003; Soderberg & Knutson, 2000). This usually involves either the examiner conducting a manual muscle test on the subject (Bogey et al., 2003; Zazulak et al., 2005), or the use of an isokinetic dynamometer (Carcia et al., 2007). This technique can be used to show how active a muscle is during a specific movement in comparison to the muscles maximum activation capacity (Burden, 2007). There are however a number of limitations to this method, the primary concern being the validity of the muscle test used due to the difficulty in isolating specific muscles (Soderberg & Knutson, 2000), and also, knowing whether the subject is actually exerting maximum force during the contraction. Merletti (1996) highlighted that without training, MVICs could be as much as 20-30% less than the true maximum. The use of sub-maximal contractions has been suggested to be a more reliable method than maximal contractions (Yang & Winter, 1984), perhaps due to signal instability above 80% of MVIC (De Luca, 1997).

Some researchers have questioned the validity of using an isometric contraction to compare against a dynamic movement, particularly a forced dynamic movement such as that during sudden inversion due to a tilt platform perturbation (Mitchell et al., 2008a). To overcome this, it has been suggested to normalise values against within-test measures such as the use of the maximum EMG values within a task (Cordova et al., 2009), or the use of the mean level of the EMG signal across the task rather than MVICs (Rash, 2003). Other researchers have suggested more dynamic methods of normalisation, for example Norcross, Blackburn and Goerger (2009) compared the reliability of single leg stance, and MVIC normalisation methods on hip muscles, and found that both methods produced good reliability in all muscles tested, with MVICs resulting in lower coefficient of variance than the single leg stance. Norcross et al. (2009) suggested that the single leg stance provided a better representation of coordinated muscle function, and is therefore more applicable than MVICs for closed kinetic chain tasks. Although the single leg stance was shown to have good reliability, Norcross et al. (2009) do not recommend it for all muscles as it showed too much variation to be considered a stable reference, particularly for the rectus femoris, vastus lateralis and biceps femoris muscles. In a similar study, Ball and Scurr (2010) compared an isometric contraction, and isokinetic contraction, a squat jump, and a 20 m sprint as methods of normalisation. This study showed that a squat jump was more reliable than the other methods producing reliable amplitudes over different days and weeks. Knutson, Soderberg, Ballantyne and Clarke (1994) found that MVICs had the best reproducibility when compared with peak and mean dynamic normalisation methods. Staudenmann et al. (2009) concluded that while EMG normalisation is unavoidable for EMG based force estimation, its validity is often limited due to poor methodology, particularly in clinical applications.

4.2.11 Summary of Muscular Reaction Time Literature

Considering the association between the incidence of ankle sprains and pronated and supinated foot structures, there is a clear need for research in this area to determine whether foot structure affects muscular reaction time. In addition, after reviewing current literature on reaction time and electromyography techniques, it is apparent that there is a need for more definitive parameters to be identified and used throughout this research area to enable accurate comparison of results. Before Study Two is carried out, reliability testing needs to be conducted in order to test reliability of electrode placement, of the tilt platform, of the reaction time measurements, and of onset detection techniques.

4.3 Pilot Study Three: Test Retest Reliability of Electrode Placement

4.3.1 Abstract

Context: Surface electrodes are widely used in current research; however there is some concern over the reliability of electrode placement.

Objective: To determine intra-tester reliability of electrode placement on the peroneus longus, tibialis anterior and gluteus medius muscles.

Subjects: Five (4 males, 1 female), age = 21 ± 1.0 years, height = 179 ± 7.9 cm, mass = 77.1 ± 16.3 kg.

Methods: Electrodes were attached to the peroneus longus, tibialis anterior and gluteus medius three times on two days. On each day, maximal voluntary isometric contractions were recorded during manual muscle tests. Intraclass correlation coefficient and standard error of measurement were then calculated for each muscle.

Results: Reliability was found to be high over two days (ICC > .70) for each muscle.

Conclusions: High ICC results indicated accurate electrode placement over two days, therefore this method of surface electrode placement was determined suitable for use in Study Two.

4.3.2 Introduction

As discussed in Section 4.2 EMG is widely used in research as a tool to measure muscle function. A key factor in ensuring accurate EMG measurement is the placement of surface electrodes; without accurate placement, the electrodes could detect interference from motor end plates or musculo-tendinous junctions (Gerdle et al.. 1999), or surrounding muscles, therefore compromising results. Previous studies have highlighted concern regarding the reliability of surface electrode placement; therefore testing was carried out in order to determine whether EMG was a suitable tool for use in Study Two.

4.3.3 Methods

Subjects:

Subject characteristics are identified in Table 4.1. The inclusion and exclusion criteria were the same as those to be used in Study Two, as identified in Section 3.2.3.

Table 4.1 Subject Characteristics

Sex	Age (years)	Height (cm)	Mass (kg)
4 Males 1 Females	21 (1.0)	179 (7.9)	77.1 (16.3)

Values are mean (SD).

<u>Procedures:</u>

Reliability of electrode placement was tested over two days for each of the three muscles to be used in Study Two; the peroneus longus, the tibialis anterior and the gluteus medius. Testing involved positioning electrodes over the muscles on only the dominant limb (as defined in Section 3.2.3) over two consecutive days. On each day electrodes were applied three times, and with each application three maximal voluntary isometric contractions (MVICs) were recorded for five seconds on each muscle using manual muscle tests. Muscle tests were performed at least two minutes apart to allow for muscle recovery. Maximal voluntary isometric contractions were performed according to standard manual muscle tests as indicated by SENIAM:

- Tibialis Anterior; support the leg just above the ankle joint with the ankle joint in dorsiflexion and the foot in inversion. Apply pressure against the medial side, dorsal surface of the foot in the direction of plantar flexion of the ankle joint and eversion of the foot.

- Peroneus Longus; support the leg above the ankle joint. Everse the foot with plantar flexion of the ankle joint while applying pressure against the lateral border and sole of the foot, in the direction of inversion of the foot and dorsiflexion of the ankle joint.

- Gluteus Medius; lying on the side with the legs spread against manual resistance (holding the ankles).

All testing was carried out using the same eight-channel DataLink EMG system (Biometrics, UK) to be used in the Study Two. Prior to the application of the electrodes, the skin was first prepared by shaving the area if necessary, then cleaned with an alcohol wipe to reduce skin impedance and left to dry. A passive reference electrode was placed around the radial styloid, and preamplified surface bipolar electrodes (Biometrics SX230; gain x 1000, bandwidth 20 Hz – 460 Hz, noise < 5 µV, input impedance > 100 MΩ, common mode rejection ratio > 96 dB) were then positioned according to SENIAM guidelines in the direction of the muscle fibres. Details of SENIAM guidelines are shown in Appendix Nine.

Data Reduction and Statistical Analysis:

The EMG signals were sampled at 1000 Hz. Raw data was processed with an RMS filter using a 20 ms moving window, this time frame was selected as it has previously been suggested that a longer time frame may be more suitable for the analysis of sustained muscle contractions such as MVICs (Konrad, 2005). Following this, the first and last second of the trace was discarded to account for the time it takes to reach a sustained MVIC from a static position, and to account for subjects reducing the contraction in anticipation of the end of the trail. Following a procedure by Bolgla and Uhl (2007) the 500 ms window with the highest average throughout the trace was then identified from the remaining three second trace using an Excel template (Microsoft Office, 2010). The peak value from the entire trace of each trial was also identified. Two measures were calculated; firstly the highest 500 ms from the three trials were averaged to create the average peak 500 ms amplitude; and secondly, the peak values from each of the three trials were averaged to create the average peak amplitude.

As data were normally distributed (Shapiro-Wilk $P > .05$), it was tested for significant differences using a paired samples t-test with the a priori alpha level set at .05. No significant differences were found. Relative and absolute reliability were then calculated using ICC and SEM respectively. All statistical analyses were calculated using Excel (Microsoft Office, 2010).

4.3.4 Results

Reliability of electrode placement results are shown in Table 4.2. All results indicate moderate to high reliability (ICC > .50) (Munro, 2000).

4.3.5 Discussion

The results of this pilot study show that the ICC of electrode placements of the peroneus longus, the tibialis anterior and the gluteus medius are moderate to very high (Munro, 2000) for both the average peak variable, and the average peak 500 ms variable. The reliability coefficients for the gluteus medius peak being the highest (ICC = .98, SEM = 0.03 mV), and the lowest being the tibialis anterior 500 ms (ICC = .64, SEM = 0.04 mV). The SEM was generally low across all scores indicating good absolute reliability. These results indicate that the electrode placement procedures as used in this pilot study are adequately reliable for use in Study Two.

Table 4.2 Test Retest Reliability of Electrode Placement

		Test		Retest		*P*	ICC	SEM
Peroneus Longus (mV)	500 ms	0.17	(0.05)	0.16	(0.06)	.25	.94	0.01
	peak	0.34	(0.09)	0.32	(0.11)	.51	.89	0.03
Tibialis Anterior (mV)	500 ms	0.24	(0.07)	0.25	(0.05)	.57	.64	0.04
	peak	0.44	(0.17)	0.47	(0.11)	.71	.71	0.07
Gluteus Medius (mV)	500 ms	0.20	(0.10)	0.27	(0.11)	.10	.96	0.02
	peak	0.34	(0.20)	0.47	(0.25)	.10	.98	0.03

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement; 500 ms: highest 500 ms from three trials averaged; peak: average of peak values in three trials. ICC > .70 indicating high correlation highlighted.

4.4 Pilot Study Four: Test Retest Reliability of the Dynamic Tilt Platform

4.4.1 Abstract

Context: Before conducting Study Two, it was vital to determine whether both the left and right parts of the tilt platform tilted at the same speed over different trials.

Objective: To determine reliability of the dynamic tilt platform.

Subjects: Six (4 male, 2 female), age = 22 ± 1.0 years, height = 176.7 ± 13.7 cm, mass = 71.7 ± 22.1 kg.

Methods: Three tilts were performed on both the left and right leg of each subject on two separate days. Time of tilt was recorded using triggers which were activated upon movement of the tilt platform. Intraclass correlation coefficient and standard error of measurement were then calculated for both the left and right parts.

Results: Both left and right tilting parts were shown to be highly reliable over two days (ICC > .99).

Conclusions: Due to high reliability found on both the left and right tilting parts, it was decided that the tilt platform was suitable for use in Study Two.

4.4.2 Introduction

The tilt platform available for use in Study Two was the same as that used by Mitchell et al. (2008a). It comprised of two independently moveable tilting parts, each moving from a neutral position to 30° inversion and 20° of plantar flexion when released. A diagram of the tilt platform is shown in Figure 4.2. In order to improve internal validity of Study Two, reliability testing was carried out on the tilt platform to ensure that both the left and right parts tilted at the same speed over different trials.

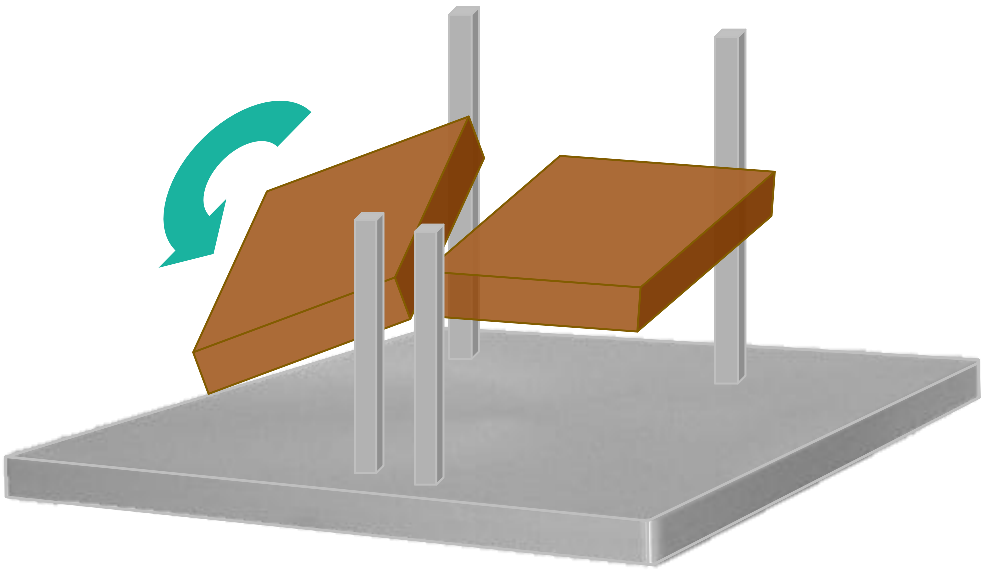


Figure 4.2. Anterior-lateral view of the dynamic tilt platform, showing a right tilt

4.4.3 Methods

<u>*Subjects:*</u>

Subject characteristics are shown in Table 4.3. The inclusion and exclusion criteria were the same as those to be used in Study Two, as identified in Section 3.2.3.

<u>*Procedures:*</u>

Testing was carried out over two days, on each day subjects were asked to stand on the tilt platform, and were told they would experience three tilts on each limb, in a randomised order. No electrodes were attached to the subjects, as no EMG data was required. The purpose of using subjects rather than simply tilting the platform without any weight was to encourage the tilt platform to function as it would with subjects standing on it.

Two triggers were used on the tilt platform to record the time of each tilt using a digital input on DataLink (Biometrics, UK). The onset trigger was activated as soon as the tilt was initiated, and the second trigger was activated once the tilt platform reached the complete tilt. The EMG signals were sampled at 1000 Hz.

Table 4.3 Subject Characteristics

Sex	Age (years)	Height (cm)	Mass (kg)
4 Males 2 Females	22 (1.0)	176.7 (13.7)	71.7 (22.1)

Values are mean (SD).

Data Reduction and Statistical Analysis:

A custom made Excel (Microsoft Office, 2010) template was used to identify the time difference between the onset of the tilt to the complete tilt. As data were normally distributed (Shapiro-Wilk $P > .05$), data were tested for significant differences using a paired samples t-test with the significance level set at .05. No significant differences were found. Correlations were then calculated using an ICC, and SEM was also calculated. All statistical analyses were calculated using Excel (Microsoft Office, 2010).

4.4.4 Results

The results of test retest reliability of both the left and right tilting parts of the tilt platform are shown in Table 4.4. Both left and right tilting parts resulted in high reliability (ICC > .70).

Table 4.4 Test Retest Reliability of the Dynamic Tilt Platform

	Test	Retest	*P*	ICC	SEM
Left Tilting Part (ms)	52.8 (4.6)	53.2 (4.7)	.50	.99	.2
Right Tilting Part (ms)	54.3 (3.8)	53.9 (5.9)	.74	.99	.8

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement. ICC > .70 indicating high correlation highlighted.

4.4.5 Discussion

The results of this pilot test show that the speed at which both the left and right tilting parts of the tilt platform move is highly reliable (Munro, 2000) over different days (ICC > .90), and is therefore suitable for use in Study Two. The SEM of the right tilt (0.8 ms) is slightly higher than that of the left (0.2 ms), however the difference of 0.6 ms was not thought to have an impact on the overall reaction times of Study Two. The tilt speeds found in this study are faster than that of Konradsen et al. (1997), who found that it took 80 ms for the loaded tilt platform to reach full tilt. The faster times in this study are likely to be due to differences in tilt platform design and materials used for construction.

4.5 Pilot Study Five: Test Retest Reliability of Onset Detection Techniques

4.5.1 Abstract

Context: The current literature, there is no standard way to quantify muscular reaction time, with existing studies using a range of different parameters. This makes comparisons between studies difficult as it is unclear which combination of parameters is most reliable.

Objective: To determine reliability of onset detection techniques to calculate muscular reaction time of the peroneus longus, tibialis anterior, and gluteus medius.

Subjects: Nine (7 males 2 females), age = 21 ± 1.6 years, height = 177 ± 9.6 cm, mass = 69.9 ± 11.4 kg.

Methods: Reaction times of the peroneus longus, tibialis anterior and gluteus medius to a tilt platform perturbation were collected three times on two separate days. Reaction times were calculated using nine different combinations of onset detection parameters. Intraclass correlation coefficient and standard error of measurement were then calculated for each combination and each muscle.

Results: Results indicated that a 10 ms RMS moving window, 150 ms baseline, 25 ms burst, and 3 standard deviations above the baseline produced the most reliable results over the three muscles (peroneus longus ICC = .85 SEM = 3.20ms, tibialis anterior ICC = .97 SEM = 2.88ms, gluteus medius ICC = .79 SEM = 4.88ms).

Conclusions: Study Two will use the combination of parameters found to be most reliable for onset detection of muscle reaction time; 10 ms RMS moving window, 150 ms baseline, 25 ms burst, and 3 standard deviations above the baseline.

4.5.2 Introduction

There is much ambiguity in the current literature regarding the best method of analysis for determining EMG reaction time. Parameters include the RMS window length when exporting data, the length of the baseline measure, the number of standard deviations above the baseline measure, and the duration for which the burst is maintained. Reaction times are a well-established measure of neuromuscular control (Beckman & Buchanan, 1995; Mitchell et al., 2008a; Ricard, Sherwood et al., 2000) however with such varied parameters it is difficult to make comparisons among different studies. Therefore testing was carried out to determine which combination of variables produced the best reliability over two days. In order to choose the variables to be included in this study, a thorough review of literature was conducted to find the most widely used variables. A summary of commonly used variables are shown in Table 4.5.

Table 4.5 Onset Detection Parameters in Previous Literature

Authors	RMS Moving Window (ms)	Baseline (ms)	Duration of Burst (ms)	Standard Deviation
Akhbari et al. (2007)	50	100	-	3
Allison et al. (1999)	-	500	-	5
Berg et al. (2007)	15	-	-	2
Cordova et al.(2009)	-	150	-	5
Di Fabio (1987)	-	50	25	3
Fernandes et al. (2000)	-	500	-	-
Hopkins et al. (2006)	10	-	-	4
Hopkins et al. (2007)	-	150	-	2
Hopkins et al. (2009)	-	150	-	3
Konradsen et al. (1993)	-	100	-	2
Konradsen et al. (1997)	-	100	-	-
Linford et al. (2006)	10	-	-	2
Osborne et al. (2001)	-	1000	10	3
Papadopoulus et al. (2008)	-	-	-	2

- Indicates unidentified parameter

4.5.3 Methods

Subjects:
Subject characteristics are shown in Table 4.6. The inclusion and exclusion criteria were the same as those to be used in Study Two, as identified in Section 3.2.3.

Table 4.6 Subject Characteristics

Sex	Age (years)	Height (cm)	Mass (kg)
7 Males 2 Females	21 (1.6)	177.0 (9.6)	69.9 (11.4)

Values are mean (SD).

Procedures:
Each subject was tested three times on two separate days on the dominant limb as defined in Section 3.2.3; this was the right leg for all subjects. Reaction time measurements were collected from the three muscles to be used in Study Two (peroneus longus, tibialis anterior and gluteus medius). Testing was carried out using the same eight-channel DataLink EMG system (Biometrics, UK) and the same electrode placement procedure as shown to be reliable in Section 4.3. The same tilt platform as shown to be reliable in Section 4.4 was also used. Subjects were asked to stand on the tilt platform in a relaxed stance, looking directly ahead. Subjects were not told when the perturbations would occur, and they were initiated at variable intervals to reduce anticipatory effects.

Data were analysed nine times for each tilt recorded; each analysis involved a different combination of onset detection parameters as identified in previous literature. The parameters selected for analysis were RMS moving window (2 ms, 50 ms, 10 ms), baseline (150 ms, 500 ms, 1000 ms), duration of burst (10 ms, 25 ms, 50 ms), and standard deviations above the baseline (2, 3, 5). Figure 4.3 shows how using different RMS window lengths affect the EMG trace.

Data Reduction and Statistical Analysis:
As data were normally distributed (Shapiro-Wilk $P > .05$), each of the nine combinations for each muscle were tested for significant differences using a paired samples t-test with the significance level set at .05; no significant differences were found. Reliability was then calculated using ICC and SEM. All statistical analyses were calculated using Excel (Microsoft Office, 2010).

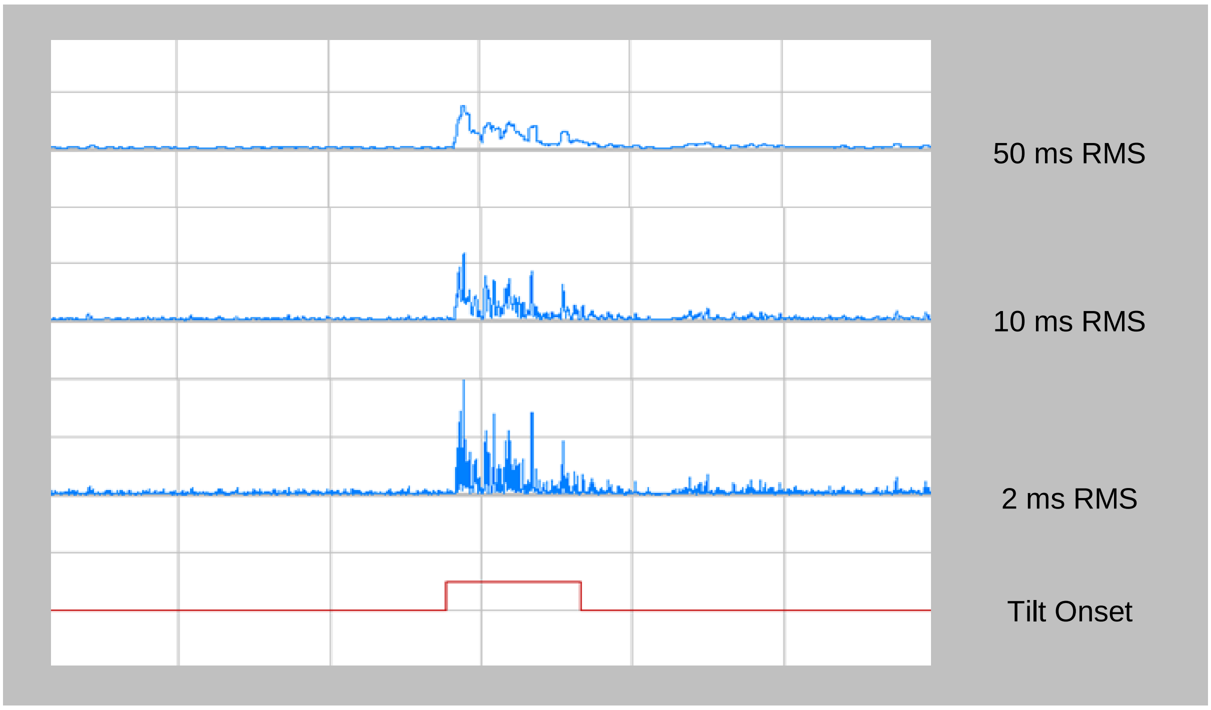


Figure 4.3. Changes to EMG Trace with Different RMS Window Lengths

4.5.4 Results

Results are shown in Table 4.7. Results indicated that a 10 ms RMS moving window, 150 ms baseline, 25 ms burst, and 3 standard deviations above the baseline produced the most reliable results over the three muscles (peroneus longus ICC = .85 SEM = 3.20ms, tibialis anterior ICC = .97 SEM = 2.88ms, gluteus medius ICC = .79 SEM = 4.88ms).

4.5.5 Discussion

It was concluded that the parameters to be used in Study Two were as follows; 10 ms RMS moving window, 150 ms baseline, 25 ms burst, and 3 standard deviations above the baseline. A diagram to explain how reaction time is calculated from these parameters is shown in Figure 4.4.

Table 4.7 Test Retest Reliability of Onset Detection Parameters

				Peroneus Longus					Tibialis Anterior					Gluteus Medius				
RMS	Base Line	Burst	SD	Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM
10	150	25	3	54.6(9.0)	53.4(7.9)	.92	.85	3.20	52.2(17.1)	49.4(17.9)	.08	.97	2.88	52.3(11.4)	53.3(10.8)	.68	.79	4.88
2	-	-	-	60.5(11.2)	59.3(11.3)	.59	.85	4.24	60.3(13.2)	57.5(14.7)	.50	.65	8.08	64.6(11.2)	69.1(18.7)	.30	.77	7.20
50	-	-	-	53.0(13.4)	46.8(12.7)	.06	.83	5.34	50.4(15.4)	44.9(19.5)	.16	.84	6.99	46.8(12.0)	40.6(23.1)	.32	.65	10.6
-	500	-	-	53.1(10.3)	54.5(9.9)	.57	.78	4.55	52.3(17.3)	51.6(13.6)	.82	.85	5.87	51.9(10.0)	53.4(13.5)	.66	.65	6.89
-	1000	-	-	53.2(10.3)	54.6(9.7)	.56	.78	4.61	53.6(14.1)	51.6(13.8)	.39	.89	4.50	54.7(9.1)	53.4(13.6)	.76	.39	8.79
-	-	10	-	48.2(13.9)	48.5(12.1)	.92	.66	7.39	47.9(17.7)	43.4(21.2)	.24	.87	6.92	40.3(13.0)	37.1(21.5)	.54	.75	8.62
-	-	50	-	53.3(10.6)	53.0(8.4)	.92	.70	5.06	54.4(12.6)	52.2(21.6)	.57	.92	4.84	55.6(12.6)	52.7(13.6)	.41	.69	7.11
-	-	-	2	51.7(11.8)	52.0(9.5)	.92	.69	5.77	50.9(17.8)	48.2(18.6)	.18	.96	3.67	46.1(13.7)	49.4(14.5)	.47	.56	9.15
-	-	-	5	55.6(9.3)	55.9(9.8)	.92	.79	4.25	55.2(15.1)	52.7(15.3)	.07	.97	2.41	58.2(8.8)	57.5(10.8)	.82	.57	6.26

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement. ICC > .70 indicating high correlation highlighted.

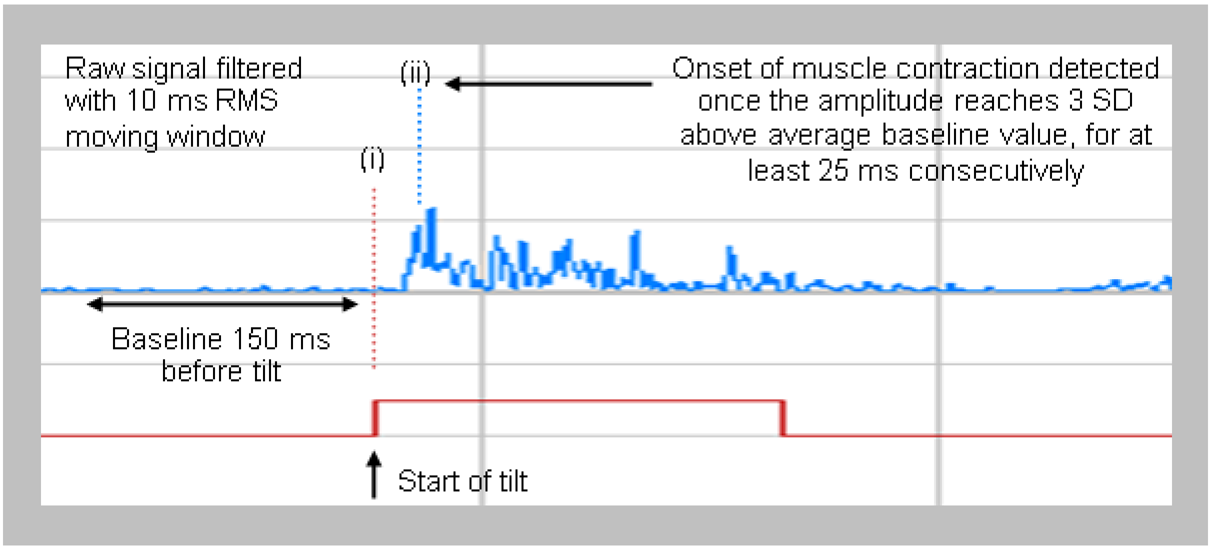


Figure 4.4 Onset Detection Using Tested Parameters. Reaction time is calculated by subtracting the time at (ii) (onset of muscle contraction) by the time at (i) (onset of tilt mechanism).

4.6 Pilot Study Six: Test Retest Reliability of Reaction Time Measurements

4.6.1 Abstract

Context: Before Study Two commenced, a final pilot test was required to incorporate all aspects of previous pilot testing (electrode placement, tilt platform and onset detection methods) to ensure that muscular reaction time measures were reliable over different days.

Objective: To determine the reliability of reaction time measurements.

Subjects: Six (3 males, 3 females), age = 21 ± 1.9 years, height = 171.2 ± 9.8 cm, mass = 69.9 ± 15.7 kg

Methods: Muscular reaction time (ms) of the peroneus longus, tibialis anterior, and gluteus medius were measured to a tilt platform perturbation over two days. Intraclass correlation coefficient and standard error of measurement were then calculated for each muscle.

Results: Reaction times for each muscle were highly reliable over two days; peroneus longus (ICC = .95); tibialis anterior (ICC = .99); gluteus medius (ICC = .91).

Conclusions: The reaction time method as used in this pilot test is suitable for use in Study Two.

4.6.2 Introduction

After determining reliability of the tilt platform, electrode placement and onset detection methods, reliability of reaction time measurements were tested to see whether results were consistent over different days. The purpose of this was to determine whether it was a suitable method for use in Study Two.

4.6.3 Methods

Subjects:

Subject characteristics are shown in Table 4.8. The inclusion and exclusion criteria were the same as those to be used in Study Two, as identified in Section 3.2.3.

Table 4.8 Subject Characteristics

Sex	Age (years)	Height (cm)	Mass (kg)
3 Males 3 Females	21 (1.9)	171.2 (9.8)	69.9 (15.7)

Values are mean (SD).

Methods:

Subjects were tested on two separate days; testing was carried out using the same EMG equipment and electrode placement procedure as used in Section 4.3. The same tilt platform as shown to be reliable in Section 4.4 was also used. Subjects were asked to stand on the tilt platform in a relaxed stance, looking directly ahead. Subjects were informed that up to eight tilts would occur and either ankle maybe tested at any one time. Subjects were not told when the perturbations would occur, and they were initiated at variable intervals to reduce anticipatory effects. Data was only recorded on the dominant leg as defined in Section 3.2.3.

Data Reduction and Statistical Analysis:

Raw EMG data was processed with an RMS filter using a 10 ms moving window. Data was then reduced using an Excel template (Microsoft Office, 2010) as shown in Appendix Ten. As shown to be reliable in Section 4.5, muscular reaction time was defined as the time between the onset of the tilt mechanism, to the onset of the EMG signal when it reached a level of three standard deviations above the baseline for 25 ms consecutively. The baseline value was the average value recorded over 150 ms immediately prior to the onset of the tilt mechanism.

As data were normally distributed (Shapiro-Wilk $P > .05$) it was tested for significant differences using a paired samples t-test with the significance level set at .05. No significant differences were found. Relative and absolute reliability were then calculated using ICC and SEM respectively, results are shown in Table 4.9. All statistical analyses were calculated using Excel (Microsoft Office, 2010).

4.6.4 Results

Results for the test retest reliability of the peroneus longus, tibialis anterior and gluteus medius are shown in Table 4.9. Reliability for each muscle is very high (ICC > .9).

Table 4.9 Test Retest Reliability of Muscular Reaction Time Measurements (ms)

Muscle	Test	Retest	*P*	ICC	SEM
Peroneus Longus	47.27 (6.08)	45.60 (5.75)	.08	.95	1.23
Tibialis Anterior	51.50 (7.06)	49.81 (8.97)	.13	.99	0.88
Gluteus Medius	47.83 (8.47)	47.53 (6.44)	.85	.91	2.18

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement. ICC > .70 indicating high correlation highlighted.

4.6.5 Discussion

The results of this study show that the ICC of reaction time measures are very high (Munro, 2000), which suggests that the testing procedure for muscular reaction time is acceptably reliable, and can therefore be used in Study Two.

4.7 Study Two: The Effects of Foot Structure on Muscular Reaction Time

This paper has been accepted for publication in the Journal of Athletic Training and is awaiting print. This study was also presented at the University of Hertfordshire Life Sciences Research Day (2012) and won an award for Best Poster Presentation (Appendix Eleven).

4.7.1 Abstract

Context: Foot structure has been shown to affect aspects of neuromuscular control including postural stability and proprioception; however no previous study has measured muscular reaction time to a simulated ankle sprain mechanism on participants with different foot structures. This is despite an association between pronated and supinated foot structures and the incidence of lateral ankle sprains.

Objective: To determine whether pronated and supinated foot structures contribute to neuromuscular deficits as measured by muscular reaction time to a simulated ankle sprain mechanism.

Subjects: Thirty volunteers were categorised into three groups according to navicular drop height measures. Ten participants had neutral feet (navicular drop height 5 – 9 mm), ten participants had pronated feet (navicular drop height ≥ 10 mm), and ten participants had supinated feet (navicular drop height ≤ 4 mm).

Methods: Muscular reaction time (ms) of the peroneus longus, tibialis anterior, and gluteus medius to a tilt platform perturbation.

Results: Those with pronated and supinated foot structures had significantly slower peroneus longus reaction times than those with neutral feet (P = .01 and P = .04 respectively). No significant differences were found for the tibialis anterior or gluteus medius.

Conclusions: Peroneus longus reaction time is influenced by foot structure. Further research is required to establish the consequences of slower peroneal reaction times in pronated and supinated foot structures. In future, researchers investigating lower limb muscular reaction time should control for foot structure as it may influence results.

4.7.2 Introduction

Medial longitudinal arch abnormalities are frequently associated with lower limb injuries (Kaufman et al., 1999; Mei-Dan et al., 2005; Williams et al., 2001); the most prevalent of which are lateral ankle sprains, with both low arches (Mei-Dan et al. 2005) and high arches (Williams et al., 2001) shown to be a risk factor. In the United States, ankle sprains have an estimated incidence rate of 2.15 per 1000 person-years (Waterman et al., 2010) with estimates of an annual healthcare cost of $2 billion (Soboroff et al., 1984). The UK also has a high incidence rate with as many as 302,000 patients attending Emergency Departments with ankle sprains each year (Bridgman et al., 2003). There is a clear demand for further research in this area both in terms of treatment, and in providing suggestions to reduce the rate of ankle sprains.

Arch height is related to foot function; a high arch is typically rigid and a characteristic of over supination, whereas a low arch is usually hypermobile and is related to over pronation (Franco, 1987). Previous studies have associated excessively pronated and supinated foot structures with deficits in some aspects of neuromuscular control in comparison to those with neutral feet (Cote et al., 2005; Tsai et al., 2006). Components of neuromuscular control include proprioception, muscle strength, postural control and muscular reaction time (Richie 2001). Both Cote et al. (2005) and Tsai et al. (2006) found evidence to suggest that postural control is affected by foot structure, however as yet; no previous study has assessed the effects of pronated and supinated foot structures on muscular reaction time.

The measurement of muscular reaction time to a tilt platform perturbation is a well-established method of analysing lower limb neuromuscular control. Either using a static (standing) platform (Beckman & Buchanan, 1995; Karlsson & Andreasson, 1992; Mitchell et al., 2008a; Ricard, Sherwood et al., 2000; Shima et al., 2005) or more recently, a dynamic (walking) platform (Hopkins et al., 2006; Hopkins et al., 2009), it is designed to simulate the mechanics of an inversion ankle sprain, therefore stressing the dynamic defence mechanism (Mitchell et al., 2008a). Typically, EMG measurements include the peroneus longus (Beckman & Buchanan, 1995; Karlsson & Andreasson, 1992; Mitchell et al., 2008a; Shima et al., 2005), and the tibialis anterior (Mitchell et al., 2008a) as these muscles have a direct role in the dynamic defence mechanism (Mitchell et al., 2008a). The gluteus medius has also been included to identify the use of hip strategy during perturbation (Beckman & Buchanan, 1995).

The purpose of this research was to determine whether pronated and supinated foot structures contribute to neuromuscular deficits as measured by muscular reaction time to a simulated ankle sprain mechanism. It was hypothesised that subjects with pronated and supinated foot structures would have significantly slower muscular reaction times than those with neutral foot structures.

4.7.3 Methods

Subjects:

Thirty volunteers participated in this study and were categorised into three groups which were dependant on navicular drop height measures. Subjects within each group were matched for height, mass, and sex. Prior to testing, ethical approval was granted from the institutional ethics committee, and all subjects read a subject briefing document and provided written informed consent. Subject characteristics are shown in Table 4.10.

Table 4.10 Subject Characteristics

Group	Sex	Age (years)	Height (cm)	Mass (kg)	Navicular Drop Height (mm)
Neutral	4 Males 6 Females	20 (2.1)	169.3 (8.8)	70.0 (11.2)	7.0 (1.0)
Pronated	4 Males 6 Females	21 (1.5)	169.8 (10.5)	70.4 (14.6)	11.0 (2.0)
Supinated	4 Males 6 Females	20 (1.6)	169.3 (6.2)	69.5 (10.7)	3.0 (1.0)

Values are mean (SD).

Subjects were included in this study if they were aged between 18 - 30 years, and took part in at least two hours of exercise each week. Subjects were excluded if any of the following applied; recent lower limb injury (within six months), history of lower limb surgery, history of ankle sprains, myositis ossificans, poor circulation, general illness, acute trauma to lower limb, soft tissue inflammation, skin infection, allergy to alcohol wipes, were under the influence of alcohol or any other psycho-active substance, or had regular use of orthotics, taping or bracing. Following recommendations by Tsai et al. (2006) to improve internal validity, subjects were also excluded if they took part in any activity involving regular balance training (e.g. ballet, gymnastics, Tai Chi) during the one year prior to testing, or for a total period of more than one year, during the 10 years prior to testing.

Subjects were categorised according to navicular drop height measures following a procedure described by Brody (1982) using the same height gauge (Axminster HG-1, United Kingdom) throughout. All measurements were taken by the same researcher as shown to be reliable during pilot testing (ICC = .97, SEM = 0.5 mm). Firstly, with the subject standing with feet shoulder width apart in a relaxed position, the most prominent point of the navicular tuberosity was marked with a pen. Secondly, the subtalar neutral position was found by pinching gently with the thumb and index finger either side of the talus, and asking the subject to slowly invert and evert their ankle until there was a feeling of equal pressure on both fingers. At this point the height of the marked navicular tuberosity from the ground was measured. Next, the subject was asked to return to a relaxed stance, and the height of the navicular tuberosity to the ground was measured again. The navicular drop height was then calculated by subtracting the relaxed stance measure, from the subtalar neutral measure. The average of three measurements was used to categorise subjects following guidelines by Cote et al. (2005) where ≥ 10 mm indicated a pronated foot, 5 – 9 mm indicated those with neutral arches and ≤ 4 mm indicated a supinated foot.

Procedures:

The tilt platform, as previously used by Mitchell et al. (2008a) comprised of two independently moveable platforms, each moving from a neutral position to 30° inversion and 20° of plantar flexion when released. Subjects were asked to stand on the tilt platform in a relaxed stance, looking directly ahead. Subjects were informed that between 6 and 8 tilts would occur and either ankle may be tested at any one time. Subjects were not told when the perturbations would occur, and they were initiated at variable intervals to reduce anticipatory effects. An average of three tilts on the dominant side was used for analysis. In accordance with Hoffman et al. (1998) the dominant side was defined as the foot used to kick a ball. If subjects were unsure, or stated that either foot could be used to kick a ball, the dominant leg was further defined as the leg the subject would prefer to recover balance on if pushed.

All testing was carried out using the same eight-channel DataLink EMG system (Biometrics, UK). Prior to the application of the electrodes, the skin was first prepared by shaving the area, then cleaned with an alcohol wipe to reduce skin impedance and left to dry. A passive reference electrode was placed on the radial styloid process, and pre-amplified surface bipolar electrodes (Biometrics SX230; gain x 1000, bandwidth 20 Hz – 460 Hz, noise < 5 µV, input impedance > 100 MΩ, common mode rejection ratio > 96 dB) were then positioned on the peroneus longus, the tibialis anterior, and the gluteus medius according to SENIAM guidelines in the direction of the muscle fibres. The EMG signals were sampled at 1000 Hz. Electrodes were not moved throughout the duration of the testing period.

Data Reduction and Statistical Analysis:

Raw EMG data was processed with an RMS filter using a 10 ms moving window. Data was then reduced using an Excel template (Microsoft Office, 2010) (Appendix Ten). Muscular reaction time was defined as the time between the onset of the tilt mechanism, to the onset of the EMG signal when it reached a level of 3 SD (Hopkins et al., 2009) above the baseline for 25 ms consecutively. The baseline value was the average value recorded over 150 ms immediately prior to the onset of the tilt mechanism.

As data were normally distributed (Shapiro-Wilk $P > .05$) a separate one-way analysis of variance was performed for each dependant variable (peroneus longus, tibialis anterior, gluteus medius) with foot structure (neutral, pronated and supinated) as the independent variable. Where a significant main effect was observed, a Dunnett's Post Hoc test was used to compare pronated and supinated foot structures against the neutral foot group. The a priori alpha level was set at .05. Effect size (η_p^2 values) was calculated where 0.01 – 0.06 indicated a small effect, 0.06 – 0.14 indicated a medium effect, and > 0.14 indicated a large effect. Observed power was also calculated. All statistical analyses were performed using IBM SPSS Statistics version 19.0 (SPSS Inc., Chicago, IL).

4.7.4 Results

The average reaction times across the different groups are shown in Table 4.11. Across the three groups, the peroneus longus had a reaction time range of 39 – 49 ms, the tibialis anterior ranged from 43 – 49 ms, and the gluteus medius ranged from 47 – 54 ms.

Table 4.11 Muscular Reaction Time Measurements

Muscle	Neutral Group	Pronated Group	Supinated Group
Peroneus Longus (ms)	39.6 (5.1)	49.7 (9.5)*	47.2 (5.8)*
Tibialis Anterior (ms)	43.6 (8.3)	45.7 (6.4)	49.2 (4.3)
Gluteus Medius (ms)	52.0 (10.2)	54.0 (10.9)	47.8 (7.2)

Values are mean (SD). *Indicates significantly slower reaction time in comparison to the neutral group ($P < .05$).

Analysis of the peroneus longus indicated a significant main effect of foot structure $F(2, 27) = 5.5$, $P = .01$, $\eta_p^2 = .29$, observed power = .82. Post hoc testing revealed a significant difference between the neutral and both the pronated group ($P = .01$) and the supinated group ($P = .04$), with both groups showing a slower peroneal reaction time in comparison to the neutral group. This is shown in Figure 4.5. No significant differences were identified within the tibialis anterior $F(2, 27) = 1.9$, $P = .17$, $\eta_p^2 = .12$, observed power = .35 or the gluteus medius $F(2, 27) = 1.1$, $P = .35$, $\eta_p^2 = .07$, observed power = .22.

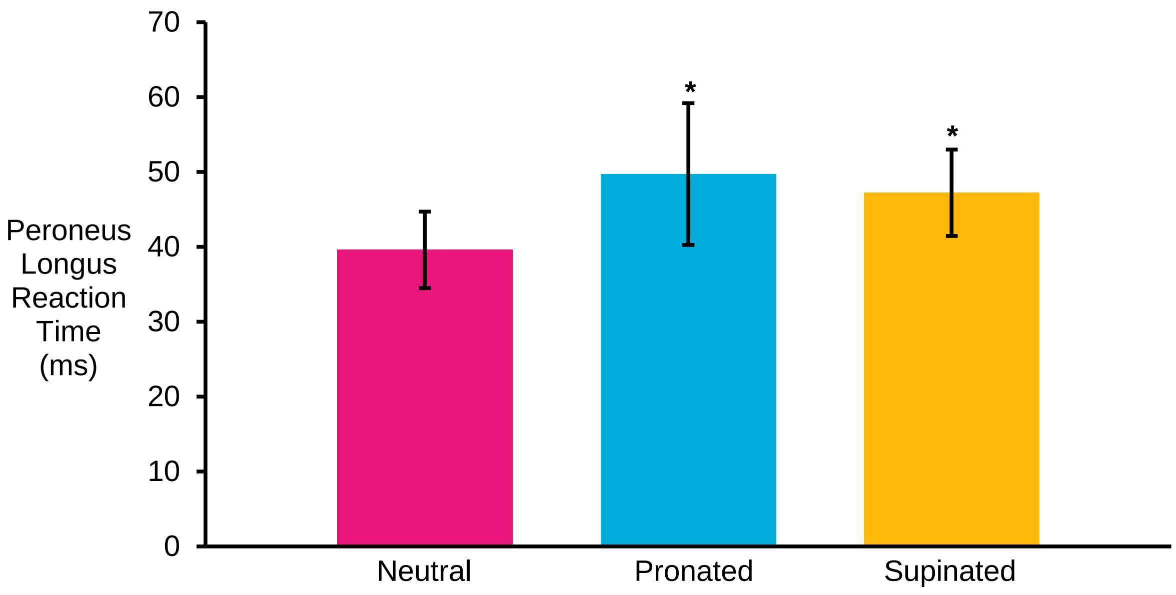


Figure 4.5 Peroneus longus reaction time (ms); showing standard deviation. ***** Indicates significantly slower reaction time in comparison to the neutral group ($P < .05$).

4.7.5 Discussion

This is the first study to have measured muscular reaction time to a tilt platform perturbation on subjects with pronated, supinated and neutral foot structures. The findings of this study showed that the mean peroneal reaction time of subjects with neutral feet was 39.6 ms, which was significantly faster ($P < .05$) than the reaction times of subjects with pronated feet and supinated feet, which were 49.7 ms (± 9.5) and 47.2 ms (± 5.8) respectively. The hypothesis that subjects with pronated and supinated feet have slower muscular reaction times of the peroneus longus than those with neutral feet was therefore accepted. In comparison to the neutral group, these values represent a 25% decrease in reaction time for the pronated group, and a 19% decrease for the supinated group. Previous studies have shown that delayed peroneal reaction times may mean that the muscles are incapable of protecting the ankle joint from sudden inversion (Richie, 2001). Whilst the incidence of ankle sprains was not in this study, the results could indicate that those with pronated or supinated foot structures may have an increased risk of obtaining a lateral ankle sprain in comparison to those with neutral feet. This is a highly important finding which has not been observed previously in the literature, and has significant implications considering that ankle sprains are among the most common sporting injuries with around 23,000 each day in the USA alone

(Kannus & Renström, 1991). Clearly a thorough epidemiological study observing foot structure and incidence of lateral ankle sprains needs to be conducted.

The theoretical basis for why pronated and supinated foot structures have slower peroneal reaction times in this study is unclear. As described by Johnson and Christensen (1999) the peroneus longus originates from the head and lateral shaft of the fibula and becomes tendinous in the middle third of the lateral compartment of the lower leg. The tendon inserts on to the plantar lateral surface of the base of the first metatarsal via a system of pulleys; the lateral malleolus, the peroneal tubercle, and the cuboid. With such a complex anatomical path in relation to other musculature it is unsurprising that slight biomechanical alterations in the foot may lead to changes in muscle activity. In addition to biomechanical differences in the foot, pronated and supinated foot structures may have repercussive effects in the lower leg; increased pronation results in excessive internal tibial rotation whereas increased supination leads to excessive external tibial rotation (Dawe & Davis, 2011).

Tiberio (1988) stated that a pronated foot reduces the mechanical advantage of the peroneus longus, perhaps due to slight shortening of the muscle in the pronated position (Franco, 1987; Karatsolis et al., 2009). In addition to the mechanical differences caused by different foot structures, previous studies have indicated EMG amplitude differences between neutral, and pronated foot structures during the stance phase of gait. Hunt and Smith (2004) found that those with pronated feet had lower EMG amplitude of the peroneus longus in comparison to neutral feet. Hunt and Smith (2004) admitted that whilst the differences were significant, they were only small; however this still may provide an insight as to why the reaction times of the pronated group were different to those with neutral feet.

Previous studies analysing other aspects of neuromuscular control have suggested differences between foot structures are caused by altered sensory feedback due to structural differences between the groups (Cote et al., 2005; Hertel et al., 2002; Tsai et al., 2006). For example a supinated foot has less ground contact than a pronated or neutral foot, and so theoretically receives less afferent input from the cutaneous receptors on the plantar surface which may affect how perturbations to stance are dealt with (Hertel et al., 2002). This seems like a feasible explanation for the slower reaction time of the supinated group, however as we did not measure plantar sensory input in this study, this remains speculative.

Previous studies have shown that patients with chronic ankle instability are more reliant on hip strategy rather than ankle strategy due to increased hypermobility in the ankle (Hertel, 2002). Ankle strategy involves shifting the centre of body mass by rotating the body about the ankle joint (Horak, 1987), whereas the hip strategy involves the use of the gluteus medius to correct posture (Leavey, Sandrey & Dahmer, 2010). It seems viable to assume that people with pronated or supinated feet may also be more reliant on hip strategy than a person with neutral feet due to decreased efficiency of the ankle strategy in maintaining balance. In this study, no statistical differences were identified in gluteus medius reaction time; this could be because the perturbation caused by the static (standing) tilt platform could be corrected by ankle strategy alone in all groups. It would be interesting to observe the reaction times of the gluteus medius on different foot structures during a more demanding task such as using a dynamic (walking) tilt platform as used by Hopkins et al. (2006) or during a perturbed dynamic landing task as used by Gutierrez and Kaminski (2012).

No significant differences were identified among foot structures for the tibialis anterior, indicating that foot structure does not affect the function of this muscle during an inversion and plantar flexion simulation. Previous studies have implied that foot structure may affect tibialis anterior function, for example Hunt and Smith (2004) identified a small increase in tibialis anterior EMG amplitude in pronated feet in comparison to neutral feet. In addition, some epidemiological studies have identified that differences in foot structure may predispose athletes to tibialis anterior strains (Williams 2001), implying that more stress is placed on the tibialis anterior muscle as a result of altered foot biomechanics. It could be argued again that the static (standing) platform used in this study did not place sufficient stress on the tibialis anterior muscle to elicit differences between foot structures.

In general the reaction times of this study appear to be faster than those reported in previous studies (Karlsson & Andreasson, 1992; Mitchell et al., 2008a; Shima et al., 2005). There are a number of reasons that could explain this, but as no previous study has investigated the effects of foot structure on muscular reaction time in this way it is difficult to make direct comparisons. One major factor is that in some previous studies, footwear has been worn (Shima et al., 2005) which has been shown to reduce the speed of inversion (Ricard, Schulties & Saret, 2000); therefore slowing the reaction time of the peroneus longus muscle. In contrast, the subjects in this study were barefoot. Further explanations for differing reaction times between studies are differences in tilt platform designs, for example, many studies in this area use tilt platforms which tilt in only one plane (Berg et al. 2007; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1991; Konradsen et al., 1993; Shima et al., 2005) whereas the tilt platform used in this study combines inversion and plantar flexion which is more applicable to the ankle sprain mechanism.

Additionally, variances in data analysis techniques may account for differences in reaction times among studies; currently there is no consensus in the literature on the best data analysis technique in studies using tilt platforms. Variables include the method used for exporting data, the length of the baseline measure, the number of standard deviations above the baseline measure, and the duration for which the burst is maintained. In some papers, these parameters are simply not reported (Konradsen & Ravn, 1991; Shima et al., 2005) making comparisons difficult; however the most widely reported value is the number of standard deviations above the baseline measure. This number has varied enormously ranging from 2 standard deviations (Berg et al. 2007; Konradsen et al., 1993) 3 standard deviations (Hopkins et al., 2009), 5 standard deviations (Cordova et al., 2010) to as high as 10 standard deviations above the baseline measure (Lynch et al. 1996). Few studies justify the parameters chosen, however Hodges and Bui (1996) advised that the standard deviation must be high enough to avoid a type I error, where the muscle is identified as active when it is not, yet low enough to avoid a type II error, where the researcher fails to identify the EMG onset when it occurs. There is a clear need for definitive parameters to be identified and used throughout this research area to enable accurate comparison of results in order to make valid conclusions.

4.7.8 Conclusions

The results of this study suggest that those with pronated and supinated foot structures have slower muscular reaction time of the peroneus longus than those with neutral feet. In light of the results of this study, future research should address whether those with pronated or supinated foot structure are more at risk of obtaining a lateral ankle sprain than those with neutral foot structures. Researchers investigating other aspects of muscular reaction time should control for foot type as differences among participants may interfere with results.

4.8 Development of Research

The work in this chapter has indicated that muscular reaction time to a tilt platform perturbation is a reliable method of measuring neuromuscular control with high levels of reliability across each component of the method including electrode placement, repeatability of the tilt platform and of the onset detection methods. In addition, Study Two has indicated that pronated and supinated foot structures have different neuromuscular responses to a simulated ankle sprain perturbation in comparison to those with neutral feet, potentially meaning that those with pronated or supinated foot structures are more vulnerable to injury during sport.

Before pursuing with the original aims of this thesis to examine the effects of foot structure and lower limb athletic taping on neuromuscular control, attention should been drawn to an interesting observation made during Studies One and Two.

4.8.1 Incidence of Asymmetry

Despite only analysing the dominant limb, measures were collected from both limbs for each subject during Studies One and Two. In doing so, it was observed that 31% of subjects had asymmetrical feet where one foot was classified as neutral, and the other foot was classified as either pronated or supinated. A breakdown of the number of participants in each subject group is shown in Table 4.12. The navicular drop height measures of each participant are identified in Appendix Twelve.

Table 4.12 Incidence of Asymmetrical Foot Structures

Foot Structure	N	Percentage
Asymmetrical	10	31%
Neutral	7	22%
Pronated	7	22%
Supinated	8	25%
Total	**32**	

This is an important observation, particularly in light of the results of Studies One and Two where foot structure on the dominant foot has been shown to affect aspects of neuromuscular control.

4.8.2 Progression of Thesis

In light of this interesting observation, Chapter Five will now address a new aim incorporated into this thesis:

- To examine the effects of asymmetrical foot structure on neuromuscular control.

Whilst this was not an original aim, it directly relates to the title of this thesis and warrants further exploration as findings may implicate further chapters.

Chapter Five

Asymmetrical Foot Structure, Limb Dominance and Lower Limb Biomechanics; A Secondary Analysis

5.1 Introduction to Chapter

In light of the recent observation of 31% of subjects in Studies One and Two having asymmetrical feet, this chapter will address the new aim added to this thesis; to examine the effects of asymmetrical foot structure on neuromuscular control.

The observation of asymmetrical foot structures poses an important question;

- Is neuromuscular control the same on both limbs in subjects with asymmetrical feet?

Before this question can be answered, in order to improve internal validity one must first exclude other potential factors which could influence results. In the current literature, the effects of limb dominance on neuromuscular control are unclear, however until this is established, one cannot conclude that potential differences in foot structure on contralateral feet are a result of differences in foot structure, or a result of limb dominance. This therefore poses a second question which must be answered initially;

- Does limb dominance affect neuromuscular control?

In attempt to answer these questions, this chapter includes two studies; Study Three examines the effects of limb dominance on neuromuscular control using only subjects with symmetrical foot structures; Study Four examines the effects of asymmetrical foot structures on neuromuscular control. Studies One and Two analysed data only from the dominant limb however data from both limbs were collected. Therefore Studies Three and Four are based on secondary data as defined by Heaton (1998) as "the use of existing data collected for the purpose of a prior study in order to pursue a research interest which is distinct from that of the original work".

Due to low subject numbers available for secondary analysis, this chapter is largely exploratory, and aims to address the high proportion of subjects included in this thesis with asymmetrical foot structures, which was unexpected during initial planning. It is intended that this chapter will highlight the need for further research in this area. Before commencing with Studies Three and Four, a thorough review of literature was conducted in order to examine current understanding of limb dominance and foot asymmetry and the effects on lower limb biomechanics.

5.2 Review of Literature on Limb Dominance and Biomechanical Symmetry

A number of search engines were used to acquire the relevant literature in this review including Scopus, Web of Knowledge and Google Scholar. Typical terms used in the search included limb *dominance*, *limb differences, muscle reaction time/postural stability on contralateral limb, asymmetrical foot structure.*

5.2.1 Defining the Dominant Leg

Often in biomechanical research, testing is carried out on a subject's 'dominant limb'. Traditionally the dominant limb is thought to have different neuromuscular properties than the non-dominant limb and so by measuring all subjects on the dominant limb it is thought to reduce potential variation, or enhance internal validity. Limb dominance is also often identified in studies where one leg serves as a control for the other leg (Hoffman et al., 1998). Although limb dominance is widely reported both in this area and other areas of sports science, there is no consensus on defining the dominant leg in current literature. Ambiguity of limb dominance may also have an impact in clinical settings where injuries are compared to the contralateral non-injured side. Hoffman et al. (1998) raised the question "What is a clinician to conclude if, during the course of progress evaluation, the balancing performance on the injured left leg is inferior to that of the dominant, uninjured right leg? Is this discrepancy due to the lingering effects of the injury, or could it be related to leg dominance?" (p. 319).

The concept of limb dominance has been discussed for over a century, with early researchers suggesting that if a left foot is used for kicking, the right foot must be dominant as it is chosen to support and steady the body (Gould, 1908). More recently, Previc (1991) suggested that one leg may not be more dominant than the other; rather one is simply used to carry out an action, whilst the other is used as support. Sadeghi, Allard, Prince and Labelle (2000) argued that in order to accept this theory one must assume that both the moving limb and the stabilising limb are placed under equal neurological demands. Instead Sadeghi et al. (2000) suggested that the limb initiating the movement usually requires more neurological skill than the supporting limb, and identified the preferred leg as the limb that is used for mobility, whereas the non-preferred leg was defined as the limb used for support. The main criticism for this is that some people may prefer to perform some movements on one leg, and use the other leg for other movements, additionally, sometimes the limb chosen to carry out a task (for example kick a ball) may be selected due to practicality rather than due to neurological reasons (Previc, 1991).

In sport there are a number of examples where it may be advantageous to be either left or right limb dominant. In athletics, 400 m hurdlers are encouraged to lead left legged regardless of limb preference in order to stay closer to the inside of the lane on an athletic track (Peters, 1988). In football a player is at an advantage if they can kick with both the left and right foot (Peters, 1988). Other potential influences on limb dominance include cultural pressures (Peters, 1988), for example being left handed was traditionally undesirable and children were forced to write with their right hands. Additionally some religions encourage limb preference, for example Hindus refrain from using the left hand when receiving gifts and when eating.

It is unclear whether upper and lower limb dominance are related; Previc (1991) identified that if the right arm is used for example when throwing, the left leg is in extension and typically used for support as the axial torque of the body is strongest during cross body motion (when using opposite leg and arm). On this basis, Previc (1991) also suggests that the limb used for antigravity flexion is usually the same side as the dominant upper limb. This is in partial agreement with Peters (1988), who stated that right handers have a clear preference for right foot kicking, however in contrast, according to Peters (1988) left handers do not appear to have a clear preference for kicking with the left foot. Previc (1991) concluded that the correlation between upper and lower limb dominance is high if the criteria for lower limb dominance distinguishes between flexion and extension. This claim is based upon the findings by Chapman, Chapman and Allen (1987) who devised a 13 item behavioural inventory of foot preference including a range of tasks such as ball kicks, standing on one leg, writing name with foot, and tapping a rhythm of a song with a foot. Previc (1991) identified that the tasks involving extension (for example standing on one leg) correlated poorly and were excluded from the overall inventory, whereas the tasks involving flexion (for example kicking) correlated highly.

In more recent literature, there are a number of ways in which researchers define the dominant limb; some studies simply ask subjects which is their dominant leg (subjective dominance), whereas others ask subjects to perform tasks (functional dominance) (Hoffman et al., 1998). Kicking a ball appears to be the most prevalent task subjects are asked to perform (Berg et al., 2007; Hopkins et al., 2006; Linford et al., 2006; Wikstrom et al., 2005). Whilst this method of defining limb dominance may be more applicable to sport than others, it could be argued that it only really applies to footballers. An alternative method to identify the dominant leg was used by Rothermel, Hale and Denegar (2004) who asked participants which leg they would prefer to catch themselves on if pushed. Hoffman et al. (1998) used three measures to identify the dominant leg; ball kick, step-up and balance recovery. Each test was performed three times, with the leg used for most trials identified as the dominant leg. It appears that more research is required to finalise the most appropriate functional test to determine limb dominance. For most valid comparisons, it is thought that studies using the dominant limb should only be compared to others with the same definition and the same functional tests used.

5.2.2 Limb Dominance and Neuromuscular Control

A number of studies have measured the effects of limb dominance on postural stability. After measuring limb dominance in three separate tests, Hoffman et al. (1998) concluded there were no differences between the dominant and non-dominant limb on ten subjects during static postural control. This is supported by Mitchell et al. (2008b) who found no differences between self-reported dominant and non-dominant limbs during a single leg stance. Mitchell et al. (2008b) suggested that more dynamic tasks may highlight potential differences between limb dominance, however Wikstrom Tillman, Kline et al. (2006) did not find any differences in a single limb forward hop task between healthy contralateral limbs. In a comprehensive study of balance, functional jumping performance, multi-joint coordination and proprioception, Ozer, Senbursa, Baltaci and Havran (2009) found no differences between dominant and non-dominant limbs.

Some studies have measured the effects of limb dominance on muscle reaction time to a tilt platform perturbation. Neither Mitchell et al. (2008a) nor Benesch et al. (2000) found significant differences in reaction time between healthy controls; however Fernandes et al. (2000) found that dominant legs had a significantly faster reaction time of 6.3 ms in comparison to the non-dominant leg during a tilt platform perturbation in male footballers.

Surprisingly, it appears at present there is little evidence to suggest that limb dominance effects neuromuscular control with only one study found to have shown a significant difference (Fernandes et al., 2000). More research is required in this area particularly in terms of postural stability where more challenging dynamic measures should be included.

5.2.3 Asymmetry in Lower Limb Biomechanics

Only two studies were found to have assessed limb symmetry using navicular drop height measures; Shultz and Nguyen (2007) concluded that in 32% of 100 cases the right foot could not be substituted for the left foot due to differences in navicular drop height which over exceed potential measurement error. This finding is very similar to the findings of Studies

One and Two where 31% of subjects were found to have asymmetrical feet. In addition, Brantingham, Adams, Cooley, Globe and Globe (2007) found some asymmetry between groups when measuring navicular drop height in subjects with lower back pain however the overall comparison found that one foot was not significantly different to the other. Using a different measure of foot type classification, Sforza, Michielon, Fragnito and Ferrario (1998) measured standardised foot prints of 46 adults and found no associations between footprint symmetry and age, body height, mass, or shoe size however on average, women had significantly more symmetric size-standardised footprints than men.

Given the number of biomechanical studies that use only one leg, it is surprising this area has not drawn more attention in current research. Such biomechanical studies are generally drawing a vast assumption that findings on one limb will equally apply to the other, however in light of the findings by Shultz and Nguyen (2007), and the similar observations of Studies One and Two, future researchers should be cautious as the effects of asymmetrical foot structures are largely unknown. Tsai et al. (2006) conducted the only study noted to have excluded subjects with asymmetrical foot structures as both feet were measured during a postural control task. In addition to researchers, practitioners in clinical settings should be cautious to use a non-injured limb as a control during assessment and rehabilitation unless foot structures on both limbs have first been identified. Shultz and Nguyen (2007) suggest that clinical judgment should be used when assessing whether differences between left and right foot structure are large enough to have clinical meaning.

5.3 Study Three: The Effects of Limb Dominance on Lower Limb Biomechanics; A Secondary Analysis

5.3.1 Abstract

Context: In the current literature, the effect of limb dominance is unclear, however in order to explore foot structure asymmetry on measures of lower limb biomechanics, the effects of limb dominance must first be ascertained.

Objectives: To measure the effects of limb dominance on lower limb biomechanics as measured by aspects of neuromuscular control using subjects with symmetrical foot structures.

Subjects: Of the thirty two volunteers whose data were available for secondary analysis; seven subjects had symmetrically neutral feet (navicular drop height 5 – 9 mm), seven subjects had symmetrically pronated feet (navicular drop height ≥ 10 mm), and eight subjects had symmetrically supinated feet (navicular drop height ≤ 4 mm). Ten subjects had asymmetrical feet and were therefore excluded from this study.

Methods: The dominant foot was defined as the foot used to kick a ball. Three dynamic hopping tasks were performed on both the dominant and non-dominant foot. Centre of pressure parameters were measured at 200 ms and three seconds including total distance, total velocity and 95% ellipse area. Peroneus longus reaction time to a tilt platform perturbation was also measured on both the dominant and non-dominant foot.

Results: No significant differences were identified between the dominant and non-dominant limb in measures of postural stability during the forward, diagonal or lateral hop, or within muscle reaction time measures.

Conclusions: According to postural stability and muscle reaction time measures, limb dominance does not appear to affect neuromuscular control. Therefore limb dominance can be excluded as a potential source of variation when comparing contralateral postural stability and muscle reaction time on subjects with asymmetrical foot structures.

5.3.2 Introduction

In order to maximise internal validity when measuring subjects with asymmetrical foot structures, the effects of limb dominance must first be examined in order to draw valid conclusions as to whether any potential differences between left and right feet in asymmetrical subjects are due to the asymmetry, limb dominance, or both.

Most previous studies indicate that limb dominance has no effect on measures of neuromuscular control (Hoffman et al. 1998; Mitchell et al., 2008b; Wikstrom, Tillman, Kline et al., 2006), however Fernandes et al. (2000) suggested it has a large effect, with non-dominant limbs having slower muscle reaction times than dominant limbs.

The purpose of this secondary research was to measure the effects of limb dominance on aspects of neuromuscular control using subjects with symmetrically neutral, pronated and supinated foot structures. It was hypothesised that limb dominance would have no effect on neuromuscular control as measured by postural stability and muscle reaction time.

5.3.3 Methods

Subjects:

Of the thirty two volunteers whose data were available for secondary analysis, ten were excluded from the current study due to having asymmetrical feet. Aside from this, the inclusion and exclusion criteria were the same as that used in Study One as identified in Section 3.2.3. As in Study One, subjects were split into three groups; neutral, pronated and supinated as measured by navicular drop height which was shown to be reliable in Section 3.2. Subject characteristics are shown in Table 5.1.

Table 5.1 Subject Characteristics

Group	Sex	Age (years)	Height (cm)	Mass (kg)	Navicular Drop Height (mm)
Neutral	3 Males 4 Females	18 (1.1)	172.7 (9.1)	67.1 (12.5)	7.0 (1.2)
Pronated	4 Males 3 Females	19 (1.1)	172.7 (6.7)	72.8 (10.4)	12.0 (1.3)
Supinated	5 Males 3 Females	21 (3.2)	173.1 (9.6)	70.1 (13.1)	2.0 (1.1)

Values are mean (SD).

Procedures:

As data from Studies One and Two were reanalysed for this study, the collection methods were the same as those identified in Sections 3.4.3 and 4.7.3, however data from the non-dominant limb was also included. For postural stability measures, only the three hopping movements; forward, diagonal and lateral were included. This was in order to optimise internal validity as these tasks resulted in the highest test retest reliability over 200 ms and three seconds during Pilot Test Two. For muscle reaction time measures, only the peroneus longus muscle was included as this was the only muscle used by Fernandes et al. (2000), and therefore the only muscle which could be compared to previous literature where a significant difference was identified.

Data Reduction and Statistical Analysis:

For postural stability, only the total distance, total velocity and 95% ellipse area parameters (as defined in Appendix Five) were included in this study. As all subjects had symmetrical feet within each group, and groups were not compared against each other, directional parameters were not of substantial interest. The same data reduction methods as identified in Section 3.4.3 were used in this study. For muscle reaction time, the same data reduction methods were used as identified in Section 4.7.3

As data were normally distributed (Shapiro-Wilk $P > .05$) paired t-tests were performed to test for significant differences between dominant and non-dominant data with a level of significance set at $P < .05$. Cohen's *d* statistics were also calculated to estimate effect size where 0.2 - 0.5 indicates a small effect, 0.5 - 0.8 indicates a medium effect, and > 0.8 indicates a large effect. All statistical analyses were performed using Excel (Microsoft Office, 2010).

5.3.4 Results

Results are shown in Tables 5.2-5.4. No significant differences in postural stability were found between dominant and non-dominant limbs during the forward, diagonal or lateral hops for any group at either 200 ms or three seconds. In addition, no significant differences were found between the peroneus longus reaction times of the dominant and non-dominant limbs.

Table 5.2 Peroneus Longus Reaction Time

Group	Dominant (ms)	Non-Dominant (ms)	*P*	*d*
Neutral	45.6 (3.2)	52.1 (5.6)	.32	1.42
Pronated	46.9 (9.7)	57.5 (9.7)	.24	1.13
Supinated	58.7 (9.5)	64.7 (7.5)	.26	0.70

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; *d*: Cohen's *d* measure of effect size.

Table 5.3 Postural Stability During Forward, Diagonal and Lateral Hop at 200 ms

		Neutral				Pronated				Supinated			
		Dominant	Non-Dominant	*P*	*d*	Dominant	Non-Dominant	*P*	*d*	Dominant	Non-Dominant	*P*	*d*
Forward Hop	T. Dist (cm)	22.8(5.0)	21.7(3.7)	.43	0.26	18.4(7.4)	18.7(5.4)	.80	0.04	23.1(11.4)	21.5(8.6)	.56	0.16
	V.Avg (cm/s)	113.9(25.1)	108.3(18.5)	.44	0.25	92.1(37.0)	93.3(27.1)	.81	0.04	115.5(56.8)	107.5(43.1)	.55	0.16
	95% EA (cm^2)	25.3(13.1)	16.4(7.5)	.11	0.84	19.0(5.1)	23.7(10.9)	.25	0.56	27.0(8.4)	22.3(8.3)	.08	0.60
Diagonal Hop	T. Dist (cm)	18.0(5.3)	24.0(6.5)	.16	1.01	18.3(4.2)	18.0(5.4)	.82	0.06	24.4(8.9)	22.8(8.2)	.19	0.19
	V.Avg (cm/s)	90.2(26.4)	120.1(32.4)	.16	1.01	91.4(21.1)	90.0(27.2)	.81	0.06	122.1(44.6)	113.9(41.1)	.19	0.19
	95% EA (cm^2)	32.3(15.8)	54.2(33.6)	.17	0.84	32.3(6.4)	40.4(23.5)	.39	0.47	29.1(11.8)	29.1(15.4)	.99	0.00
Lateral Hop	T. Dist (cm)	21.2(6.6)	21.7(4.0)	.80	0.09	17.4(3.6)	19.0(4.2)	.13	0.41	23.3(6.5)	23.6(7.0)	.93	0.03
	V.Avg (cm/s)	105.7(33.1)	108.4(19.8)	.79	0.10	87.1(17.7)	95.0(20.9)	.12	0.41	116.6(32.7)	117.7(35.2)	.93	0.03
	95% EA (cm^2)	58.2(23.5)	59.8(32.0)	.93	0.05	40.3(15.0)	49.3(22.2)	.39	0.48	33.6(8.3)	46.4(14.4)	.10	1.08

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; *d*: Cohen's *d* measure of effect size; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Table 5.4 Postural Stability During Forward, Diagonal and Lateral Hop at Three Seconds

		Neutral				Pronated				Supinated			
		Dominant	Non-Dominant	*P*	*d*	Dominant	Non-Dominant	*P*	*d*	Dominant	Non-Dominant	*P*	*d*
Forward Hop	T. Dist (cm)	75.0(7.3)	74.2(6.2)	.82	0.12	62.4(7.6)	65.0(6.1)	.15	0.37	72.4(14)	69.7(13.0)	.47	0.20
	V.Avg (cm/s)	25.0(2.4)	24.7(2.1)	.82	0.12	20.8(2.5)	21.7(2.0)	.15	0.37	24.1(4.7)	23.2(4.3)	.47	0.20
	95% EA (cm^2)	29.4(16.2)	31.8(13.0)	.65	0.16	26.7(6.4)	29.7(7.4)	.26	0.44	28.3(11.8)	31.7(13.8)	.58	0.26
Diagonal Hop	T. Dist (cm)	67.6(7.4)	75.0(17.6)	.26	0.55	64.9(7.4)	65.2(6.4)	.93	0.03	72.6(8.6)	71.6(5.6)	.64	0.14
	V.Avg (cm/s)	22.5(2.5)	25.0(5.9)	.26	0.56	21.6(2.5)	21.7(2.1)	.94	0.03	24.2(2.9)	23.9(1.9)	.64	0.14
	95% EA (cm^2)	29.0(9.5)	28.7(11.0)	.95	0.02	30.6(8.5)	35.7(9.1)	.17	0.58	34.4(14.4)	32.0(10.5)	.60	0.19
Lateral Hop	T. Dist (cm)	69.9(6.1)	76.6(8.6)	.12	0.90	64.0(8.6)	65.3(7.5)	.50	0.16	72.6(10.8)	70.8(8.8)	.59	0.18
	V.Avg (cm/s)	23.3(2.0)	25.5(2.9)	.12	0.90	21.3(2.9)	21.8(2.5)	.49	0.16	24.2(3.6)	23.6(2.9)	.60	0.18
	95% EA (cm^2)	32.8(3.4)	44.7(17.0)	.13	0.97	42.5(11.7)	40.0(11.8)	.59	0.21	33.5(13.4)	39.0(13.9)	.34	0.40

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; *d*: Cohen's *d* measure of effect size; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

5.3.5 Discussion

This is the first study to have measured the effects of limb dominance on dynamic postural stability and muscle reaction time using subjects with neutral, pronated and supinated foot structures. It was hypothesised that limb dominance would have no effect on postural stability or muscle reaction time; the hypothesis is therefore accepted as no significant differences were identified between dominant and non-dominant limbs during postural stability or muscle reaction time to a tilt platform perturbation.

Previous studies measuring the effects of limb dominance on static postural control found no differences between dominant and non-dominant limbs (Hoffman et al., 1998; Mitchell et al., 2008b) and suggested that more challenging dynamic tasks may be required in order to elicit differences between contralateral limbs (Mitchell et al., 2008b). Wikstrom Tillman, Kline et al. (2006) measured postural stability during a forward hop task and found no differences between dominant and non-dominant limbs. This study used the same forward hop task as used by Wikstrom Tillman, Kline et al. (2006), and also a diagonal and lateral hop, and found no differences between dominant and non-dominant limbs. Other tasks such as a forward run or 45° cut could be used as measures of postural stability; however these tasks naturally have more variance due to the dynamic nature of the movements therefore making it more difficult to identify small differences between dominant and non-dominant limbs.

Fernandes et al. (2000) was the only study found to have identified significant differences between the dominant and non-dominant limb. In the current study, the reaction times of the dominant limb were generally faster than the non-dominant limb, however as no significant differences were found; the findings do not support those of Fernandes et al. (2000). The effect size of both the neutral group and the pronated group were large indicating that the difference between groups is at least one standard deviation. This suggests that a larger number of subjects may be required, however as the current study was based on secondary analysis a limited number of subjects per group were available.

Interestingly the reaction times of the neutral group are faster than those of the pronated and supinated groups. Whilst this was not analysed statistically in the current study, it goes to support the work of Study Two where pronated and supinated groups were found to have significantly slower peroneus longus reaction times in comparison to the neutral group.

5.3.6 Conclusion

In conclusion, this study observed no differences between the neuromuscular response of dominant and non-dominant limbs as measured by postural stability and muscle reaction time. Therefore limb dominance can be excluded as a potential source of variation when comparing contralateral postural stability and muscle reaction time on subjects with asymmetrical foot structures. In light of this study, and several other studies which have found no differences between dominant and non-dominant limbs, it is suggested that when using subjects with symmetrical feet researchers and clinicians are safe to assume equal neuromuscular responses between dominant and non-dominant limbs.

5.4 Study Four: The Effects of Asymmetrical Foot Structures on Lower Limb Biomechanics; A Secondary Analysis

5.4.1 Abstract

Context: In previous studies over 30% of subjects have been found to have asymmetrical foot structures, where one foot is neutral, the other either pronated or supinated. No study has been found to have measured the effects of asymmetrical foot structures on neuromuscular control.

Objectives: To measure neuromuscular control on subjects with asymmetrical foot structures.

Subjects: Of the thirty two volunteers whose data were available for secondary analysis, ten had asymmetrical feet; six subjects had one neutral foot (navicular drop height 5 – 9 mm) and one pronated foot (navicular drop height ≥ 10 mm), and four subjects had one neutral foot (navicular drop height 5 – 9 mm) and one supinated foot (navicular drop height ≤ 4 mm). The remaining 22 subjects were excluded from this study due to having symmetrical feet.

Methods: Three dynamic hopping tasks; forward, diagonal and lateral were performed on both legs. Centre of pressure parameters were analysed at 200 ms and three seconds, including total distance, total velocity and 95% ellipse area. Peroneus longus muscle reaction time to a tilt platform perturbation was also measured on both legs.

Results: There were some apparent differences in the means between contralateral feet in both the neutral and pronated, and neutral and supinated groups in measures of muscle reaction time and postural stability. The pronated and supinated feet appeared to have poorer postural stability and slower muscle reaction times than the neutral feet.

Conclusions: Further research is required in order to understand the effects of foot asymmetry on neuromuscular control. Due to low sample numbers this study provides only preliminary data in the area of foot asymmetry and its effects on lower limb biomechanics. This area warrants further study with a much larger sample size to enable statistical analysis.

5.4.2 Introduction

The medial longitudinal arch is commonly measured in both scientific and clinical settings in order to determine whether a participant has neutral, pronated or supinated feet. The most common measurement taken to quantify the arch of the foot is navicular drop height measure. It was first identified by Brody (1982) as an indicator of excessive subtalar joint pronation measuring the displacement of the navicular bone between the subtalar neutral position and a relaxed stance. Alongside identifying excessive pronation, it has since been used to indicate neutral and excessively supinated foot structures where a navicular drop height of ≥ 10 mm indicates a pronated foot, 5 – 9 mm indicates those with neutral feet and ≤ 4 mm indicates a supinated foot (Cote et al., 2005). This measure has previously been shown to have very high reliability (ICC = .98, SEM = .5 mm) according to guidelines (Munro, 2000) as shown in Pilot Study One.

Shultz and Nguyen (2007) noted that 32% of subjects have been found to have asymmetrical foot structures, where one foot is neutral, the other either pronated or supinated. This finding is supported by unpublished observations of Studies One and Two where 31% of subjects participating in biomechanical research had asymmetrical feet. The implications of asymmetrical foot structures are unclear, no study has yet analysed the effects of postural stability and muscle reaction time on subjects with asymmetrical feet.

The purpose of this study was to measure the effects of foot structure asymmetry on aspects of neuromuscular control, specifically postural stability and muscular reaction time. Affecting such a high proportion of subjects in this thesis, asymmetry could not be ignored and so despite low subject numbers available in this secondary analysis, it was decided to be included in this thesis as a preliminary study. Following the results of Study One and Two, it was thought that the pronated and supinated feet would have higher postural stability and muscle reaction time measures indicating poorer neuromuscular control in comparison to the neutral feet.

5.4.3 Methods

<u>Subjects:</u>

This study used the ten subjects which were excluded from Study Three due to having asymmetrical foot structures. Subjects were divided into groups according to foot structure using navicular drop height measures as shown to be reliable in Section 3.2. The inclusion and exclusion criteria were the same as that used in Studies One and Two as identified in Section 3.2.3. Subject characteristics are shown in Table 5.5.

<u>Procedures:</u>

As the data from Studies One and Two were reanalysed, the collection methods were the same as those identified in Sections 3.4.3 and 4.7.3; however data from the non-dominant limbs were also included. As in Study Three, for postural stability, only the three hopping movements; forward, diagonal and lateral were included, and for muscle reaction time measures, only the peroneus longus muscle was included.

Table 5.5 Subject Characteristics

Group	Sex	Age (years)	Height (cm)	Mass (kg)	Navicular Drop Height (mm)	
Neutral & Pronated	2 Males 4 Females	21 (2.8)	167.7 (13.2)	68.0 (17.1)	Neutral Pronated	8.4 (0.7) 11.3 (1.9)
Neutral & Supinated	1 Male 3 Females	21 (2.1)	169.7 (10.3)	65.8 (8.1)	Neutral Supinated	5.6 (0.8) 2.5 (1.3)

Values are mean (SD).

Data Reduction and Statistical Analysis:

For postural stability, total distance, total velocity and 95% ellipse area parameters (as defined in Appendix Five) were included in this study. The same data reduction methods as identified in Section 3.4.3 were used to analyse data at both 200 ms and three seconds. For muscle reaction time, the same data reduction methods were used as identified in Section 4.7.3.

Due to low subject numbers which would result in low statistical power, only descriptive statistics were calculated. All analyses were performed using Excel (Microsoft Office, 2010).

5.4.4 Results

Results are shown in Tables 5.6-5.8. For the peroneus longus reaction times, the pronated and supinated feet are both slower than the neutral feet. At 200 ms, there are more observable increases in means in the pronated and supinated feet in comparison to the neutral feet than at the three second time frame.

Table 5.6 Peroneus Longus Reaction Time (ms)

Group	Neutral Foot	Pronated or Supinated Foot
Neutral & Pronated	52.7 (13.4)	59.3 (13.6)
Neutral & Supinated	60.7 (9.0)	70.0 (8.4)

Values are mean (SD).

Table 5.7 Postural Stability During Forward, Diagonal and Lateral Hop at 200 ms

		Neutral & Pronated		Neutral & Supinated	
		Neutral Foot	Pronated Foot	Neutral Foot	Supinated Foot
Forward Hop	T. Dist (cm)	23.9(10.2)	24.0(15.2)	19.5(9.1)	22.5(10.8)
	V.Avg (cm/s)	103.2(36.2)	88.9(24.6)	81.0(12.0)	112.8(54.1)
	95% EA (cm^2)	34.7(14.0)	24.7(7.0)	16.5(7.8)	20.4(7.8)
Diagonal Hop	T. Dist (cm)	22.7(13.4)	26.1(10.9)	23.2(7.8)	22.8(12.7)
	V.Avg (cm/s)	102.9(47.0)	125.9(46.5)	115.8(38.8)	114.1(63.2)
	95% EA (cm^2)	23.2(11.5)	30.7(9.0)	30.7(7.7)	32.0(18.7)
Lateral Hop	T. Dist (cm)	21.6(9.3)	26.6(8.3)	21.8(8.6)	20.7(4.6)
	V.Avg (cm/s)	108.0(46.3)	133.1(41.4)	109.1(43.1)	103.2(23.1)
	95% EA (cm^2)	45.1(35.7)	40.8(15.4)	35.2(8.4)	41.0(10.2)

Values are mean (SD). T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Table 5.8 Postural Stability During Forward, Diagonal and Lateral Hop at Three Seconds

		Neutral & Pronated		Neutral & Supinated	
		Neutral Foot	Pronated Foot	Neutral Foot	Supinated Foot
Forward Hop	T. Dist (cm)	75.7(11.8)	76.0(15.7)	66.5(8.2)	72.6(9.4)
	V.Avg (cm/s)	25.3(3.9)	25.8(6.1)	22.2(2.8)	24.2(3.2)
	95% EA (cm^2)	29.8(16.3)	32.6(20.1)	30.0(7.7)	28.3(10.3)
Diagonal Hop	T. Dist (cm)	78.8(17.4)	78.2(13.0)	72.6(6.6)	72.0(10.8)
	V.Avg (cm/s)	26.3(5.8)	26.0(4.3)	24.2(2.2)	24.0(3.6)
	95% EA (cm^2)	37.6(17.2)	28.5(16.8)	27.7(4.1)	28.3(4.5)
Lateral Hop	T. Dist (cm)	77.5(13.6)	79.8(10.9)	69.6(6.5)	68.8(5.0)
	V.Avg (cm/s)	25.9(4.5)	26.6(3.6)	23.2(2.2)	22.9(1.6)
	95% EA (cm^2)	33.4(10.3)	30.1(6.8)	33.8(11.6)	30.0(13.9)

Values are mean (SD). T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

5.4.5 Discussion

This is the first study to have measured the effects of dynamic postural stability and muscle reaction time on subjects with asymmetrical foot structures. Due to low subject numbers, this acts as a preliminary study which aims to promote further research in this area. It was thought that the pronated and supinated feet would have higher postural stability and slower muscle reaction time measures indicating poorer neuromuscular control in comparison to the neutral feet. Low subject numbers are known to affect the power of a statistical test; insufficient power may result in a type II error whereby the false null hypothesis is accepted (Vincent & Weir, 2012). Because of this, statistical differences were not calculated and so comparisons can only be made with descriptive data.

The mean muscle reaction time data indicate differences in both groups between the neutral foot and both the pronated and supinated foot albeit with high standard deviations. The findings therefore support those of Study Two as the neutral feet in both groups indicated faster reaction time in comparison to the pronated and supinated feet, however further analysis on a larger sample is required to confirm this.

The postural stability data indicates some large differences between the neutral and pronated foot, particularly at 200 ms where the neutral foot shows lower postural stability (indicating better postural control) in all three parameters in comparison to the supinated foot during the forward hop task, and in comparison to the pronated foot in the diagonal hop task. At three seconds the neutral foot shows lower measures in all three postural stability parameters in comparison to the pronated group during the forward hop task. This supports the results of Study One where the pronated and supinated groups indicated poorer postural stability than the neutral group in some tasks. In Study One, specific differences were identified between the neutral and supinated groups during the diagonal hop at 200 ms, and between the neutral and both the pronated and supinated groups during the lateral hop at 200 ms. In comparison, within the current study the supinated feet do not have increased postural stability parameters at 200 ms when compared to the neutral feet, however during the lateral hop at 200 ms, large differences were identified in the supinated feet with a higher 95% ellipse area in comparison to the neutral feet. Also in the pronated feet, where both the total distance and total velocity measures were higher than the neutral feet. Again statistical analysis on a large sample is required to confirm these observations.

In light of these findings, the initial observation of 31% of subjects having asymmetrical foot structures remains of high interest, particularly as few other studies have highlighted this.

Perhaps the incidence of asymmetrical foot structures is widely unreported due to many researchers only testing the foot structure of participants on one limb. In agreement with Shultz and Nguyen (2007), it is recommended that both limbs should now be measured and classified in clinical and research settings in order to ensure valid comparison.

In the current study, subjects had either one neutral foot and one pronated foot, or one neutral foot and one supinated foot. No subjects were identified to have one pronated foot and one supinated foot however it is possible that if a wider sample were available, this may be observed. No previous studies were found to have observed this before, however it is an interesting area for future research.

As emphasised in the introduction, a known limitation of this study was the low subject numbers available for secondary analysis, however as such a high proportion of subjects were identified to have asymmetrical feet, it was still thought to be worthwhile to explore the data. The results of this study reinforce the worth of this research area as they indicate that neuromuscular control is not the same on contralateral limbs in subjects with asymmetrical feet. To confirm these indications, this study needs to be repeated on a much larger scale in order to conduct statistical analysis.

5.4.6 Conclusion

The results of this study indicate that the contralateral limbs of subjects with asymmetrical feet may respond differently to both postural stability and muscle reaction time tasks, however this speculation needs to be confirmed with statistical analysis. Until the effects of asymmetrical foot structures are further researched, it is advised that researchers and clinicians do not generalise measures from one limb to the contralateral limb unless foot symmetry has been established.

5.5 Development of Research

Whilst diverting from the original aims of this thesis, the work within this chapter contributes to current literature by highlighting the incidence of foot structure asymmetry among a healthy sporting population. In addition this chapter has explored the effects of limb dominance and foot asymmetry on neuromuscular control. With reference to the two questions identified in the introduction of this chapter, Study Three has indicated that limb dominance does not affect neuromuscular control, and Study Four has indicated that neuromuscular control does not appear to be the same on the contralateral limbs of subjects with asymmetrical foot structures as measured by postural stability and muscle reaction time. Further research is required using a larger sample in order to confirm these findings. Once the effects of asymmetrical foot structures are established, suggestions can then be made in order to prevent potential injuries from occurring during sport such as the use of prehabilitation exercises, or interventions such as athletic taping. In line with the original aims of this thesis, the next chapters will assess whether lower limb athletic taping affects neuromuscular control on subjects with pronated, supinated and neutral foot structures.

Chapter Six

The Effects of Athletic Taping on Lower Limb Biomechanics

6.1 Introduction to Chapter

In light of Studies One to Four, it is clear that foot structure affects neuromuscular control as measured by muscle reaction time and dynamic postural stability. In addition, study three has highlighted that there appear to be no differences between the dominant and non-dominant leg in these measures. It is common in sporting practice to apply athletic taping to areas vulnerable to injury to provide protection, and also to help correct biomechanical abnormalities which may contribute to injury. Whilst athletic taping is common practice in sport, little is known about the effects of lower limb taping on neuromuscular control on athletes with different foot structures. Therefore Study Five aims to address this by measuring postural stability and muscular reaction time on individuals with neutral, pronated and supinated feet under four conditions; no-tape, arch tape, ankle tape and proprioceptive tape.

Before conducting Study Five, a thorough review of existing literature was conducted in order to further understand the scientific reasoning for current taping techniques used. Despite the many variations of lower limb athletic taping, this literature review focuses on arch and ankle taping. Arch taping was included as it attempts to hold the foot in a neutral position therefore correcting biomechanical abnormalities associated with pronated and supinated foot structures. In light of Study Two, ankle taping was also included in order to see whether it is a suitable preventative measure for subjects with pronated or supinated foot structures. Proprioceptive taping techniques are also discussed.

6.2 Review of Literature on Lower Limb Athletic Taping

A number of search engines were used to acquire the relevant literature in this review including Scopus, Web of Knowledge and Google Scholar. Typical terms used in the search included *athletic taping*, *closed basket weave*, *low-Dye* and *neuromuscular effects of taping*.

6.2.1 Arch Taping

Arch taping is a lower limb taping technique designed to support the medial longitudinal arch of the foot. It has been shown to produce a number of mechanical effects including changes in plantar pressure (Russo & Chipchase, 2001; Vicenzino et al., 2006), and changes in navicular drop height (Vicenzino et al., 1997). Plantar pressure measurements are regarded as the most objective method, where an increase in lateral pressure and a reduction in medial pressure indicates effective anti-pronation taping (Ator et al., 1991; Lange et al., 2004; Russo & Chipchase, 2001). Although often referred to as an 'anti-pronation' technique, O'Sullivan, Kennedy, O'Neil and Mhainin (2008) used 3D motion analysis and found evidence to suggest that arch taping restricts supination at the rearfoot, as well as pronation. As yet, there is little supporting evidence however this is a recent study using modern technology, and anthropometric methods which are more frequently used in research may be unable to detect specific movement of the rearfoot. Mechanical changes caused by arch taping have been shown to be effective in reducing pain in subjects with plantar fasciitis (Landorf, Radford, Keenan & Redmind, 2005); however pain in this study was measured using a visual analogue scale which can be criticised for being subjective.

6.2.2 Low-Dye Arch Taping

The traditional low-Dye technique created by Dr. R. Dye in 1937 has been shown to reduce symptoms associated with excessive pronation (Lange et al., 2004; Radford et al., 2006). Tape is applied below the ankle generating a supinating force that controls the amount of pronation occurring at the subtalar joint (Ator et al., 1991; Childs, Olson, McPoil & Cornwall, 1996). Several studies have found the low-Dye method to significantly alter peak and mean plantar pressure values; in particular, pressure has been shown to increase on the lateral midfoot (Lange et al., 2004; Scranton, Pedegana & Whitesel, 1982) suggesting a decrease in pronation.

The traditional low-Dye taping technique has been criticised for losing its effectiveness during activity (Vicenzino et al., 1997). Vicenzino et al. (1997) modified the technique, after suggesting that the original low-Dye lost its effectiveness due to the poor leverage exerted by tape solely applied to the foot. Figure 6.1 shows a comparison of the original low-Dye taping and the modified (or augmented) low-Dye taping technique. With the addition of calcaneal slings and reverse sixes which extend up the leg, Vicenzino et al. (1997) found that the tape provided better control of pronation immediately after application, and after an agility exercise protocol. Holmes, Wilcox and Fletcher (2002) support the short term use of the modified low-Dye technique and found that it reduced pronation by holding the subtalar joint in a near-neutral position; however identified that the tape did reduce in effectiveness after approximately 10 minutes of light exercise.

Additional studies investigating the modified low-Dye technique have shown it to significantly increase arch height when both standing still, and when walking (Franettovich, Chapman & Vicenzino, 2008; Vicenzino, Franettovich, McPoil, Russell & Skardoon, 2005). In a further study, Vicenzino, McPoil, and Buckland (2007) found the modified low-Dye technique to significantly alter plantar pressures of the foot by increasing both peak and mean maximal pressures in the lateral midfoot, and by reducing mean maximum pressure in the medial forefoot and rearfoot.

(i) (ii)

Figure 6.1 Comparison of the original low-Dye taping technique (i) and the modified low-Dye taping technique (ii) following procedures by Vicenzino et al. (1997).

6.2.3 Alternative Arch Taping Techniques

An alternative method of anti-pronation taping is the X-arch technique, like the original low-Dye the tape is applied below the ankle to support the arch (Perrin, 2005). A more commonly used modification of this is the 'double X-arch' (Del Rossi et al., 2004; Hunt et al., 2004; Hunt, Stowell, Alnwick & Evans, 2007). This technique has been shown to significantly increase arch height and reduce symptoms caused by excessive pronation, as well as symptoms of Sever's disease and plantar fasciitis (Hunt et al., 2004; Hunt et al., 2007). Other less popular methods of arch taping include 'circular arch', 'weave arch', and 'loop arch' (Beam, 2006), however there is little evidence to support the effectiveness of these techniques. A previous study found to compare alternative taping methods looked at both the low-Dye and double X-arch techniques, it concluded that both techniques are initially effective in increasing arch height; however both reduce in effectiveness after around 10 minutes exercise (Ator et al., 1991). More recently a study by Denyer and Mitchell (2012) compared the augmented low-Dye technique to the double X-arch taping technique and a temporary orthotic, and found that the augmented low-Dye technique was most effective in reducing excessive pronation (taping techniques are shown in Appendix Thirteen).

6.2.4 Reliability of Arch Tape Application

A common concern with previous studies which have used taping is that it is difficult to assess the reliability of the tape applied. Effectiveness of the taping can be enhanced by applying the tape directly to the skin, without use of an under-wrap, and by using adhesive spray (Ator et al., 1991; Vicenzino et al., 1997). Most studies attempt to control reliability of results by allowing only one tester to apply tape to all participants. However it is still difficult to ensure that each taping method is applied with exactly the same tension and at exactly the same anatomical positioning. Russo and Chipchase (2001) attempted to control the taping consistency by measuring taped plantar pressures over two consecutive days. The results were compared to an expert with 20 years of clinical experience and were then analysed to compare both intra and inter-tester consistency. Results showed high intra-tester reliability (ICC > .75) in all foot areas, and high correlations between the two tapers in medial and lateral plantar pressures (ICC > .75) however low correlations were prevalent in the midfoot. Lange et al. (2004) also measured intra-tester reliability of arch tape application using the same method, and found moderate to very high reliability (ICC .67 - .98) in 9 out of 10 areas of the foot, however as with Russo and Chipchase (2001) poor reliability was detected in the midfoot; it is therefore advised that midfoot plantar pressure data should be treated with caution.

6.2.5 Potential Limitations of Arch Taping

Arch taping is opposite to the medial-lateral application of ankle tape used to prevent lateral ankle sprains (Vicenzino et al., 2006). Arch taping has therefore been highlighted as a potential risk factor for causing lateral ankle sprains as the technique effectively pulls the subtalar joint out of excessive pronation towards supination. Vicenzino et al. (2006) measured plantar contact area during standing, walking and running, and found that although augmented low-Dye taping altered static foot posture, there were minimal changes in plantar contact area measures during walking and running, indicating that arch taping does not appear to put athletes at an increased risk of an ankle sprain during sport. This is the only study found to have assessed the potential risk of ankle sprains when wearing arch taping, this study could be improved by measuring the effects of taping during a movement more specific to the ankle sprain mechanism, such as a tilt platform perturbation.

6.2.6 Arch Taping or Orthotics?

Orthotics are often used as an alternative technique to arch taping aimed at reducing lower limb structural deformities (Franco, 1987). Alongside taping, orthotics have been shown to reduce lower limb pain (Nicolopoulos et al., 2000), and effectively limit excessive pronation (Del Rossi et al., 2004), however moulded orthotics are more expensive than taping methods, and can be time consuming to produce (Branthwaite, Payton & Chockalingam, 2004). Several studies have compared the use of orthotics to taping as a means of reducing excessive pronation. Vicenzino, Griffiths, Griffiths and Hadley (2000) compared the modified low-Dye technique to a temporary orthotic and found that both methods significantly increased arch height, compared to the control. Some studies have shown that orthotics can provide a fast return to sport (Nicolopoulos et al., 2000); Hunt et al. (2007) however, found that taping allowed an earlier return to sport compared to other conservative methods including orthotics. A study by Hadley, Griffiths, Griffiths and Vicenzino (1999) found that taping was superior to orthotics in controlling tibial rotation caused by pronation, both after application of tape and during exercise. Scranton et al. (1982) compared barefoot gait, to gait with a medial arch orthotic support and low-Dye taping and found that the low-Dye taping provided support to the medial arch, which greatly reduced force on the medial plantar fascia and plantar tarsal ligaments. The medial arch orthotic shifted the plantar pressure forces laterally, however did not reduce the strain to the same extent as the taping. The study concluded that both taping and medial arch orthotics can change injury pattern, potentially treating early symptoms of overuse injuries. This study only observed plantar pressures when walking across a force plate, so the findings cannot be applied to more dynamic sports. Some studies suggest that taping should be used prior to the use of orthotics to ensure that the orthotics will produce the desired effect, and to save the patient time and money (Smith, Brooker, Vicenzino & McPoil, 2004; Vicenzino, 2004).

6.2.7 Ankle Taping

Ankle taping is widely used in sport for the prevention of injuries, immediately after an ankle injury as a form of compression, and during rehabilitation of ankle injuries for protection (Karlsson, Swärd & Andreasson, 1993; McCluskey, Blackburn & Lewis, 1976; Perrin, 2005). Ankle taping has often been chosen simply by athlete’s preference or superstition (Callaghan, 1997), however with increasing research in the area, there is now a strong scientific argument for wearing ankle taping. Ankle taping is designed to relieve excessive stresses on ankle joint ligaments, whilst enabling normal joint function (Wilkerson, 2002). Ankle taping has been shown to reduce range of motion in both inversion and eversion, as well as plantar flexion and dorsi flexion (Paris, Vardaxis & Kokkaliaris, 1995). Reduced range of motion as a result of ankle taping indicates increased stability at the ankle joint (Riemann et al., 2002).

6.2.8 Basket Weave Taping

There are a number of different ankle taping techniques used, both in clinical practice, and in biomechanical research. Variations in the methods of application make comparisons of different taping studies difficult. One of the most widely used techniques is the basket weave, which was devised by Gibney (1895). It was originally designed as a treatment for ankle sprains but is now commonly used as a preventative measure (Delahunt, O'Driscoll & Moran, 2009; Perrin, 2005; Sawkins, Refshauge, Kilbreath & Raymond, 2007). There are several variations of the basket weave; open basket weave taping does not join anteriorly, leaving a small gap at the front of the lower leg and ankle, and is therefore useful for taping acute sprains by allowing space for swelling without compromising blood flow (McCluskey et al., 1976; Perrin, 2005). On the other hand, the closed basket weave covers the lower leg and foot entirely. Both techniques restrict inversion and eversion at the ankle, however in addition, the closed basket weave also reduces plantar flexion and dorsi flexion (Paris, 1995), therefore making the technique more robust than the open basket weave.

Although some researchers prefer the use of the open basket weave (Ozer et al., 2009), the closed basket weave technique is most prevalent in current literature (Delahunt et al., 2009; Karlsson & Andreasson, 1992; Martin & Harter, 1993; Sawkins et al., 2007). There appears to be much variation with regards how the technique is applied, and also the number of repetitions for the various components, although most ankle taping techniques use 38 mm zinc oxide athletic tape (Shima et al., 2005). Generally, the closed basket weave consists of two anchor strips, one around the base of the calf, and the other slightly posterior to the base of the fifth metatarsal (Delahunt et al., 2009; Perrin, 2005; Sawkins et al., 2007). Following the anchor strips, stirrups are usually placed from the medial surface on the superior anchor strip, wrapping underneath the foot, to the lateral surface of the same anchor strip. A horseshoe strip is then placed from the medial surface of the inferior anchor strip, around the heel, attaching to the lateral surface of the anchor strip. Taping is usually applied in a medial to lateral direction to evert the foot away from the inversion ankle sprain mechanism, towards pronation (Martin & Harter, 1993), however tape may also be applied in a lateral-medial direction for the prevention of eversion ankle sprains (Perrin, 2005). The stirrups and horseshoes are repeated three times, in a weaving manner, overlapping the previous strip by approximately one centimetre (Delahunt et al., 2009; Perrin, 2005).

Much of the variation in the literature occurs at this point, Perrin (2005) recommends closing the leg with horizontal strips and then applying medial and lateral heel locks, followed by two figure of eight strips; this is in agreement with Ricard, Sherwood et al. (2000) and Shima et al. (2005). This technique is shown in Figure 6.2. Delahunt et al. (2009) on the other hand applied a figure of 8 with pre-wrap before applying the basket weave pattern, next applied two heel locks and two subtalar slings before applying locking strips to fill in any gaps. Wilkerson (2002) supports the use of subtalar slings; however Wilkerson (2002) used semi-elastic tape to resist subtalar inversion, rather than non-elastic tape. Alternatively, Karlsson and Andreasson (1992) used a closed basket weave technique without a heel lock; it was stated that this was because they were interested in the functional properties of the tape, however it could be argued that the functional properties of the tape can still be assessed using a complete taping technique. It has been suggested that ankle taping may cause skin irritation (Verhagen, van der Beek, & van Mechelen, 2001). Karlsson et al., (1993) advised that tape should be applied in strips rather than a continuous piece as this may result in unequal pressure on the skin, and may impair blood flow. Karlsson et al. (1993) also advised that tape should be removed immediately after exercise; additionally Perrin (2005) suggests that tape applied to acute sprains should be removed if ice is applied to the ankle, and should not be worn overnight.

Figure 6.2 The closed basket weave ankle taping technique as described by Perrin (2005).

6.2.9 Comparison of Ankle Taping Techniques

Several studies have compared the various ankle taping techniques; Rarick, Bigley, Karst and Malina (1962) compared four methods of ankle taping and found the basket weave in combination with stirrups and heel locks most effective in providing mechanical support. Using a wooden model of a foot, Pope, Renström, Donnermeyer and Morgenstern (1987) compared four different types of ankle taping, and found the basket weave with three layers of a figure of eight to be most effective at withholding a tension of 420 N, which was suggested to be the typical torque applied to the ankle during sports. It was suggested that failure of the tape occurred due to loss of contact with the skin rather than breakage of the tape. However, whilst it may be more ethical to test the tension of tape on a wooden foot than to risk injury on a human foot, the methodology of this study can be criticised as the results may not be applicable to a human ankle. Frankeny, Jewett, Hanks and Sebastianelli (1993) also compared four different ankle taping techniques and found the Hinton-Boswell technique to be most effective in restricting inversion, however no supporting literature was found using this technique. The basket weave technique was found to be second most effective. The results of this study can also be questioned as the various taping techniques were compared using a non-weight bearing test, therefore not reflecting the nature of an inversion ankle sprain.

6.2.10 Ankle Taping and Under-Wrap

Some researchers have combined zinc oxide taping with under-wrap, or with elastic tape (Perrin, 2005; Wilkerson, 2002). Under-wrap is used to help with the removal of the taping technique, particularly among male subjects due to leg hair (Karlsson, et al., 1993; Sawkins et al., 2007), though many researchers advise not to use under-wrap as it prevents the tape from securely adhering to the skin reducing the ‘pulling effect’ (Kaminski & Gerlach, 2001). In a previous study however, under-wrap was not found to influence the amount of inversion when compared to tape applied directly to the skin (Ricard, Sherwood et al., 2000). Refshauge, Kilbreath and Raymond (2000) advise placing tape directly onto the skin without under-wrap in order to assure the strongest cutaneous cues.

In sports where athletes wear studded boots, such as rugby or football, athletes may prefer to have athletic taping over the boot; this technique is known as spatting (Pederson, Ricard, Merrill, Schulthies & Allsen, 1997). In a comparison of three techniques; spatting alone, traditional taping alone and a combination of the two, Pederson et al. (1997) found that the combination of the two was most effective in reducing the rate of inversion before and after exercise, followed by spatting alone, and then taping alone. This is an interesting finding, and is often used in practice, however unfortunately there is little supporting literature.

6.2.11 Reliability of Ankle Tape Application

As with arch taping, it is difficult to measure the reliability of ankle taping techniques, in terms of the tension applied by the technique, and also in terms of anatomical positioning. Most studies ensure that the same experienced tester applies the tape on each subject (Kaminski & Gerlach, 2001; Martin & Harter, 1993; Paris et al., 1992; Riemann et al., 2002; Shima et al., 2005), however no studies were found to have attempted to measure the reliability of ankle tape application.

6.2.12 Potential Limitations of Ankle Taping

One major concern with the prolonged use of ankle taping, is that excessive use could potentially weaken the muscles around the ankle joint, therefore making the ankle more at risk of injury if tape is not worn (Callaghan, 1997). However Cordova, Cardona, Ingersoll and Sandrey (2000) found no evidence to suggest that long term use of an ankle brace will affect the peroneus longus muscle during a sudden inversion tilt perturbation, so it could be speculated that the same would occur with the long term use of taping, however this is yet to be researched.

A further concern for the use of athletic taping is that it may impair performance. Paris et al. (1992) compared performance when wearing ankle bracing and taping over four dynamic tasks, a 50m sprint, a balance task, an agility task, and a vertical jump task. The study showed that there were no differences in performance during the 50m sprint, the balance, or the agility tasks for both the tape and bracing, however during the vertical jump task; a significant decrease in jump height was prevalent with the use of the ankle brace. This study implies that ankle taping may be of more benefit to an athlete in comparison to bracing as it provides ankle support without having a detrimental effect on performance. Sawkins et al. (2007) found similar results when assessing the effect of ankle taping on two functional tests on subjects with functional ankle instability; it was suggested by Sawkins et al. (2007) that the dynamic hopping task, and the star excursion balance test may not have been sensitive enough to identify differences that may have occurred between the conditions. During a landing task, Riemann et al., (2002) found that with the use of ankle taping, time to reach peak forces was significantly less when compared to a no-tape control, this suggests that higher stresses are imposed on the musculoskeletal system during dynamic activity with ankle taping due to decreased energy absorption on impact. This is consistent with Yi et al. (2003) who indicated that the use of ankle taping may result in injury during landing due to restriction to ankle motion causing a lack of shock absorption on impact.

6.2.13 Ankle Taping or Bracing?

Refshauge et al. (2000) suggested taping is used more frequently than bracing as it is a cheaper alternative and the benefits are similar, however the long term use of taping has been criticised for being expensive compared to other techniques such as bracing. Paris et al. (1992) highlighted that on average, approximately two thirds of a roll would be used to tape one ankle using the closed basket weave technique, with each roll costing around $2 (Canadian), equating to approximately £1.20. Pope et al. (1987) calculated that some American football teams may spend up to $50,000 on athletic tape each season, although this is not exclusively for ankle taping. Alt, Lohrer and Gollhofer (1999) assessed the effectiveness of two ankle taping techniques; the traditional closed basket weave, and a modified shorter version using approximately 1.5m less tape and concluded that although it was less expensive, the shorter version was not as effective in stabilising the ankle joint. The literature is inconclusive as to whether bracing is as effective as taping; a cadaveric study showed that braces are not as effective as newly applied tape; however the ankle brace lasted longer than the taping technique which deteriorated with use (Shapiro, Kabo, Mitchell, Loren & Tsenter, 1994). Martin and Harter (1993) compared the use of ankle taping to a lace up brace, and a semi-rigid orthotic, and found that the taping technique was unable to restrict inversion after a 20 minute period of exercise on a laterally tilted treadmill, supporting the use of bracing. Ozer et al. (2009) on the other hand could not identify superiority between taping and bracing suggesting that they may both play an important role in supporting the ankle joint during activity.

6.2.14 Ankle Taping and Foot Structure

As yet, no studies have been found to assess the effects of ankle taping on different foot types. This is surprising considering the fact that ankle taping is applied in such a way that it moves the foot into pronation (Martin & Harter, 1993; Nishikawa et al., 2002); so theoretically, taping may have a different effect on a subject with pronated feet, in comparison to a subject with neutral or supinated feet, however this is yet to be researched. In light of the findings of Study Two, this area requires further research particularly if ankle taping is to be suggested as a preventative measure to reduce the risk of ankle sprains on individuals with pronated and supinated foot structures.

6.2.15 Neuromuscular Effects of Lower Limb Taping

Most studies on arch and ankle taping highlight the mechanical effect of taping on the joint (Ator et al., 1991; Holmes et al., 2002; Lange et al., 2004; Lohrer et al., 1999; Martin & Harter, 1993); however there is evidence to suggest that athletic taping may also produce a neuromuscular effect. This was thought to occur as it was found that the stabilising effect of ankle taping is still present, even after periods of exercise when the mechanical effects of the taping had reduced (Robbins, Waked & Rappelt, 1995). It has been suggested that ankle taping may prevent injury through improving components of neuromuscular control. Robbins et al. (1995) propose that by uniting the skin of the foot and the leg with ankle taping it provides cutaneous cues of plantar surface position and orientation through traction of the tape on the skin. Matsusaka, Yokoyama, Tsurusaki, Inokuchi and Okita (2001) suggest that the disturbance of the afferent input from mechanoreceptors would affect not only a sense of movement and position, but also the subsequent neuromuscular reflex to control posture and coordination. In accordance with this, the effects of taping on lower limb neuromuscular control have been measured in literature using postural stability, muscle reaction time and joint position sense measurements (Leanderson et al., 1996; Kaminski & Gerlach, 2001; Karlsson & Andreasson, 1992).

6.2.16 Taping and Postural Stability

Several previous studies have used postural stability to measure the effects of taping on neuromuscular control (Bennell & Goldie, 1994; Leanderson et al., 1996; Yi et al., 2003). Leanderson et al. (1996) showed that ankle taping successfully reduced postural sway, showing both decreased mean and maximum postural sway in the mediolateral and anteroposterior directions. On the other hand, Bennell and Goldie (1994) found that ankle taping had adverse effects of postural stability during a single leg stance; subjects were less steady and had more frequent touch downs that when performing the task with no taping. Yi et al. (2003) also measured the effects of ankle taping on postural stability, and found that vertical ground reaction force significantly increased when compared with untapped trials during a drop from a 40 cm bench onto a force platform, Yi et al. (2003) concluded that this may have detrimental effects to the athlete.

Few studies have examined the neuromuscular effects of arch taping, however a study by Wall et al. (2005) tested modified low-Dye taping during four functional tasks (back pedal, cutting, drop jump and hopping), it concluded that arch taping may support the subtalar joint by increasing the efficiency of force dissipation on landing. No other studies were found to assess the effects of arch taping on postural stability; however Janin and Dupui (2009) looked at the effects of a medial arch orthotic and found that during bilateral stance, plantar pressure increased in the contralateral foot, and COP shifted towards the opposite direction to that of the arch support. Janin and Dupui (2009) suggested that stimulation of the plantar surface may cause the perception that the centre of mass has shifted towards the stimulated foot, causing a shift in the COP in the opposite direction in order to maintain stability. As several studies have reported a similarity between the effects of arch taping and medially placed orthotics (Scranton et al., 1982; Vicenzino et al., 2000), it is possible that these results may also occur with the use of arch taping; however this is yet to be examined. More research is required to confirm the effects of both arch and ankle taping on postural stability.

6.2.17 Taping and Muscle Reaction Time

Several studies have used muscle reaction time to measure the effects of taping on neuromuscular control. Karlsson and Andreasson (1992) measured the effects of ankle taping on the reaction time of the peroneal muscles during a simulated ankle sprain, and found that reaction time was significantly reduced. This is an important finding as it suggests that ankle taping can speed the activation of the dynamic defence mechanism which is used to counteract movements leading to a lateral ankle sprain. However as reported by Karlsson and Andreasson (1992), these results did not occur with all subjects; in 6 of the 20 subjects involved in the study showed insignificant shortened peroneal reaction time.

Conversely, Shima et al. (2005) found that ankle taping delayed peroneal reaction time, it was speculated that this may be due to mechanical restriction of the tape, which would slow the speed of inversion, therefore reducing the activation of the peroneals. Shima et al. (2005) did not measure velocity of the sudden inversion; however the theory is in support of Alt et al. (1999) who also found a decrease in peroneus longus stretch reflex caused by a decrease in inversion velocity found with ankle tape application. Ricard, Sherwood et al. (2000) also found a decrease in both average and maximum inversion velocity as a result of ankle taping. Lohrer et al. (1999) stated that by decreasing the speed of inversion, it allows time for the functional reflexes to initiate in time to protect the joint from injury. Although the effects of ankle taping on muscle activation have been studied, research is inconclusive; a wider range of lower limb muscles should be measured in order to understand the effects of ankle taping on muscular reaction time.

Some other studies have measured changes in muscle activity on the contralateral limb as a result of taping; Loos and Boelens (1984) tested peroneal muscle activity with tape over the left ankle during three tasks; bilateral stance, unilateral stance, and unilateral stance with 2 kg weight. The study showed that during the bilateral stance, there was no increase in muscle activity on the right leg; however the activity of the peroneus longus on the left leg increased significantly during the unilateral stance with and without the extra weight. This suggests that although taping may not influence muscle activity on the contralateral limb, it does provide support for the neuromuscular effects of taping on the limb to which it is applied.

No studies were found to assess the effects of arch taping on muscular reaction time to a tilt platform perturbation; however some studies have measured the effects of arch taping on muscular activity. Franettovich et al. (2008) tested subjects with pronated feet and found a decrease in muscular activity of the tibialis anterior and posterior during walking, indicating that the taping method is successful in reducing the load of these muscles. Also using subjects with pronated feet, Wall et al. (2005) found significantly lower peak muscle activity of the soleus and peroneus longus during dynamic tasks with the use of arch taping. This finding of reduced peroneus longus activity is of particular interest as it implies that with arch taping the muscle may not be able to contract sufficiently in order to protect the ankle joint by moving it out of a vulnerable plantar flexion and inversion position which may result in an ankle sprain. However, as yet, no studies have measured the effects of taping on subjects with neutral, pronated and supinated arch heights when assessing muscle reaction time to sudden ankle inversion.

6.2.18 Taping and Joint Position Sense

Measures of joint position sense have been widely used in literature to examine the effects of taping (Kaminski & Gerlach, 2001; Robbins et al., 1995; Spanos, Brunswic & Billis, 2008). Kaminski and Gerlach (2001) compared the closed basket weave ankle taping technique with a neoprene ankle support and found that neither technique were effective in enhancing joint position sense. It may be that the tests for joint position sense are not sensitive enough to detect alterations in neuromuscular control in healthy subjects; Kaminski and Gerlach (2001) suggested that further research needs to target those with neuromuscular deficits such as those with functional ankle instability. Refshauge et al. (2000) used subjects with recurrent ankle sprains, and tested the ability to detect passive movements in plantar flexion and dorsi flexion. It was hypothesised that ankle taping would increase cutaneous cues therefore enhancing the perception of movement, however this study found that taping had no effect on movement perception on subjects with recurrent ankle sprains. As suggested by the researchers, it may be that no differences were found in this plane of movement as ankle taping is designed to primarily restrict inversion and eversion movements. In a more recent study, Refshauge, Raymond, Kilbreath, Pengal and Heijnen (2009) tested joint position sense on subjects with recurrent ankle sprains and found that ankle taping reduced the ability to detect inversion and eversion movements, suggesting a reduction in neuromuscular capabilities as a result of the ankle taping.

Contrary to Refshauge et al. (2009), Spanos et al. (2008) found that ankle taping improves ankle position awareness among athletes with a history of inversion ankle sprains. This was measured in a non-weight bearing position which was suggested to be most representative of an ankle sprain when compared with full weight bearing, as most ankle sprains occur when the foot is just touching the ground before full weight bearing occurs, however this is a questionable justification. Robbins et al. (1995) also found that ankle taping improves neuromuscular control at the joint by improving the judgement of position and orientation of the plantar surface. No study was found to have measured joint position sense with the use of arch taping. The results of joint position tests are clearly inconclusive. As discussed in Section 2.6.2, other measures such as postural stability or muscle reaction time are thought to be more functional assessments of neuromuscular control.

6.2.19 Kinesio Tape

In recent years there has been an increase in popularity of kinesio tape. Kinesio tape is elastic and can be left on the skin for up to five days. It is claimed to improve local circulation, reduce oedema, facilitate muscle activity, and improve joint function by enhancing sensory mechanisms (Briem et al. 2011). Few studies were found to have measured the effects of kinesio tape on the foot and ankle, however Briem et al. (2011) compared kinesio tape to non-elastic tape and a no-tape condition and found that kinesio tape did not affect muscle activity during a sudden inversion perturbation; however muscle activity of the peroneus longus did increase with the non-elastic tape in comparison to the no-tape condition. Halseth, McChesney, DeBeliso, Vaughn and Lien (2004) used joint position sense to measure the effects of kinesio taping on neuromuscular control and found that it had no effect in comparison to a no-tape control. No study was found to have measured the effects of kinesio tape on dynamic postural stability. Despite popular use it appears that kinesio tape has little scientific backing in terms of muscle facilitation however further research is required particularly using measures of postural stability to verify this.

6.2.20 Evidence for Neuromuscular Effects of Taping

Despite the lack of conclusive research on the neuromuscular effects of lower limb taping, there is evidence to suggest that the application of tape has neuromuscular effects elsewhere in the body, for example the shoulder (Morrissey, 2000) and the quadriceps (Macgregor, Gerlach, Mellor & Hodges 2005; Tobin & Robinson 2000), so it would seem feasible that the same may occur with arch and ankle taping. By applying tape across the patella, Macgregor et al. (2005) noticed an increase in vastus medialis oblique (VMO) muscle activity indicating that cutaneous stimulation caused by the tape affects muscle function. The vastus lateralis was also measured in this study however it was not affected by the tape. Using a different quadriceps taping technique, Tobin and Robinson (2000) found that vastus lateralis activity could be inhibited, however concluded that further research was required to determine the effects of taping on VMO activity.

Two studies were found to have examined the neuromuscular effects of taping at the ankle by reducing the restrictive nature of the tape. Simoneau (1997) measured changes in ankle joint neuromuscular control after applying two strips of tape either side of the talocrural joint, one strip positioned posteriorly over the Achilles tendon, and the other placed anteriorly. In a similar study, Matsusaka et al. (2001) also measured ankle joint neuromuscular control after subjects participated in a 10 week proprioceptive training regime wearing two strips of tape designed to eliminate the mechanical effects of taping, yet add cutaneous sensory feedback around the ankle joint. Contrary to Simoneau (1997) the strips of tape in this study were positioned either side of the lateral malleolus along the anterior and posterior margins from the distal third of the lower leg to the sole of the foot. Despite using different measures both studies concluded that the strips of tape provided increased sensory feedback; Simoneau (1997) used joint position sense measures whereas Matsusaka et al. (2001) used postural stability during a single leg stance. Interestingly, in the study by Matsusaka et al. (2001) although the taping technique was worn throughout the 10 week training regime, it was not worn during assessment suggesting that the taping has long term neuromuscular benefits, furthermore, the subject group that had worn the tape during training improved two weeks earlier, than the control group.

The taping techniques by Simoneau (1997) and Matsusaka et al. (2001) were thought to create a pulling effect on the skin during ankle movements, however due to the differences in positioning between the two taping techniques, one would speculate that the technique designed by Simoneau (1997) would pull on the skin more so during plantar flexion and dorsi flexion movements, whereas the technique by Matsusaka et al. (2001) would pull more so during inversion and eversion movements. This remains speculation as Simoneau (1997) only measured joint position sense during plantar flexion and dorsi flexion, and Matsusaka et al. (2001) used rectangular area values without giving changes of postural sway in specific directions. As ankle sprains are predominantly caused by excessive inversion, the taping technique by Matsusaka et al. (2001) seems more relevant when assessing ankle joint neuromuscular control. The taping technique used by Matsusaka et al. (2001) is shown in Appendix Fourteen.

Sawkins et al. (2007) tested the placebo effect of taping, by putting only one strip of tape on the subjects ankle, the strip was approximately 10cm long and positioned on the lateral aspect of the lower leg above the lateral malleolus, aligned vertically over the tendon of the peroneus longus. The subject was informed that this would have a proprioceptive effect to enhance performance when in reality it was thought to have little physiological effect. The study showed that neither the real ankle taping method, nor the placebo taping had an effect on performance; however subject's perceptions of stability, confidence, and reassurance had increased with both techniques compared to the no-tape control. The methodology of this study could be questioned as the altered perceptions could have been influenced by sensory feedback caused by the tape on the skin, rather than the perception that tape will prevent injury as advised by the researchers.

The scientific reasoning behind the neuromuscular effects of taping is unclear; some have suggested it may be caused by enhanced stimulation of the cutaneous receptors caused by the close contact between the tape and the skin (Feuerbach & Grabiner, 1993), whereas others suggest that the tape may increase cutaneous input, which combine together on muscle afferents causing an increase in excitability of the motor neurone pool (Refshauge et al., 2009). A recent study by Alexander, McMullan and Harrison (2008) attempted to establish the scientific basis of the neuromuscular effects of taping; the study involved two applications of tape, one across the direction of the muscle fibres, and the other along the muscle fibres. Tape across muscle fibres failed to affect motor neurone excitability, tape along muscle fibres reduced excitability (19 - 13%) suggesting that changes to movement patterns cannot be explained by facilitation of motor neurone excitability. The results of the study by Alexander et al. (2008) are in contrast to Tobin and Robinson (2000), and Morrissey (2000). Tobin and Robinson (2000) found evidence to suggest that application of tape perpendicular to the muscle fibres can inhibit muscle activity, which may be useful during rehabilitation when specific muscles need to be targeted, whereas Morrissey (2000) found evidence to suggest that application of tape in line with muscle fibres may facilitate muscle activity.

6.2.21 Summary of Athletic Taping Literature

Although the scientific basis is not always clear, it is generally agreed that taping produces a combination of mechanical and neuromuscular effects (Wilkerson, 2002) including changes in range of motion (Nishikawa et al., 2002; Paris et al., 1995), changes in muscle activity (Franettovich et al., 2008) and changes in postural stability (Leanderson et al., 1996). There is a clear need for research in this area to measure the effects of athletic taping on subjects with different foot structures in aspects of neuromuscular control. Before conducting Study Five, pilot testing was required to determine the reliability of the application of lower limb athletic taping.

6.3 Pilot Study Seven: Test Retest Reliability of Tape Application

6.3.1 Abstract

Context: Athletic taping is widely used in sport, however previous literature has highlighted concern over the reliability of the application of taping techniques. It is imperative that this is measured before conducting Study Five in order to establish internal validity.

Objective: To determine intra-tester reliability of the application of the taping techniques to be used in Study Five.

Subjects: Six (3 males 3 females), age = 21 ± 1.6 years, height = 171.2 ± 9.8 cm, mass = 69.9 ± 15.7 kg.

Methods: The arch and ankle taping techniques were applied three times on two seperate days. With arch taping, navicular drop height was measured three times for each tape application. With ankle taping, range of motion was measured using a goniometer three times for each application. Intraclass correlation coefficient and standard error of measurement were then calculated. The proprioceptive taping could not be tested for reliability as it did not cause a mechanical affect across the ankle or foot and therefore could not be measured objectively.

Results: Reliability for both the arch and ankle taping was found to be very high over two days for both taping techniques (ICC > .95).

Conclusions: Due to very high measures of reliability, it was concluded that both taping techniques were suitable for use in Study Five. Although the proprioceptive technique could not be tested for reliability, it was also to be included in Study Five.

6.3.2 Introduction

In order to make the results of the Study Five comparable to other literature, it was decided that the three taping techniques to be used were the modified low-Dye arch taping technique, the closed basket weave ankle taping technique, and a proprioceptive taping technique as used by Matsusaka et al. (2001). The modified low-Dye arch taping technique was chosen on account of previous work by Denyer and Mitchell (2011) where it was found to be most effective in supporting the arch of the foot during dynamic movements in comparison to double X-arch taping and shop bought orthotics (Appendix Thirteen). Following extensive practicing of the tape applications, reliability testing was conducted on arch and ankle taping. Reliability testing was not carried out on the proprioceptive taping technique as it was deemed unnecessary due to the simple nature of the taping technique, additionally, as the proprioceptive technique did not cause a mechanical effect, the reliability of the application could not be measured objectively. As arch taping has been shown to increase navicular drop height on application (Vicenzino et al., 1997; Vicenzino et al. 2000), it was decided that this would be the most suitable measure of arch taping reliability, as it indirectly indicates the tension of the tape applied. It was thought that the application of the taping would be consistent if the navicular drop height was the same after each application. As ankle taping has previously been shown to restrict ankle range of motion in plantar flexion, dorsi flexion, inversion and eversion (Paris et al., 1995), it was decided that this could be used as a measure of ankle tape application reliability. It was thought that if the taping reduced the range of motion at the ankle by the same amount on each application, it would show that the taping technique was applied with the same tension each time.

6.3.3 Methods

Subjects:

Subject characteristics are shown in Table 6.1. The inclusion and exclusion criteria were the same as those to be used in Study Five, as identified in Section 3.2.3.

Table 6.1 Subject Characteristics

Sex	Age (years)	Height (cm)	Mass (kg)
3 Males 3 Females	21 (1.6)	171.2 (9.8)	69.9 (15.7)

Values are mean (SD).

<u>*Procedures:*</u>
The same procedure was repeated three times on two separate days. On each day, arch and ankle taping techniques were to each subject on the dominant foot as defined in Section 3.2.3. Arch taping was applied following a procedure by Vicenzino et al. (1997) and was applied using zinc oxide 2.5 cm tape as shown in Figure 6.3. After the application of the arch taping, navicular drop height was measured using the same procedure as shown to be reliable in Section 3.2. Ankle taping was applied following a procedure by Perrin (2005) using zinc-oxide 4 cm tape as shown in Figure 6.4. After the application of the ankle taping, ankle range of motion was measured using a hand held universal goniometer. Three measurements were taken for each application of tape including total plantar flexion and dorsi flexion, and total inversion and eversion. The plantar flexion and dorsi flexion method followed a seated protocol by Venturini, André, Aguilar and Giacomelli (2006) who showed that measurements were reliable (ICC > .70) over different testers and different days. A similar procedure by Menadue, Raymond, Kilbreath, Refshauge and Adams (2006) was used for the inversion and eversion measurements; again this technique was also shown to have strong intra-tester reliability over different days (ICC > .80).

Figure 6.3. Modified low-Dye arch taping

Low-Dye Arch Taping Technique - Procedure used by Vicenzino et al. (1997).

- With the foot actively held in dorsi flexion and slight supination, a horizontal horseshoe was placed from the most medial aspect of the head of the first metatarsal, directed posteriorly, and then attached onto the most lateral aspect of the head of the fifth metatarsal.

- A series of mini-stirrups were then placed from the lateral aspect of the horseshoe, running underneath the plantar surface attaching to the medial aspect of the horseshoe, covering the entire midfoot.
- An anchor was then placed around the base of the calf, and three reverse sixes were applied starting at the medial malleolus coursing laterally across the dorsum of the foot, beneath the midfoot and continuing up to the anchor strip.
- Each reverse six was overlapped anteriorly by approximately half the width of tape.
- Finally, two calcaneal slings were applied; each starting from the anterior aspect of the anchor strip, passing distally towards the medial malleolus, passing underneath the foot and continuing back up to its origin on the anchor strip, again the second strip was overlapped anteriorly by approximately half the width of tape.

Figure 6.4. Closed basket weave ankle taping

Closed Basket Weave Taping Technique - Procedure used by Perrin (2005).

- With the subject maintaining 90° of dorsi flexion, the taping technique consists of two anchor strips, the first on the distal leg around the base of the calf (approximately one third up the length of the lower leg), the second around the foot slightly posterior to the head of the fifth metatarsal.
- Following the anchor strips, a stirrup was then placed from the medial aspect of the most proximal anchor, underneath the foot, and attached on to the lateral aspect of the same anchor.
- A horizontal horseshoe was then placed from the medial aspect of the distal anchor strip, directed posteriorly around the heel, and then attached to the lateral aspect of the same anchor.

- Two further stirrups and horseshoes were applied in a weaving manner creating the basket weave effect.
- More horseshoes were applied up to the proximal anchor strip, enclosing the entire ankle.
- Medial and Lateral heel locks were then applied, and the taping technique was completed by a figure of eight.

Data Reduction and Statistical Analysis:

The average of the three measurements from each tape application taken on both days was used for analysis. Data was initially tested for normality (Shapiro-Wilk $P > .05$), as data were normally distributed it was then tested for significant differences using a paired samples t-test with the a priori alpha level set at .05. No significant differences were found so both relative and absolute reliability were then calculated using an ICC and SEM respectively. All statistical analyses were performed using Excel (Microsoft Office, 2010).

6.3.4 Results

Results of the arch and ankle taping application are shown in Tables 6.2. Each measure resulted in high reliability over two separate days.

Table 6.2 Test Retest Reliability of Arch and Ankle Taping Application

	Test	Retest	*P*	ICC	SEM
Arch Tape Navicular Drop Height (mm)	5.0 (2.8)	5.0 (2.6)	.90	.99	.15
Ankle Tape Plantar Flexion - Dorsi Flexion ROM (°)	57.9 (3.3)	58.3 (3.8)	.44	.96	.67
Ankle Tape Inversion - Eversion ROM (°)	36.1 (4.0)	36.8 (3.6)	.10	.98	.52

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement; ROM: range of motion. ICC > .70 indicating high correlation highlighted.

6.3.5 Discussion

The results indicate that both arch taping and ankle taping application are highly reliable (Munro, 2000) when applied over different days by the same tester (ICC > .95). The SEM are also low for each measure indicating good absolute reliability. Therefore it can be concluded that these taping methods are suitable for use in the Study Five. Although the proprioceptive taping technique could not be measured for reliability, it will still be included in Study Five following the same method as used by Matsusaka et al. (2001).

6.4 Study Five: The Effects of Athletic Taping on Lower Limb Biomechanics

6.4.1 Abstract

Context: In current literature, the effects of athletic taping are unclear. Additionally, few studies have measured the effects of taping on subjects with different foot structures.

Objective: To measure the effects of three athletic taping techniques on subjects with neutral, pronated and supinated foot structures on measures of neuromuscular control.

Subjects: 30 volunteers with symmetrical feet were categorised into three groups which were dependant on navicular drop height measures. 10 had neutral feet (navicular drop height 5 – 9 mm), 10 had pronated feet (navicular drop height > 10 mm), and 10 had supinated feet (navicular drop height < 4 mm).

Methods: Three athletic taping techniques (arch, ankle, proprioceptive) were applied to subjects with different foot structures and with each technique two aspects of neuromuscular control were measured; dynamic postural stability and muscle reaction time to a tilt platform perturbation.

Results: Ankle taping slowed peroneus longus reaction time in pronated (P = .01) and neutral groups (P = .04). In addition, both ankle and arch taping affected postural stability during dynamic hopping tasks. The proprioceptive taping technique had no effect on either measure of neuromuscular control.

Conclusions: Ankle and arch taping appear to cause a different neuromuscular response dependant on the foot structure to which it is applied. In light of this, sports clinicians should consider the foot structure of athletes before applying tape.

6.4.2 Introduction

Athletic taping is widely used to help support structures of the foot and ankle to reduce the risk of injuries associated with lower limb abnormalities including abnormalities of the medial longitudinal arch (Perrin, 2005; Vicenzino et al., 1997). Alongside the restrictive mechanical effects of athletic taping, it has been suggested that taping may also result in neuromuscular effects including changes in muscle activity and changes in postural stability (Karlsson & Andreasson, 1992; Shima et al., 2005). Some research has been carried out on the neuromuscular effects of ankle taping, but arch taping is less researched and previous studies on both taping techniques are inconclusive.

Whilst there is limited research on the neuromuscular effects of lower limb taping techniques, changes in muscle activity as a result of taping have been identified elsewhere in the body including the quadriceps (Macgregor et al., 2005) and the shoulder (Morrissey, 2000) indicating that lower limb taping such as arch and ankle taping may also influence lower limb neuromuscular control. Further evidence for neuromuscular effects of taping were shown by Matsusaka et al. (2001) and Simoneau (1997) who applied strips of tape across the ankle joint without restricting range of motion and measured postural stability and joint position sense; the results indicated that the tape provided increased sensory feedback enhancing neuromuscular control at the ankle.

The purpose of this study was to measure the effects of athletic taping on neuromuscular control on individuals with neutral, pronated and supinated foot structures. It was hypothesised that athletic taping would significantly affect both postural stability and muscle reaction time measures on all subject groups.

6.4.3 Methods

<u>Subjects:</u>

Thirty volunteers participated in this study and were categorised into three groups which were dependant on navicular drop height measures using the same method as shown to be reliable in Section 3.2. Prior to testing, ethical approval was granted from the institutional ethics committee, and all subjects read a subject briefing document (Appendix One), provided written informed consent (Appendix Two) and completed a health screen form (Appendix Three). Subject characteristics are shown in Table 6.3. The inclusion and exclusion criteria were the same as those identified in Section 3.2.3.

Table 6.3 Subject Characteristics

Group	Sex	Age (years)	Height (cm)	Mass (kg)	Navicular Drop Height (mm)
Neutral	4 Males 6 Females	20 (2.1)	169.3 (8.8)	70.0 (11.2)	7.0 (1.0)
Pronated	4 Males 6 Females	21 (1.5)	169.8 (10.5)	70.4 (14.6)	11.0 (2.0)
Supinated	4 Males 6 Females	20 (1.6)	169.3 (6.2)	69.5 (10.7)	3.0 (1.0)

Values are mean (SD).

Procedures:

Postural stability and muscle reaction time measures were repeated with each testing condition (arch tape, ankle tape and proprioceptive tape). Subjects rested for ten minutes between each condition to avoid a fatigue effect. Postural stability measures were collected using the same method as identified in Section 3.4.3 however only the three hopping tasks were included. The same muscle reaction time method as identified in Section 4.7.3 was also used in this study; electrodes were not moved throughout the duration of the one hour testing period.

The arch taping technique used in this study was the modified low-Dye taping technique using a procedure described by Vicenzino et al. (1997) this technique used zinc oxide 2.5 cm tape. The ankle taping technique used in this study was the closed basket weave procedure as described by Perrin (2005) using zinc-oxide 4 cm tape. Both the arch and ankle taping techniques were applied in the same procedure as shown to be reliable in Pilot Study Seven. The proprioceptive taping followed a procedure by Matsusaka et al. (2001) where two strips of zinc-oxide 2.5 cm tape were applied parallel to each other either side of the lateral malleolus, starting on the distal third of the lower leg, running distally to the plantar surface of the foot; this is shown in Appendix Fourteen. This technique was designed to provide cutaneous stimulation during even the slightest inversion and eversion movements, whilst eliminating any mechanical effect of taping as there was no restriction to range of motion at the ankle joint.

Data Reduction and Statistical Analysis:

Recording for postural stability data began as soon as the pressure on the force platform exceeded 50 N, and data were collected at 1000 HZ. Centre of pressure data were collected, and six variables were calculated at two time frames (200 ms and three seconds) including peak COP, mean COP, total COP, total distance, average velocity, and 95% ellipse area using an Excel template (Microsoft Office, 2010) (Appendix Four). This procedure was shown to be reliable in Pilot Study Two. Definitions of each COP variable are shown in Appendix Five.

Muscle reaction time data was processed with an RMS filter using a 10 ms moving window. Data was then reduced using an Excel template (Microsoft Office, 2010); this is shown in Appendix Ten. As shown to be reliable in Section 4.5, muscle reaction time was defined as the time between the onset of the tilt mechanism, to the onset of the EMG signal when it reached a level of three standard deviations above the baseline for 25 ms consecutively. The baseline value was the average value recorded over 150 ms immediately prior to the onset of the tilt mechanism.

As data were normally distributed (Shapiro-Wilk $P > .05$), for each dependant variable (COP parameters, peroneus longus, tibialis anterior and gluteus medius reaction time), a separate repeated measures analysis of variance was performed with taping technique (no-tape, ankle, arch and proprioceptive) as the independent variable. The a priori alpha level was set at .05. Where Mauchly's test of sphericity was significant, a Greenhouse-Geisser correction was applied. Where a significant difference was observed, pre-planned paired samples t-tests were used to identify the source of significance by comparing arch, ankle and proprioceptive data against the no-tape control. Bonferroni adjustment for pairwise comparisons was applied where appropriate ($P = .05 \div 3 = .016$). Effect size (η_p^2 values) was calculated where 0.01 - 0.06 indicated a small effect, 0.06 - 0.14 indicated a medium effect, and > 0.14 indicated a large effect. Observed power was also calculated. All statistical analyses were performed using SPSS (version 17.0; SPSS Inc, Chicago, IL).

6.4.4 Results

Postural Stability results are shown in Appendix Fifteen. Muscle reaction time results are shown in Appendix Sixteen.

Analysis of the postural stability tasks found significant differences only at the 200 ms time frame. During the forward hop task, a significant difference between taping techniques was identified in the pronated group; $F(3, 30) = 5.6$, $P = .01$, $\eta_p^2 = 0.36$, observed power = 0.92. As shown in Figure 6.5 pre-planned follow up tests indicated that both ankle and arch taping resulted in significantly lower mean anterior COP (indicating increased postural stability) in comparison to the no-tape control ($P = .004$, $P = .001$ respectively). During the same measure, differences were also identified within the supinated group; $F(3, 21) = 4.0$, $P = .02$, $\eta_p^2 = 0.36$, observed power = 0.76. Follow up tests indicated that ankle taping resulted in significantly lower mean anterior COP in comparison to the no-tape control ($P = .001$); this is also shown in Figure 6.5. During analysis of average velocity, differences were also identified within the supinated group $F(3, 21) = 7.1$, $P = .01$, $\eta_p^2 = 0.50$, observed power = 0.95. Further tests indicated that ankle taping resulted in significantly higher average velocity in comparison to the no-tape control ($P = .003$) (Figure 6.6).

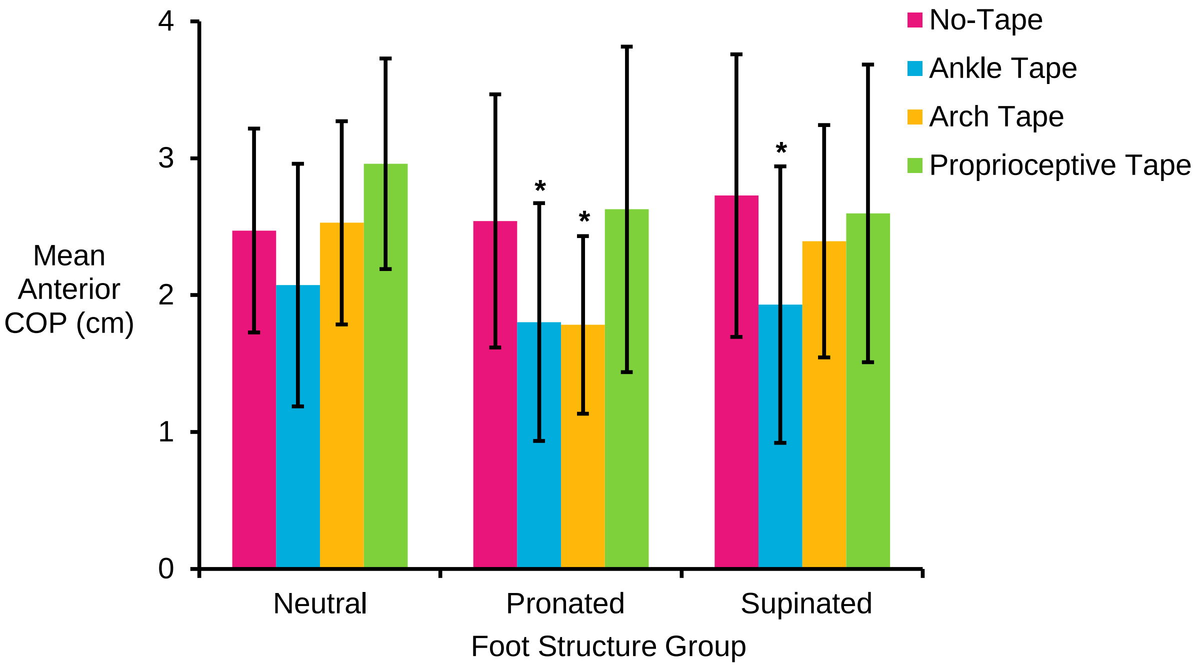


Figure 6.5 Mean anterior COP during the forward hop at 200 ms with lower limb athletic taping on neutral, pronated and supinated foot structures; showing standard deviation.

* Indicates significant difference in comparison to the no-tape condition ($P < .016$).

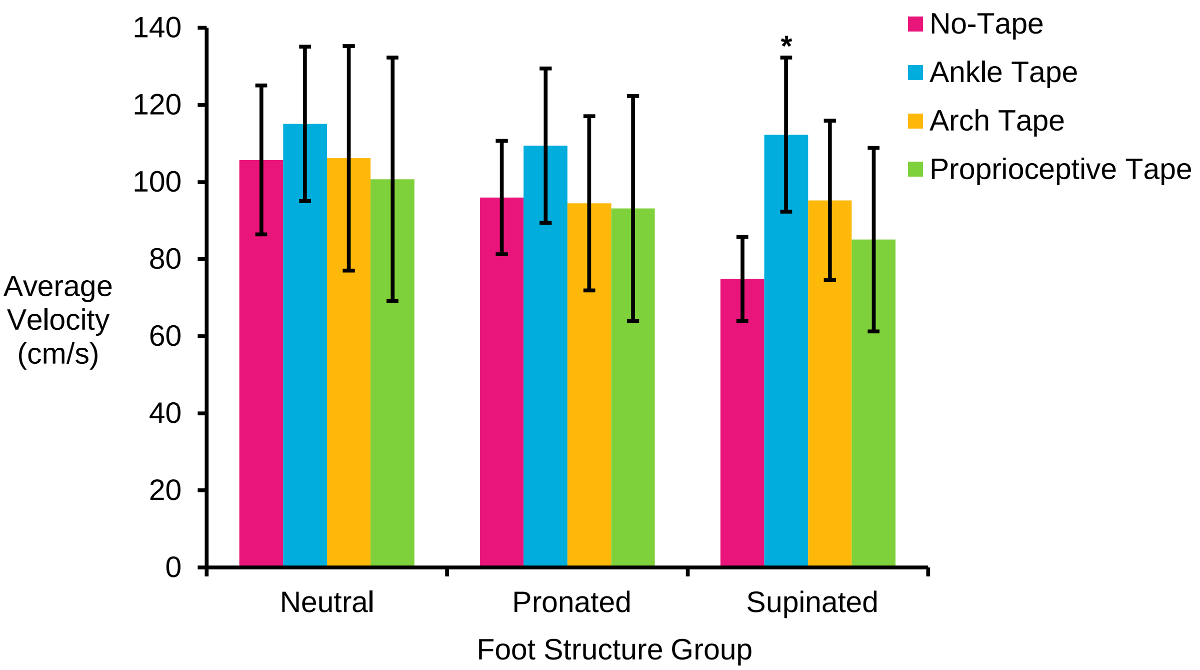


Figure 6.6 Average velocity during the forward hop at 200 ms with lower limb athletic taping on neutral, pronated and supinated foot structures; showing standard deviation. ***** Indicates significant difference in comparison to the no-tape condition ($P < .016$).

During the diagonal hop task, the only difference identified was in the 95% ellipse area within supinated group; $F(3, 21) = 3.2$, $P = .04$, $\eta_p^2 = 0.31$, observed power = 0.65. Follow up tests indicated that arch taping resulted in a lower 95% ellipse area than the no-tape control (*P* = .005) (Figure 6.7).

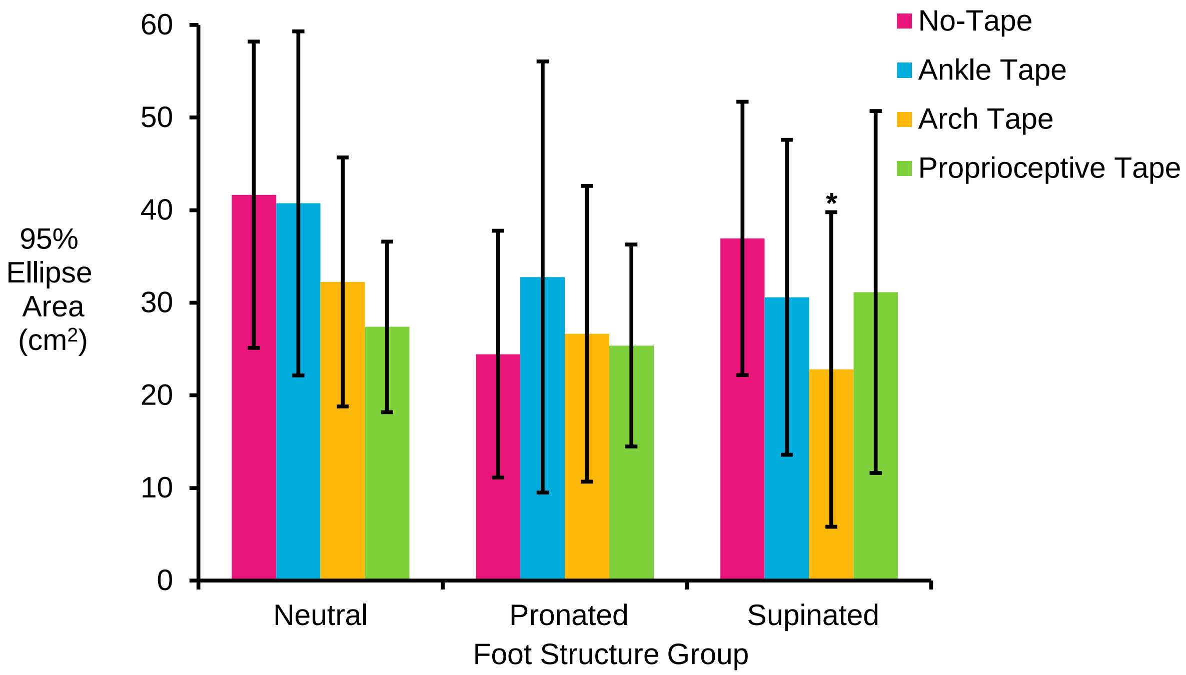


Figure 6.7 95% ellipse area during the diagonal hop at 200 ms with lower limb athletic taping on neutral, pronated and supinated foot structures; showing standard deviation. ***** Indicates significant difference in comparison to the no-tape condition (*P* < .016).

During the lateral hop task, the only difference identified was in the supinated group; $F(3, 21) = 4.2$, $P = .01$, $\eta_p^2 = 0.37$, observed power = 0.78. As shown in Figure 6.8, further analysis revealed that the ankle taping resulted in significantly lower peak medial COP in comparison to the no-tape control (*P* = .010). No differences were found within the neutral or pronated groups in the diagonal or lateral hop tasks.

Muscle reaction time analysis of the neutral group revealed a significant difference between taping techniques for the peroneus longus muscle; $F(3, 27) = 3.7$, $P = .02$, $\eta_p^2 = 0.29$, observed power = 0.74. Pre-planned follow up tests indicated that ankle taping resulted in significantly slower peroneus longus reaction time in comparison to the no-tape control (*P* = .008). No significant differences were identified for the tibialis anterior or gluteus medius muscles. Analysis of the pronated group indicated a significant difference between the taping techniques for the peroneus longus muscle; $F(3, 27) = 3.1$, $P = .04$, $\eta_p^2 = 0.26$, observed power = 0.65. Further analysis indicated that ankle taping resulted in significantly slower peroneus longus reaction time in comparison to the no-tape control (*P* = .002). The differences identified for the peroneus longus muscle are shown in Figure 6.9. All other reaction time data are shown in Appendix Sixteen. No differences were identified for the gluteus medius muscle. No significant differences were identified within the supinated group.

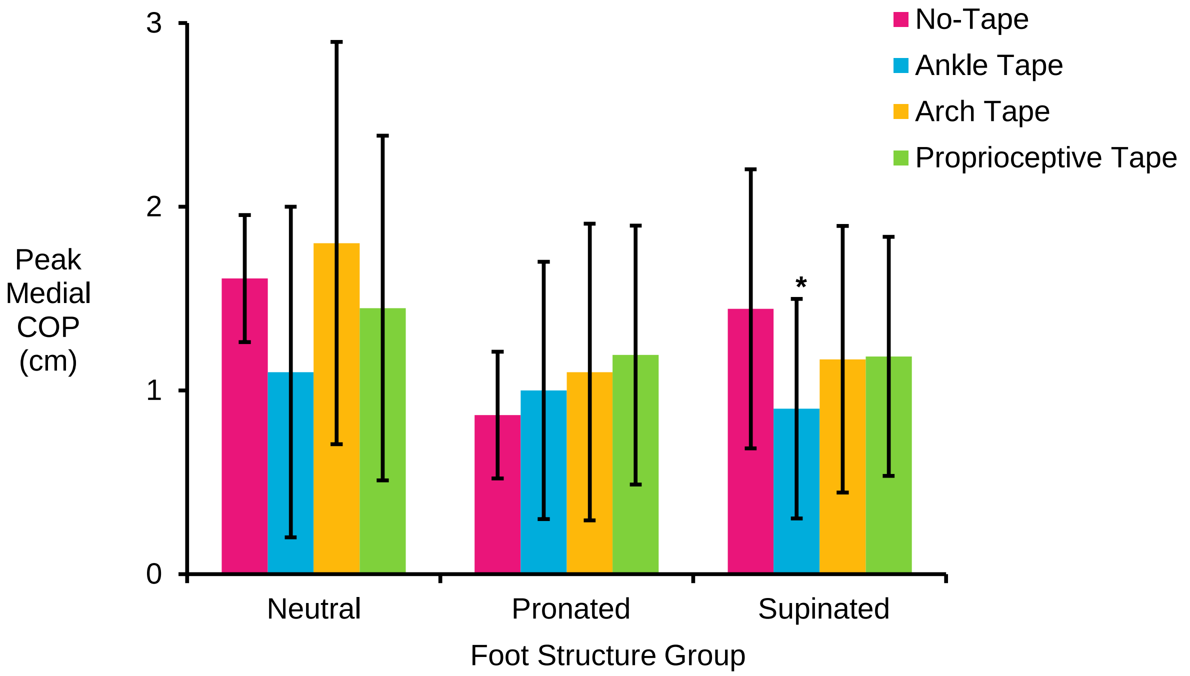


Figure 6.8 Peak medial COP during the lateral hop at 200 ms with lower limb athletic taping on neutral, pronated and supinated foot structures; showing standard deviation. ***** Indicates significant difference in comparison to the no-tape condition (*P* < .016).

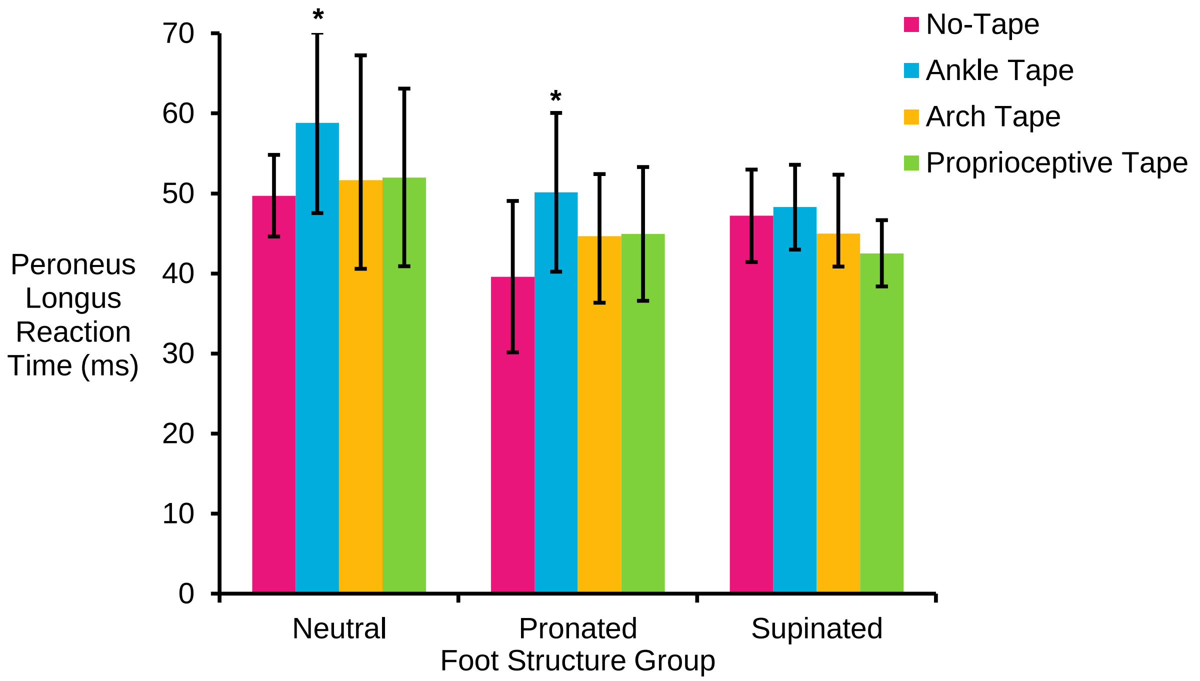


Figure 6.9 Reaction time of the peroneus longus muscle with lower limb athletic taping on neutral, pronated and supinate foot structures; showing standard deviation. ***** Indicates significant difference in comparison to the no-tape condition (*P* < .016).

6.4.5 Discussion

This is the first study to have measured the neuromuscular effects of three different lower limb taping techniques on subjects with neutral, pronated and supinated foot structures. The results of this study indicate that aspects of neuromuscular control as measured by postural stability and muscular reaction time are affected by lower limb athletic taping. Therefore the hypothesis was partially accepted as athletic taping did significantly affect both postural stability and muscle reaction time measures across the three subject groups.

The muscular reaction time data indicated that ankle taping slows peroneus longus reaction time in pronated and neutral groups in comparison to the no-tape control. Several previous studies have analysed the effects of ankle taping on muscle reaction time, the results of this study are in agreement with Shima et al. (2005) who also found that ankle taping slows peroneal reaction time on both healthy subjects, and subjects with a history of ankle injury. Shima et al. (2005) found reaction times of 83.8 ms with ankle taping, which was shown to be significantly slower than the no-tape condition (80.8 ms). In contrast, Karlsson and Andreasson (1992) found that peroneal reaction time was significantly faster with the use of ankle taping on subjects with ankle instability from 84.5 ms to 73.4 ms with taping. Interestingly in the study by Karlsson and Andreasson (1992) no significant differences were present with taping among 6 of the 20 subjects used in the study with the lowest degree of ankle instability, and the taping did not affect reaction time on stable feet. In the current study, no significant difference was found for the peroneus longus within the supinated group with ankle taping. The arch and proprioceptive taping techniques had no effect on muscular reaction time.

The postural stability data indicated that both ankle and arch taping appear to affect individuals with pronated and supinated foot structures however no differences were apparent in the neutral group. Differences within the pronated group were only identified during the forward hop task, whereas the supinated group highlighted differences across all three tasks. Differences in postural stability were only identified in this study at the 200 ms time frame, this is in line with Study One, and confirms the need for future postural stability research to analyse this time frame as it is within the first 200 ms that injuries such as ankle sprains occur (Fong et al., 2009).

With ankle taping, during the forward hop both the pronated and supinated groups resulted in lower mean anterior COP in comparison to no-tape control at 200 ms, however this difference was not apparent in the neutral group. Interestingly, despite resulting in lower mean anterior COP, during the same task the supinated group also resulted in increased average velocity in comparison to the no-tape control. Whilst it was not objectively measured, it was observed that with ankle taping, individuals with supinated feet appeared to have greater gross body movements on landing in comparison to the neutral and pronated groups. One explanation for this may be that subjects with supinated foot structures are more dependent on hip strategy, rather than ankle strategy in order to maintain stability when wearing ankle taping. Ankle strategy involves shifting the centre of body mass by rotating the body about the ankle joint (Horak, 1987) whereas the hip strategy involves the use of the gluteus medius to correct posture (Leavey et al., 2010). This speculation is supported by the fact that those with supinated foot structures typically have rigid and inflexible feet (Franco, 1987). The use of ankle taping would add to the rigidity of the foot and ankle increasing the need to correct posture using hip strategy. In comparison, those with neutral feet naturally have more subtalar joint mobility (Franco, 1987), and may be able to correct posture more efficiently with ankle strategy. This remains speculation as gluteus medius muscle activity was not recorded during the postural stability tasks.

During the forward hop, arch taping resulted in reduced mean anterior COP in comparison to the no-tape control in the pronated group for the forward hop task. Arch taping also affected the supinated group during the diagonal hop, resulting in a lower 95% ellipse area in comparison to the no-tape control. Both results indicate that arch taping appears to improve postural stability in supinated group and pronated groups. The arch taping had no affect during the lateral hop task suggesting that the technique does not provide enough support during lateral movement. Wall et al. (2006) measured the effects of arch taping on subjects with pronated feet and found that arch taping had positive effects in dissipating forces during landing on dynamic tasks. No other studies were found to have considered foot structure when measuring the effects of arch taping on postural stability.

The mechanism causing the ankle taping to affect lower limb neuromuscular control is still unclear. It has been reported that the restrictive nature of ankle taping may slow the speed of inversion at the ankle which in turn would slow the muscle reaction time (Karlsson & Andreasson, 1992). However one would expect this to occur equally across all foot structures, whereas in this study, only reaction times of the pronated and neutral groups were affected by ankle taping. Similarly within the measures of postural stability, differences were only identified in the pronated and supinated groups, and not in the neutral group making it difficult to conclude that the differences in reaction time are caused solely by the mechanical restriction of the tape, as one would expect this difference to be apparent among all foot structures.

Both Matsusaka et al. (2001) and Simoneau (1997) have suggested that the pull of the tape on the skin provides cutaneous cues of plantar surface position and orientation, yet differences were not apparent in either the postural stability or muscle reaction time measures with the proprioceptive taping. This contradicts result of Matsusaka et al. (2001) who found that the same proprioceptive taping technique improved postural stability as measured during a single leg stance. Perhaps during more dynamic tasks the proprioceptive taping condition does not result in enough cutaneous cues in comparison to the ankle and arch taping techniques due to having a much smaller surface area of direct skin contact.

6.4.6 Limitations

O'Sullivan et al. (2008) suggested that in studies where no neuromuscular effects are found, one explanation may be that participants may require time to adjust to the biomechanical and neuromuscular changes caused by arch taping techniques; therefore effects may not be present when a subject is tested immediately after tape application. In this study, this is a potential limitation as subjects were tested immediately after the application of tape, and so O'Sullivan's explanation may account for why some measures did not appear to be affected by the taping techniques.

A further limitation in testing subjects immediately after the application of taping is that athletic taping has been shown to loosen during exercise (Ricard, Sherwood et al., 2000) which reduces the mechanical restriction of tape across the ankle (Robbins et al., 1995). Therefore the results of this study only apply immediately after tape application and cannot be generalised to during exercise. Through exercise, previous studies have shown that neuromuscular effects of taping may outlast mechanical effects of taping, (Robbins et al., 1995) however as the mechanical effects of taping were not reduced in this study as they would be through exercise, it is difficult to conclude that the results of this study represent the full neuromuscular effects of taping. Repeating the same testing procedure after a period of exercise which reduces the mechanical restriction of the tape may help to clarify whether results were caused by a neuromuscular effect of the taping, a mechanical effect, or by a combination of the two.

6.4.7 Recommendations for Future Research

This area should be researched further in order to determine if changes in reaction time and postural stability caused by lower limb taping techniques are apparent after a period of exercise where the mechanical effects of taping have been reduced. Additionally, whilst this study used a static tilt platform, further research among subjects with different foot structures may utilise a dynamic tilt platform such as that used by Hopkins et al. (2007) to observe perturbed walking, or such as that used by Gutierrez and Kaminski (2010) to examine the effects of a perturbation during a dynamic landing task. Further research into the effects of foot structure on postural stability should continue to analyse data at 200 ms as the current study suggests that this time frame may provide more useful information in terms of injury prevention in comparison to longer time frames used by other studies.

6.4.8 Conclusions

The results of this study suggest that both arch and ankle taping affect neuromuscular control on subjects with neutral, pronated and supinated foot structures. In light of this study, sports clinicians should consider the foot structure of the athlete before administering athletic taping. Further research should aim to determine whether the differences in postural stability and reaction times identified in this study are also apparent after a period of exercise where the mechanical restriction of the taping has been reduced.

6.5 Development of Research

The work in this chapter has indicated that the application of both arch and ankle taping are reliable. Additionally, Study Five has indicated that arch and ankle taping techniques affect neuromuscular control as measured by postural stability and muscle reaction time. A limitation of this study is that the effects of taping were measured immediately after application of tape. It could be argued that in order to be more applicable to sport and to improve external validity, the effects of taping should be measured after a period of exercise.

The final study in this thesis will analyse the effects of athletic taping after exercise, to determine whether the results found in Study Five are also applicable after a period of dynamic exercise.

Chapter Seven

The Effects of Athletic Taping on Lower Limb Biomechanics after Exercise

7.1 Introduction to Chapter

This chapter addresses the effects of athletic taping on lower limb biomechanics both before and after a period of exercise. The purpose of this is to increase external validity in comparison to Study Five by making the results of this thesis more applicable to sport. The second purpose of including the exercise protocol is to attempt to reduce the mechanical effects of taping in order to expose potential neuromuscular effects which may occur as a result of the interaction of the tape on the skin. This chapter includes a literature review in order to gain to further understanding of the effects of foot structure and athletic taping on lower limb biomechanics after exercise. Following this, pilot testing was conducted to determine the most effective exercise protocol to be used in Study Six. Study Six is the final study in this thesis and combines methodology of Studies One to Five in order to establish the effects of athletic taping on lower limb biomechanics in subjects with neutral, pronated and supinated foot structures both before and after exercise.

7.2 Review of Literature on Neuromuscular Control, Taping and Exercise

A number of search engines were used to acquire the relevant information including Scopus, Web of Knowledge and Google Scholar. Typical search terms included *exercise and postural stability, exercise and reaction time, exercise and taping,* and *exercise protocols.*

7.2.1 Exercise and Athletic Taping

Holmes et al. (2002) tested modified low-Dye arch tape before and after a ten minute walking protocol and found that whilst navicular drop height had reduced, the taping was still effective in maintaining the subtalar joint neutral position; however Holmes et al. (2002) admit that this may not occur through more vigorous or longer periods of exercise. Vicenzino et al. (2000) measured navicular drop height after application of the modified low-Dye taping technique and found it resulted in a 19% increase in arch height in comparison to the no-tape control. Whilst this increase reduced in size to 3.5% after 10 minutes of jogging around a figure of 8 track 30 meters in length Vicenzino et al. (2000) concluded that it was still superior to the no-tape control. These results suggested that it was still a functional modality after exercise which may be useful in the treatment of lower limb injuries that are associated with pronation (Vicenzino et al., 2000). In contrast, Ator et al. (1991) found that the modified low-Dye taping technique had no effect on navicular drop height after a ten minute jogging protocol.

Ankle taping has also been shown to reduce in effectiveness after exercise (Martin & Harter, 1993; Pederson et al., 1997; Paris et al., 1995). To overcome this many studies ensure that measurements are taken immediately after the application of tape (Shima et al., 2005), however this limits the external validity of the study as results cannot be directly applied to sport. The effects of ankle taping post-exercise have been measured using range of motion, postural stability, and muscle reaction time. One of the first studies to measure the effects of taping post-exercise was Rarick et al. (1962), who found that taping reduced in effectiveness by up to 40% in just 10 minutes of exercise as measured by ankle joint range of motion; however it is likely that the strength and adhesive properties of athletic taping has developed in the past 50 years. Consistent with Rarick et al. (1962), Paris et al. (1995) also found that ankle taping reduced in effectiveness during just 15 minutes of activity on a treadmill showing significant increases in plantar flexion and inversion range of motion throughout 15 minute intervals. Martin and Harter (1993) also found that after 20 minutes of exercise with ankle taping, rearfoot inversion range of motion increased. The ankle taping technique was least successful in restricting ankle movement in comparison to a lace-up brace and a semi-rigid orthotic (Martin & Harter, 1993). More recently, Fleet, Galen and Moore (2009) applied ankle taping and measured strength of tape over a 24 hour period during activities of daily living, this study found that over 24 hours, inversion range of motion increased, and the moment required to rotate the foot into inversion decreased. This showed that the taping had reduced in effectiveness over the 24 hour period.

In contrast to the previously mentioned studies, after 30 minutes of basketball, Robbins et al. (1995) found that whilst the mechanical properties of ankle taping had reduced, the taping still had a positive effect on joint position sense in comparison to the no-tape control, giving evidence for neuromuscular effects of taping caused by the interaction between the tape on the skin rather than mechanical restriction. Consistent with this, Pederson et al. (1997) found that ankle taping reduced inversion range of motion at the ankle by 35% before, and 20% after exercise, indicating that the taping technique was still effective in reducing the amount of inversion after 30 minutes of rugby drills.

Using measures of postural stability, Leanderson et al. (1996) found that ankle taping resulted in decreased parameters after application, yet after exercise there were no differences between ankle taping and the no-tape control. Leanderson et al. (1996) also found that subjects without taped ankles showed a decrease in postural stability parameters after exercise, suggesting that less uncontrollable movements occur after a period of exercise. Because of this, Leanderson et al. (1996) concluded that ankle taping may be most important during the initial stages of exercise, as at this point the athlete may be most susceptible to injury caused by poorer postural stability. More recently, Lohkamp, Craven, Walker-Johnson and Greig (2009) measured the effects of ankle taping before and after a 45 minute treadmill protocol simulating the change of speed in football match play. It was identified that reaction times to an inversion perturbation on a NeuroCom Balance System (NeuroCom International Inc, Clackmas, OR) were longer after 22.5 minutes of exercise. In the same study, no significant differences were identified in measures of centre of gravity displacement. Riemann et al. (2002) measured the effects of ground reaction forces on ankle taping, and found that during dynamic movements, the tape resulted in a decrease in the time to reach peak impact forces. This indicated that the foot is subjected to peak loads within a shorter time period which may have negative implications in terms of injury. However this requires further research.

Using a combination of treadmill running and mat based strength exercises, Alt et al. (1999) found no difference between pre and post-exercise measures on muscle reaction time with ankle taping. However they did find reduced angle of inversion, and reduced inversion amplitude post-exercise indicating that the mechanical effects of the taping restrict the amount of inversion occurring at the ankle during an inversion perturbation. Lohrer et al. (1999) also measured the effects of ankle taping on muscle reaction time to a tilt platform perturbation and found that after 20 minutes of exercise, the proprioceptive amplification ratio (a calculation based on inversion time and amplitude) had significantly reduced. It was thought that this reduction was caused by both fatigue and mechanical loosening of the tape. Ricard, Sherwood et al. (2000) also found reduced average inversion velocity, maximum inversion velocity, and time to reach maximum inversion to a tilt platform perturbation with the use of ankle taping. These differences were still present after exercise implying that the neuromuscular system may have additional time to respond to an inversion perturbation with tape in comparison to without tape.

A limitation to measuring range of motion post-exercise is that several previous studies have found that without tape, aspects of lower limb biomechanics are affected by exercise. Range of motion increased post-exercise due to increased extensibility of connective tissue caused by a warming effect of increased blood flow (Ricard, Sherwood et al., 2000). In addition to affecting range of motion, fatigued muscles are less efficient at maintaining postural stability; Gribble and Hertel (2004) used this principle to examine the interaction between the hip and ankle during single limb stance on healthy adults by fatiguing different muscle groups. The results of their study showed that fatigue to more proximal muscle groups resulted in a greater reduction of postural control than fatigue to distal muscle groups, highlighting the import role of hip muscles in maintaining postural control. They also found that ankle invertors and evertors were not as important as plantar flexors and dorsi flexors in maintaining single limb stance as the inversion and eversion muscle groups were not affected by the fatiguing task. This study did not involve the use of a tilt platform and so it remains to be seen how fatigue affects the peroneus longus, tibialis anterior and gluteus medius during perturbed stance.

Robbins et al. (1995) measured joint position sense on subjects without the use of tape both before and after exercise, and found that position awareness of the foot declines with exercise, reinforcing the need for preventative measures such as athletic taping. Benesch et al. (2000) measured the effects of a warm up (cycle ergometer followed by stretching) on muscular reaction time to a tilt platform perturbation and found that the reaction times of the peroneus brevis were significantly reduced; however no differences were identified for the peroneus longus. In the same study, following two minutes of rope skipping, the reaction time of the peroneus brevis and longus significantly increased. This study implies that the type of exercise may have a varied effect on muscle activity.

As range of motion, postural stability and muscle reaction time are known to be affected by exercise, when differences are found with ankle or arch taping post-exercise, it can be unclear whether results are due to loosening of the tape, increase in the mobility caused by exercise, or a combination of the two. Therefore exercise protocols need to be monitored in order to maximise internal validity and ensure that any differences identified between taping conditions can be attributed to the tape, rather than due to fatigue caused by the exercise protocol.

7.2.2 Exercise Protocols Used in Current Literature

A common protocol in taping studies is for subjects to undergo a period of exercise after tape application before taking various biomechanical measurements. The purpose of this is to determine how long the mechanical effects of taping last during exercise (Ator, et al., 1991; Del Rossi et al., 2004; Paris et al., 1995), or to try and reduce the mechanical effects of the tape in order to expose potential neuromuscular effects (Leanderson et al., 1996; Robbins et al., 1995). A thorough review of the literature indicated four main methods by which researchers have attempted this.

The first of which is to use a sport specific protocol such as ‘2.5 – 3 hours of football practice’ (Fumich, Ellison, Guerin & Grace, 1981), ‘a volleyball session’ (Greene & Hillman, 1990), ‘a two hour football practice session’ (Leanderson et al., 1996), or ‘30 minutes of running and basketball' (Robbins et al., 1995). Whilst this method is applicable to sport and the typical conditions in which taping is used, the vague detail in this method can be highly criticised for being unrepeatable. No two exercise sessions will be the same and so there is no way to quantify whether each subject is stressing the taping in the same way. There are various movement analysis studies in support of this from a range of sports. For example, Bloomfield et al. (2007) found that defenders in FA Premier League soccer performed the highest amount of jogging, skipping and shuffling movements and spent a significantly less amount of time sprinting and running in comparison to midfielders and strikers implying that the amount of stress place on athletic taping on athletes of different positions within the same sport will vary enormously.

The second method found to be common among taping studies is to use an agility based exercise session. There are various agility tests that have been used in current literature; these often involve a combination of running and cutting drills (Delahunt et al., 2009; Frankeny et al., 1993; Manfroy, Ashton-Miller & Wojtys, 1997; Martin & Harter, 1993; Meana, Alegre, Elvira & Aguado, 2008; Pederson, et al., 1997) or 'figure of 8' courses (Hadley et al., 1999; Laughman, Carr, Chao, Youdas & Sim, 1980; Vicenzino et al., 2000). Agility protocols are typically more controlled than sport specific protocols as researchers usually set a specified distance or time limit. They can still be criticised however, as there is no way to regulate whether subjects place the same amount of stress on the tape each time they complete the protocol. Vicenzino et al. (2000), attempted to control this by ensuring each subject completed the same amount of circuits on an agility track over two separate days, however there is no control for between subject differences such as speed or stride length which make results questionable.

The third most prevalent method of stressing athletic taping is to use a marked running course (Ator et al., 1991; Holmes et al., 2003) or treadmill based protocol (Del Rossi et al., 2004; Lohkamp et al., 2009; Riemann et al., 2002; Paris et al., 1995; Yi et al., 2003). These protocols have included 'ten minutes of normal walking pace around a ¼ mile track' (Holmes et al., 2002), however Yi et al. (2003) suggested that walking isn't strenuous enough to loosen tape. A further criticism of the use of a running track is that the subjects presumably only walk in only one direction, meaning that as they walk around the track, they only turn to the left therefore placing more stress on the lateral side of the left foot, and the medial side of the right foot. It could be argued that this is sport specific for track events, however not for other dynamic sports. Rather than using a running track, Ator et al. (1991) instructed subjects to 'jog continuously for ten minutes in a 50 yard indoor hallway', however this can again be criticised as there was no control for speed or distance covered within the ten minutes.

Several researchers have used treadmills in order to elicit more control over stressing the tape, as the speed and distance can be monitored. However despite this, of the studies found, several can be criticised for having poor between subject reliability, for example Del Rossi et al. (2004) asked subjects to run at a 'self-selected running speed', and whilst this was controlled for within subject reliability as subjects ran at the same speed over two days, the self-selected pace of one athlete may vary considerably to that of another. Similarly, Riemann et al. (2002) used a treadmill protocol where the speed was set to ensure that subjects were working between 65 - 70% of their maximum heart rate for 20 minutes. However, again whilst this is controlled for within-subject effects, there is no control over the amount the tape is stressed between subjects.

Treadmill based protocols can be criticised for not being applicable to sport. Lohkamp et al. (2009) tried to overcome this by including 195 discrete changes of speed throughout a fifteen minute cycle of activity on a treadmill to represent match play during soccer. However a criticism of this method, and most others performed on a treadmill, is that the tape is only stressed in one plane, which is not applicable to most sports which place multi-planar demands on the ankle joint. A study by Paris et al. (1995) attempted to overcome this by including lateral movement on a treadmill. Subjects speed-walked on a treadmill inclined to 9° at a speed of 3 mph for 10 minutes with a combination of forwards walking and left and right carioca; carioca was defined by Paris et al. (1995) as a left/right facing crossover strides.

Some other studies have used a combination of the protocols previously mentioned, for example, Alt et al. (1999) used a combination of treadmill running, and mat based exercises including drop jumps, and slope jumps, and Ricard, Sherwood et al. (2000) used a combination of treadmill running, agility exercises and lower limb stretching. It is thought that a combination of exercises designed to stress the athletic taping in different planes is most representative of sport.

7.2.3 Summary of Literature

This literature review has highlighted that measures of neuromuscular control including postural stability and muscle reaction time are affected by exercise. In addition the effects of athletic taping are also influenced by exercise. There are a number of existing protocols used in the literature to reduce the mechanical restrictive nature of taping yet further research is required to establish which is most effective in doing so, whilst also limiting the effects of fatigue which may also influence postural stability and muscle reaction time measures.

7.3 Pilot Study Eight: Development of an Exercise Protocol

7.3.1 Abstract

Context: Many studies use exercise protocols in order to make results more applicable to sport and to stress lower limb athletic taping methods to reduce the mechanical restrictive nature of the tape, however there is no consistent method used across the literature.
Objective: To determine the most effective exercise protocol to reduce the mechanical effects of taping.
Subjects: Eight subjects (2 males, 6 females, age = 20 ± 2.2 years, height = 167.8 ± 7.9 cm, mass = 60.2 ± 7.8 kg).
Methods: Three exercise protocols (treadmill, figure of 8, and a combined treadmill and mat exercise protocol) were compared by measuring the reduction of the mechanical restriction of tape between pre and post exercise. Arch tape was measured by comparing pre and post-exercise navicular drop height, and ankle tape was measured by comparing pre and post-exercise plantar flexion – dorsi flexion and inversion – eversion range of motion using a universal goniometer.
Results: Each exercise protocol resulted in significant differences ($P < .05$) between pre and post-exercise measures for the three range of motion measures. The combined treadmill and mat exercise protocol resulted in the largest percentage decrease between pre and post-exercise measures.
Conclusions: The combined treadmill and mat exercise protocol was deemed must suitable for use in Study Six.

7.3.2. Introduction

There is no consensus within the literature on which exercise protocol is most suitable in reducing the mechanical effects of taping in order to expose potential neuromuscular effects whilst maintaining external validity. It is recognised that there are already concerns to testing with athletic taping with regards to the placement of the tape and the tension at which it is applied (as addressed in Pilot Study Seven). Therefore it was deemed crucial to identify an exercise protocol that was as controlled as possible in order to reduce variation among subjects during testing. Yet at the same time, it was also deemed important to stress the tape in a method applicable to sporting activities. An additional concern was that postural stability and muscle reaction time results may be altered by fatigue caused by an exercise protocol which could potentially mask any differences caused by the taping techniques.

The purpose of this pilot testing was to compare two exercise protocols as previously established in the literature, against a third protocol specifically designed for this study; a combined treadmill and mat exercise protocol. The combined protocol aimed to strike a balance between being sport specific (externally valid), yet consistent across subjects (internally valid). The most effective exercise protocol was determined by the protocol which resulted in increased range of motion which signifies reduced mechanical restriction. In addition to range of motion, the protocol with the most consistent step counts among subjects and most consistent rate of perceived exertion (RPE) score was also observed. This was to ensure each subject stressed the tape by the same amount, to minimise fatigue to prevent postural stability and muscular reaction times being compromised.

7.3.3 Methods

Subjects:

Subject characteristics are shown in Table 7.1. The inclusion and exclusion criteria were the same as those to be used in Study Six, as identified in Section 3.2.3.

Table 7.1 Subject Characteristics

Sex	Age (years)	Height (cm)	Mass (kg)
2 Males 6 Females	20 (2.2)	167.8 (7.9)	60.2 (7.8)

Values are mean (SD).

Procedures:

Each subject was tested during three sessions on separate days; one session for each of the testing protocols (treadmill, figure of 8, and combined treadmill and mat exercise protocol) in a randomised order. During each session, arch taping was placed on the subjects left foot, and ankle taping on the subject’s right foot. Before the subjects undertook each exercise protocol, measurements of range of motion were collected from both feet. Navicular drop height was used to measure range of motion on the left foot with arch taping using the same method as shown to be reliable in Pilot Study One. Plantar flexion - dorsi flexion and inversion - eversion measures were collected using a universal goniometer on the right foot with ankle taping using the same method as described in Pilot Study Seven. Subjects then undertook the exercise protocol with a pedometer attached to record the amount of steps taken during the exercise. The same Yamax Digi-Walker SW-200 pedometer as shown to be reliable by Schneider, Crouter and Bassett (2004) was used for this testing. Immediately after each protocol, step count and RPE were recorded and the measures of range of motion were repeated on both feet. The RPE scale is shown in Appendix Seventeen.

The exercise protocols selected were a treadmill based protocol as defined by Paris et al. (1995), an agility based figure of 8 protocol as defined by Laughman et al. (1980), and a combined treadmill and mat exercises protocol specifically designed for this study. In the procedure defined by Paris et al. (1995), the treadmill was set at 9° incline at 3 mph, subjects exercised for ten minutes in total; two minutes facing forwards, three minutes left facing carioca, three minutes right facing carioca and two minutes facing forwards again. In line with Paris et al. (2005), subjects were instructed to take maximum strides and to maintain foot contact for as long as possible. The figure of 8 protocol by Laughman et al. (1980) was selected for this pilot testing as it was clearly defined and therefore repeatable. Subjects were instructed to run at a 'self-selected' pace for 15 minutes, along a figure of 8 course 10 meters long and 2.5 m wide with 10 turns.

The combined treadmill and mat exercise protocol utilised aspects of several exercise protocols described in the current literature. It involved six minutes of treadmill based exercise (1.5 minutes forwards walking, two minutes left facing carioca, 30 seconds forwards walking and two minutes right facing carioca). The treadmill was set at 1% incline which was selected as it is shown to be representative of outdoor running (Jones & Doust, 1996), and 3 mph was selected as it was a safe speed to perform carioca on the treadmill without causing an undue risk of tripping. Mat based exercises were also incorporated, these included forward, diagonal and lateral lunges, drop jumps from a 30 cm bench, step up/step downs on to a 30 cm bench, toe raises off of a 10 cm platform, and maximal vertical jumps. Each exercise was performed ten times on each leg. This part of the exercise protocol lasted approximately six minutes in total.

Data Reduction and Statistical Analysis:

All data were normally distributed (Shapiro-Wilk $P > .05$), paired samples t-tests were used to determine whether there were any significant differences between range of motion pre and post-exercise for each exercise protocol. The a priori alpha level was set at .05. All statistical analyses were performed using SPSS (version 17.0; SPSS Inc, Chicago, IL). Percentage change was calculated to show differences between pre and post exercise for each exercise protocol.

7.3.4 Results

Results are shown in Table 7.2. Each exercise protocol resulted in a significant increase ($P <$ .05) in range of motion post-exercise in comparison to pre-exercise measures for both arch and ankle taping.

7.3.5 Discussion

The largest differences between pre and post-exercise range of motion for all three measures were after the combined treadmill and mat exercise protocol. This protocol also resulted in the smallest standard deviations for RPE (12 ± 1.2) and step count (1674 ± 53.3) in comparison to the other protocols which indicated that it is most consistent between subjects. The combined treadmill and mat exercise protocol was therefore selected for use in Study Six.

Table 7.2 Comparison of Three Exercise Protocols

	Navicular Drop Height (mm) (Arch Taping)			Range of Motion PF-DF (°) (Ankle Taping)			Range of Motion Inv-Ev (°) (Ankle Taping)			RPE	Total Steps
	Pre-Exercise	Post-Exercise	% Change	Pre-Exercise	Post-Exercise	% Change	Pre-Exercise	Post-Exercise	% Change		
Treadmill Protocol Paris (1995)	0.41 (0.1)	0.51* (0.1)	20%	38.19 (2.9)	45.75* (5.5)	17%	30.63 (5.3)	35.33* (7.6)	13%	13 (1.4)	1251 (61.4)
Figure of 8 Protocol Laughman (1980)	0.40 (0.1)	0.56* (0.1)	29%	41.75 (9.5)	48.63* (9.0)	14%	29.38 (5.5)	41.75* (3.5)	30%	10.5 (1.3)	1067 (128.8)
Treadmill and Mat Exercise Protocol	0.40 (0.1)	0.60* (0.1)	33%	39.25 (7.3)	52.25* (7.8)	25%	30.50 (5.2)	44.50* (4.9)	31%	12 (1.2)	1674 (53.3)

Values are mean (SD). PF: plantar flexion; DF: dorsi flexion; Inv: inversion; Ev: eversion; RPE: rate of perceived exertion. *Indicates significant difference in comparison to pre-exercise range of motion (*P* < .05).

7.4 Study Six: The Effects of Athletic Taping on Lower Limb Biomechanics after an Exercise Protocol

7.4.1 Abstract

Context: Many studies examine the effects of athletic taping immediately after application, however in doing so; results cannot be generalised to sport as it is unclear whether any effects identified are also apparent during exercise. In addition, it is unclear whether changes in postural stability and muscle reaction time are caused by the mechanical restrictive nature of the tape, or by a potential neuromuscular effect caused by the interaction of the tape on the skin.

Objective: To determine whether the effects of taping on postural stability and muscle reaction time are apparent after exercise.

Subjects: Thirty three volunteers with symmetrical feet were categorised into three groups according to navicular drop height measures. Fourteen subjects had neutral feet (navicular drop height 5 – 9 mm), ten subjects had pronated feet (navicular drop height ≥ 10 mm), and nine subjects had supinated feet (navicular drop height ≤ 4 mm).

Methods: Postural stability and muscle reaction time to a tilt platform perturbation were measured before and after exercise under three conditions; no-tape, arch tape and ankle tape.

Results: In the no-tape condition, postural stability parameters were poorer and muscle reaction times were slower after exercise. With arch and ankle taping, some measures of neuromuscular control improved after exercise in comparison to before exercise immediately after tape application.

Conclusions: This study confirms that without taping, neuromuscular control reduces after exercise; in addition, improvements in neuromuscular control caused by athletic taping appear to diminish after exercise. Therefore athletic taping may only be useful in terms of injury prevention immediately after application.

7.4.2 Introduction

A limitation of the majority of studies measuring the effects of athletic taping on lower limb biomechanics is that measurements are typically taken immediately after the application of tape. Whilst this provides an insight to the effects of newly applied tape, it has been shown that exercise has a loosening effect on tape (Leanderson et al., 1996; Martin & Harter, 1993; Paris et al., 1995), indicating that the results of newly applied tape cannot be generalised to during sport. In addition to not being applicable to sport, testing the effects of athletic taping after application also makes it difficult to determine whether any biomechanical differences observed are due to the mechanical restrictive nature of taping, or due to an underlying neuromuscular effect which may occur as a result of the interaction of tape on the skin. In order to overcome this, some studies have measured the effects of athletic taping after a period of exercise. This helps to increase external validity by testing subjects in a method more applicable to sport, and therefore more representative of the conditions in which injuries such as ankle sprains are more likely to occur. In pilot testing, the most effective exercise protocol in reducing the restrictive nature of tape, whilst maintaining consistent RPE and step count among subjects was a combined treadmill and mat exercise protocol.

The purpose of this study was to measure the effects of athletic taping on neuromuscular control in individuals with neutral, pronated and supinated foot structures pre and post-exercise. It was hypothesised that the no-tape condition would result in slower reaction times and higher postural stability parameters post-exercise in comparison to pre-exercise. In addition it was hypothesised that arch and ankle taping techniques would result in faster reaction times and lower postural stability parameters post-exercise in comparison to pre-exercise. These were expected to occur across all foot structures.

7.4.3 Methods

<u>*Subjects:*</u>

Thirty three volunteers participated in this study and were categorised into three groups dependant on navicular drop height measures using the same method as shown to be reliable in Section 3.2. The inclusion and exclusion criteria were the same as those identified in Section 3.2.3. Subject characteristics are shown in Table 7.3.

Table 7.3 Subject Characteristics

Group	Sex	Age (years)	Height (cm)	Mass (kg)	Navicular Drop Height (mm)
Neutral	5 Males 9 Females	21 (3.9)	171.6 (9.7)	68.4 (13.2)	7.2 (1.5)
Pronated	5 Males 5 Females	19 (1.1)	169.7 (9.6)	70.3 (10.4)	11.4 (1.1)
Supinated	5 Males 4 Females	21 (3.1)	172.3 (9.3)	69.5 (12.4)	2.0 (1.1)

Values are mean (SD).

Prior to testing, ethical approval was granted from the institutional ethics committee, and all subjects read a subject briefing document (Appendix One), provided written informed consent (Appendix Two) and completed a health screen form (Appendix Three).

Procedures:

Postural stability and muscle reaction time measures were repeated with each testing condition (no-tape, arch tape, and ankle tape) on three separate days. Following the application of tape, pre-exercise measures of postural stability and muscle reaction time measures were taken. Postural stability measures were collected using the same method as identified in Section 3.4.3 however only the three hopping tasks were included. The same muscle reaction time method as identified in Section 4.7.3 was also used in this study; electrodes were not moved throughout the duration of the one hour testing period. Subjects then underwent a treadmill and mat exercise protocol as identified in Pilot Study Eight. Following this, the postural stability and muscle reaction time measures were repeated before the tape was removed.

The arch taping technique used in this study was the modified low-Dye taping technique using a procedure described by Vicenzino et al. (1997) this technique used zinc oxide 2.5 cm tape. The ankle taping technique used in this study was the closed basket weave procedure as described by Perrin (2005) using zinc-oxide 4 cm tape. Both the arch and ankle taping techniques were applied following the same procedures as shown to be reliable in Pilot Study Seven.

<u>Data Reduction and Statistical Analysis:</u>

Recording for postural stability data began as soon as the pressure on the force platform exceeded 50 N, and data were collected at 1000 HZ. Centre of pressure data were collected, and six variables were calculated at two time frames (200 ms and three seconds) including peak COP, mean COP, total COP, total distance, average velocity, and 95% ellipse area using an Excel template (Microsoft Office, 2010) (Appendix Four). This procedure was shown to be reliable in Pilot Study Two. Definitions of each COP variable are shown in Appendix Five.

Muscle reaction time data was processed with an RMS filter using a 10 ms moving window. Data was then reduced using an Excel template (Microsoft Office, 2010); this is shown in Appendix Ten. As shown to be reliable in Section 4.5, muscle reaction time was defined as the time between the onset of the tilt mechanism, to the onset of the EMG signal when it reached a level of three standard deviations above the baseline for 25 ms consecutively. The baseline value was the average value recorded over 150 ms immediately prior to the onset of the tilt mechanism.

As data were normally distributed (Shapiro-Wilk $P > .05$), for each dependant variable (COP parameters, peroneus longus, tibialis anterior and gluteus medius reaction time), a separate two way repeated measures analysis of variance was performed with taping technique (no-tape, ankle tape, arch tape) and exercise (pre and post) as the independent variables. The a priori alpha level was set at .05. Where Mauchly's test of sphericity was significant, a Greenhouse-Geisser correction was applied. Where a significant taping by exercise interaction was observed, pre-planned paired samples t-tests were used to identify the source of significance by comparing pre and post-exercise measures for each taping technique. Effect size (η_p^2 values) was calculated where 0.01 - 0.06 indicated a small effect, 0.06 - 0.14 indicated a medium effect, and > 0.14 indicated a large effect. Observed power was also calculated. All statistical analyses were performed using SPSS (version 17.0; SPSS Inc, Chicago, IL).

7.4.4 Results

Postural stability and muscle reaction time results are shown in Appendices Eighteen-Nineteen.

Analysis of the three postural stability tasks found significant differences between each of the taping conditions at both pre and post-exercise. During the forward hop task, at 200 ms, significant interactions between taping and exercise were identified in both the neutral and pronated groups; the neutral group indicated an interaction within the 95% ellipse area measure $F(2, 26) = 5.4$, $P = .01$, $\eta_p^2 = 0.29$, observed power = 0.81. Pre-planned follow up tests indicated that the arch taping resulted in higher parameters post-exercise indicating a reduction in postural stability in comparison to pre-exercise ($P = .006$). Within the pronated group, a significant interaction was identified within the mean anterior measure $F(2, 18) = 4.09$, $P = .03$, $\eta_p^2 = 0.31$, observed power = 0.65. Further analysis revealed higher parameters post-exercise with the arch taping condition ($P = .005$).

During the forward hop task, at three seconds, differences were present within the supinated group in the no-tape condition in two measures; the 95% ellipse area $F(2, 16) = 4.8$, $P = .02$, $\eta_p^2 = 0.38$, observed power = 0.72, and mean posterior COP $F(2, 16) = 7.1$, $P = .01$, $\eta_p^2 = 0.47$, observed power = 0.88. Pre-planned follow up tests indicated that both measures were higher post-exercise, suggesting that postural stability reduces following the exercise protocol (P = .025, P = .025 respectively).

During the diagonal hop task, at 200 ms, tape by exercise interactions were present in the neutral and supinated groups. The neutral group identified an interaction in the average velocity measure $F(2, 26) = 3.6$, $P = .04$, $\eta_p^2 = 0.22$, observed power = 0.62. Further analysis indicated that in the no-tape condition, parameters were higher post-exercise indicating that postural stability had reduced (P = .025). In the supinated group, an interaction was identified in the measure of total lateral centre of pressure; $F(2, 16) = 3.7$, $P = .04$, $\eta_p^2 = 0.31$, observed power = 0.59. Follow up tests indicated that both the no-tape (P = .021) and ankle tape (P = .001) conditions resulted in reduced postural stability post-exercise.

During the diagonal hop task, at three seconds, in the pronated group, interactions were identified in the mean medial $F(2, 18) = 4.5$, $P = .02$, $\eta_p^2 = 0.34$, observed power = 0.70, and mean posterior measures $F(2, 18) = 3.6$, $P = .04$, $\eta_p^2 = 0.28$, observed power = 0.59. Analysis revealed that the arch taping condition resulted in increased parameters post-exercise in comparison to pre exercise (P = .012, P = .025 respectively). No differences were identified in the lateral hop condition.

Analysis of muscular reaction time measures indicated that differences present were in the peroneus longus muscle in both the neutral $F(2, 26) = 3.5$, $P = .041$, $\eta_p^2 = 0.21$, observed power = 0.61 and pronated groups $F(2, 18) = 5.14$, $P = .01$, $\eta_p^2 = 0.36$, observed power = 0.75. Follow up analysis revealed that in both groups, differences were identified in the no-tape condition (P = .004, P = .020), and the ankle tape condition (P = .019, P = .023). As shown in Figures 7.1-2, both groups indicated slower reaction time post-exercise in the no-tape condition, suggesting that reaction time worsens with exercise. Both groups also showed that ankle taping results in faster reaction times post-exercise, suggesting than when wearing ankle taping, reaction times improve with exercise.

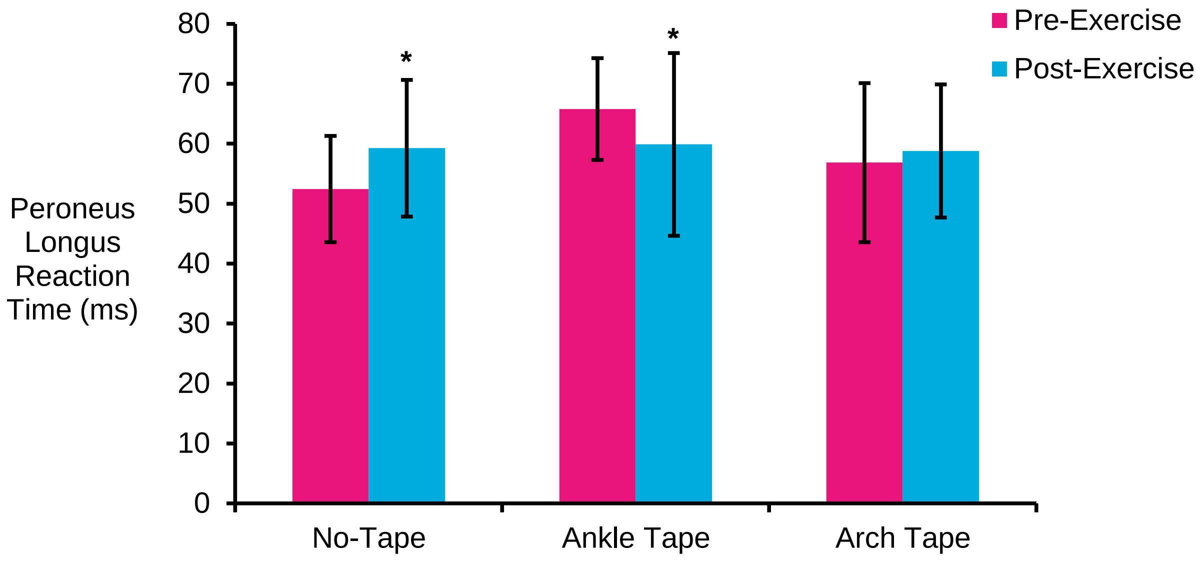


Figure 7.1 Reaction time of the peroneus longus muscle with lower limb athletic taping pre and post-exercise on subjects with neutral feet: showing standard deviation. ***** Indicates significant difference in comparison to pre-exercise condition (*P* < .05).

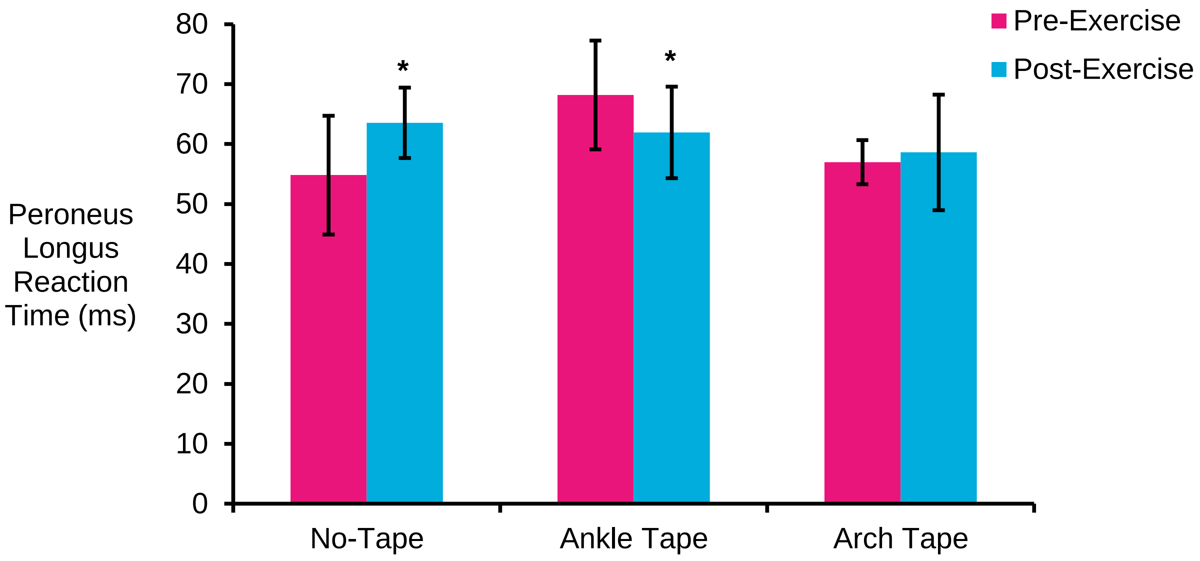


Figure 7.2 Reaction time of the peroneus longus muscle with lower limb athletic taping pre and post-exercise on subjects with pronated feet: showing standard deviation. ***** Indicates significant difference in comparison to pre-exercise condition (*P* < .05).

7.4.5 Discussion

This is the first study to have measured the effects of athletic taping pre and post-exercise on subjects with neutral, pronated and supinated foot structures using postural stability and muscle reaction time parameters. It was hypothesised that the no-tape condition would result in slower reaction times and higher postural stability parameters post-exercise in comparison to pre-exercise. In addition it was hypothesised that arch and ankle taping techniques would result in faster reaction times and lower postural stability parameters post-exercise in comparison to pre-exercise. Both hypotheses were expected to occur across all foot structures. Therefore the hypotheses are partially accepted as results indicated slower peroneus longus reaction times and higher postural stability parameters in the no-tape control post-exercise. In addition, faster peroneus longus reaction times were identified in the ankle tape condition post-exercise and higher postural stability parameters in both the arch and ankle taping conditions post-exercise.

In the current study, it was identified that in the no-tape condition, the reaction time of the peroneus longus muscle was significantly slower post-exercise in the neutral and pronated groups. In addition, in measures of postural stability, all differences in the no-tape condition indicated higher post-exercise values indicating that postural stability worsens during exercise. Differences were identified in the neutral and pronated groups in measures of muscle reaction time, and in neutral, pronated and supinated groups in measures of postural stability. Although other studies have not considered the effects of foot structure on postural stability post-exercise, these findings are in support of Gribble and Hertel (2004), Benesch et al. (2000) and Lohkamp et al. (2009) and imply that individuals are more vulnerable to injury after a period of exercise emphasising the need for preventative measures such as athletic taping. However as the rate of injury was not measured in this study, this speculation is yet to be researched. These findings are in contrast to Leanderson et al. (1996) who found that subjects had improved postural stability after exercise suggesting that warmer muscle resulted in less uncontrollable movements, however, a limitation to this study is that subjects were tested after a 'two hour soccer practice session', which can be criticised for being both vague and unrepeatable.

With ankle taping, the peroneus longus reaction times in both the neutral and pronated groups were significantly faster post-exercise in comparison to pre-exercise. Interestingly, the ankle taping condition was the only taping technique found to result in significantly different reaction times in comparison to the no-tape condition in Study Five. It appears that whilst ankle tape results in an immediate increase in peroneus longus reaction time, after a 12 minute exercise protocol this increase diminishes. Whilst it was not assessed statistically, it appears that the reaction times when wearing ankle taping post-exercise are similar to those with no-tape, indicating that the tape has no effect after a 12 minute exercise protocol. No other studies were found to have identified differences in reaction time post-exercise, however Alt et al. (1999) found that with ankle taping, the speed of inversion was slower, yet after a similar exercise protocol to that used in the current study, the speed of inversion was faster. A similar finding was reported by Lohrer et al. (1999) who found that the proprioceptive amplitude ratio had reduced post-exercise. Both Alt et al. (1999) and Lohrer et al. (1999) attributed these findings to the loosening of the tape after exercise. Lohrer et al. (1999) suggested that the initial reduction in inversion velocity caused by athletic taping enables time for the dynamic defence mechanism to come into action in time to protect the joint. However it appears that in order for this effect to continue during exercise, according to the results of the current study, the athletic tape must be reapplied at least every 12 minutes.

Both arch and ankle taping resulted in significantly higher postural stability parameters post-exercise in comparison to pre-exercise measures suggesting that postural stability worsened after exercise. Most previous studies examining the effects of athletic taping post-exercise use range of motion measures in non-weight bearing positions to assess measures of neuromuscular control rather than measures of muscular reaction time, or postural stability. This makes comparisons difficult; however, the results of this study are in support of Leanderson et al. (1996) who also found that the effects of ankle taping on postural stability significantly reduced after exercise. The results are also in support of Ator et al. (1991) who found that the effects of arch taping in maintaining the subtalar neutral position significantly reduced after just ten minutes of jogging.

The differences which were identified in Study Five are also apparent in the current study, however as the differences between taping techniques were not compared in this study this finding is only based on the observations of means. Of the differences identified in Study Five, the only measure found to have resulted in reduced stability post-exercise was the forward hop task where the pronated group showed poorer postural stability as measured by mean anterior centre of pressure post-exercise. All other differences identified in Study Five were not shown to be affected by the exercise protocol indicating that the other differences identified in Study Five are still apparent after exercise.

Previous studies have indicated that some neuromuscular effects are still apparent after exercise when the restrictive nature of tape has been reduced (Robbins et al., 1995). In the current study, the findings of reduced postural stability post-exercise with the use of tape do not provide support for the theory that athletic taping provides a neuromuscular effect caused by cutaneous stimulation. However the current study does give support for the mechanical effects of athletic tape which work to control postural stability by restricting the range of motion across the foot and ankle. As only one of the differences identified in Study Five reduced after exercise in the current study, some support is provided for the neuromuscular effects of taping, as one would expect all of the parameters identified as significant in Study Five to show significant reductions in stability in Study Six if no neuromuscular effects were caused.

As previously discussed, all reaction time parameters identified as significant in Study Five were shown to reduce in the current study therefore not supporting the theory that athletic taping has a neuromuscular effect on lower limb biomechanics. Perhaps the neuromuscular effects are not apparent in muscle reaction time measures, but are in other measures including postural stability as evident in this study, and in measures of joint position sense as suggested by Robbins et al. (1995). Clearly this area requires further research in order to determine whether arch and ankle taping are effective in improving postural stability.

A reason for the current study indicating a significant decrease in postural stability in comparison to previous studies may be due to the exercise protocol used. The exercise protocol in the current study was specifically designed to stress the athletic taping techniques in a range of planes, whereas exercise methods used in previous studies typically only stress the tape in one plane (Ator et al., 1991; Holmes et al., 2002). By only stressing the tape in one plane, it is feasible that the restriction of the tape will remain high in other planes, therefore allowing further movement in the plane that has been stressed in comparison to others. With the example of arch taping which has previously been measured using navicular drop height measures pre and post-exercise (Holmes et al., 2002), if the tape has only been stressed in the anterior-posterior plane when walking forwards on a treadmill, the restriction in the medial and lateral plane is not likely to have changed. This may indicate why some previous studies have not identified differences post-exercise. The same applies for ankle taping where some studies have only measured inversion – eversion range of motion after a forward running protocol (Martin & Harter, 1993). Aside from running on an athletics track, stressing taping in multiple planes as done in the current study is more representative of a range of sports, indicating that this study has improved external validity in comparison to previous studies only stressing the tape in one plane.

In the current study, differences in postural stability were identified at both 200 ms and three seconds, this is in contrast to Studies One and Five where differences were only observed in the first 200 ms. Balance was maintained for three seconds in order to make results comparable to previous literature, however as few sports require athletes to maintain single limb balance for so long during match play, it is thought that the initial 200 ms of landing provides a more valuable insight into the postural stability deficits. It is also important as it is beneath the level of conscious control (Schmidt & Lee, 2005), and the time frame within which injuries occur (Fong et al., 2009). In addition, 200 ms indicated higher levels of reliability among most postural stability parameters in comparison to the same parameters at three seconds during Pilot Study Two.

7.4.6 Limitations

A limitation of this study is that whilst it used an exercise protocol which was more representative of sport than many others used in previous literature, it only lasted for approximately 12 minutes. This is in line with the time frames of many other exercise protocols used in this area of research, however it means that the results are only applicable to the first 12 minutes of sport, and so it cannot be assumed that the same results apply to longer time frames.

7.4.7 Recommendations for Future Research

It would be interesting to observe the effects of taping over more intermittent time frames, in order to observe the effects of taping after the 12 minutes shown in this study. In addition it would be interesting to determine at what point the taping results in changes in neuromuscular control. This time frame would be key in identifying when tape may need to be reapplied in order to sustain the benefits of taping identified in Study Five. The effects of ankle taping on inversion velocity also appears to be key in determining whether taping is useful in the prevention of ankle sprains and therefore warrants further research on subjects with neutral, pronated and supinated foot structures.

7.4.8 Conclusion

This study has shown that the reaction times of the peroneus longus muscle in the neutral and pronated groups are delayed through exercise, in addition postural stability parameters also increase, indicating poorer neuromuscular control in comparison to pre-exercise measures. With ankle tape application, the peroneus longus reaction time is delayed in the neutral and pronated groups; however this effect diminished after exercise as the athletic taping loosens. Some postural stability parameters are also reduced post-exercise with ankle and arch taping, as the taping loosens through use. Most differences in neuromuscular control identified in Study Five were not shown to be different in Study Six post-exercise, indicating that these effects may still be present after a period of exercise. Further research is required in order to determine whether athletic taping results in a neuromuscular effect as a result of the interaction of the tape on the skin. In the meantime, it appears that regular re-application of tape may be required in order to maintain the benefits identified in Study five.

Chapter Eight

Discussion

8.1 Introduction to Discussion

The original aims of this thesis were to examine the effects of foot structure on lower limb biomechanics as measured by aspects of neuromuscular control, and to examine the effects of foot structure and athletic taping on lower limb biomechanics as measured by aspects of neuromuscular control. A further aim to examine the effects of asymmetrical foot structure on lower limb biomechanics was added to the thesis in light of the observation that 31% of subjects in Studies One and Two had asymmetrical foot structures.

In order to explore these aims, it was hypothesised that individuals with pronated and supinated foot structures will have reduced neuromuscular control on measures of dynamic postural stability (poorer) and muscular reaction time (slower) in comparison to those with neutral feet. In addition it was hypothesised that athletic taping will affect lower limb neuromuscular control on measures of dynamic postural stability and muscular reaction time in comparison to a no-tape control.

The aims were addressed over six studies showing progression from the original concept of variances in foot structure, through to techniques of injury prevention on individuals who may be more susceptible to injury during sport. Within this discussion, these aims will be reviewed in reflection of the research undertaken. The clinical implications, contributions to literature, limitations, and recommendations for future research are also discussed.

8.2 Aim: To Examine the Effects of Foot Structure on Lower Limb Biomechanics

The aim to examine the effects of foot structure on lower limb biomechanics was addressed throughout this thesis with each study involving the comparison of subjects with neutral, pronated and supinated feet. Perhaps the most important aspect in addressing this aim was accurate identification of subjects with different foot structures. Despite a range of measures used in previous literature, navicular drop height was selected as it was most widely used in previous literature thus enabling comparisons to previous studies; in addition, it is relatively easy to perform without requiring expensive equipment. The reliability of the navicular drop height measurement was addressed in Pilot Study One which indicated that navicular drop height was consistent when measured by the same examiner over different days. Navicular drop height was subsequently used in Studies One-Six in order to classify subjects into neutral, pronated and supinated groups.

Following an extensive review of literature, it was identified that the effects of foot structure on postural stability had previously been measured (Cote et al., 2005; Hertel et al., 2002; Tsai et al., 2006), however the previous measures were based on static tasks, which could be criticised for having low external validity and therefore not being applicable to sport. Study One therefore sought to address this gap in the literature by measuring the effects of neutral, pronated and supinated foot structures on dynamic postural stability. Pilot Study Two measured the reliability of seven dynamic postural stability tasks using six different postural stability parameters at two time periods. It was identified that parameters within the 200 ms time frame appears to be most reliable, however as no previous studies were found to have measured postural stability at 200 ms, three seconds was also included in further studies in order to make comparisons to previous literature. Additionally, it was found that more dynamic tasks resulted in reduced reliability. Again despite this, it was decided that all tasks would be used for further study in order to research previously unexplored areas of postural stability during sport specific movements.

Using the seven dynamic postural stability tasks identified in Pilot Study Two, the results of Study One indicated that individuals with pronated and supinated foot structures had reduced postural stability in comparison to individuals with neutral feet during diagonal and lateral hopping tasks. Whilst this was the first study to have measured foot structure during these dynamic tasks, the results were in support of previous studies using static measures of postural stability where it was also identified that pronated and supinated foot structures resulted in reduced postural stability (Cote et al., 2005; Hertel et al., 2002; Tsai et al., 2006). Interestingly, in Study One, most differences were identified in the initial 200 ms on landing. As sporting injuries such as ankle sprains occur within this time frame (Fong et al., 2009), findings of instability on landing among individuals with pronated and supinated foot structures is particularly important as it implies that the may be more vulnerable to injury in comparison to those with neutral feet. In addition, differences in Study One were only identified in tasks involving diagonal and lateral movement. These movements are most representative of the mechanisms of injuries such as lateral ankle sprains during sport which adds further support to the theory that individuals with pronated and supinated feet may be more at risk of injury that those with neutral feet. The results of Study One confirmed the need for further research into the effects of foot structure on lower limb biomechanics.

To further explore the effects of foot structure on lower limb biomechanics, Study Two measured the effects of neutral, pronated and supinated foot structures on muscular reaction time to a tilt platform perturbation. Before this study was conducted, a series of pilot tests were carried out in order to determine the reliability of the proposed methods. Pilot Study

Three used MVIC’s to determine the reliability of electrode placement on the peroneus longus, tibialis anterior, and gluteus medius muscles. It was found that by following the guidelines by SENIAM, electrode placement was highly reliable over two days and was therefore used in subsequent studies within this thesis. Pilot Study Four indicated that both the left and right tilting components of the dynamic tilt platform were reliable for use in Studies Two-Six. Pilot Study Five compared different methods for the onset detect of muscle reaction time and found that a 10 ms RMS moving window, 150 ms baseline, 25 ms burst and 3 standard deviations above the baseline were most consistent over different days and were therefore used in subsequent studies within this thesis. Pilot Study Six combined all aspects of the muscle reaction time methods and showed that it was a highly reliable method over different days; the same method was subsequently used in Studies Two-Six.

Using the methods identified in Pilot Studies Three-Six, Study Two indicated that both those with pronated and supinated feet had slower peroneus longus reaction times in comparison to those with neutral feet. This is another highly important finding which supports the results of Study One and adds further evidence to the argument that individuals with pronated and supinated feet may be more at risk of injury in comparison to individuals with neutral feet. This was the first study to have measured the effects of muscle reaction time on individuals with neutral, pronated and supinated foot structures, and therefore it is hoped that this finding will stimulate further research in the area in order to clarify whether individuals with pronated and supinated feet are more vulnerable to ankle sprains in comparison to those with neutral feet.

The reason why individuals with pronated and supinated feet have different responses to lower limb biomechanical tasks is unclear; previous studies have suggested they may be caused by either mechanical differences, or differences in peripheral input. Hertel et al. (2002) suggested that individuals with cavus feet have no anatomical block when pronating in comparison to a neutral or planus feet. This is a feasible explanation as on landing it would mean that individuals with cavus feet would have a wider range of movement until the medial aspect of the foot and the ground make contact. In support of this, in Study One the peak medial centre of pressure during the lateral hop was significantly higher in the supinated group in comparison to the neutral group. However one would expect more differences in the medial plane if this explanation for differences in foot structure was the sole cause. Additionally, this theory implies that individuals with pronated feet would have increased postural stability caused by excessive ground contact, however as shown in Study One, the pronated group also resulted in reduced postural stability in some parameters during diagonal and lateral tasks. When considering the pronated foot, a theory by Tsai et al. (2006)

is more likely, whereby the hypermobility in the subtalar joint results in increased postural stability parameters. This is consistent with Cote et al. (2005) who suggested that those with pronated feet are more reliant on soft tissue structures for stability.

Hertel et al. (2002) suggested that differences in postural stability between foot structures may be caused by differences in peripheral input; individuals with cavus feet have less plantar surface contact area than those with neutral and pronated feet and therefore have reduced sensory cues resulting in reduced mechanisms to maintain balance. Again this theory implies that individuals with pronated feet receive increased sensory input due to increased plantar contact area, however this does not appear to benefit individuals during dynamic postural stability or muscle reaction time tasks. Other suggestions circulate around supinated feet having a smaller base of support therefore having less stability than neutral or pronated feet which have wider bases of support. However again these theories are questionable as Study One also resulted in differences in postural stability in the pronated group, and also, this theory implies that individuals with smaller feet are less stable than individuals with larger feet. This seems unlikely, however as no previous studies were found in this area, it requires further research.

A further theory that requires exploration is that differences in lower limb neuromuscular control could be associated with the peroneus longus muscle. The peroneus longus was the only muscle found to result in significant differences between foot structures in Studies Two, Five and Six. Differences in muscle activity caused by foot structure are unsurprising as the peroneus longus has a complicated anatomical path which wraps underneath the foot and inserts on the lateral surface of the base of the first metatarsal via a system of pulleys around the lateral malleolus, the peroneal tubercle and the cuboid (Johnson & Christensen, 1999). In addition to its key function of everting the foot the peroneus longus is also a stabilising muscle which serves to maintain balance during the stance phase of gait (Louwerens, van Linge, de Klerk, Mulder & Snijders, 1995). With this in mind, it is possible that the biomechanical alterations of pronated and supinated foot structures could affect the peroneus longus muscle therefore explaining both the reduced muscle reaction time and reduced postural stability. No other studies were found to have associated foot structure with the function of the peroneus longus muscle and so this speculation is yet to be researched.

8.2.1 Clinical Implications

The clinical implications of the findings of Studies One and Two are substantial. Individuals with pronated and supinated foot structures were found to have poorer postural stability and slower muscle reaction times in comparison to individuals with neutral feet. Both findings suggest that individuals with pronated and supinated foot structures may be more vulnerable to injuries such as lateral ankle sprains in comparison to individuals with neutral feet. In light of this, clinicians should consider the use of preventative measures such as athletic taping and bracing in order to prevent injury during sport on athletes with pronated and supinated feet. The use of specific exercises during prehabilitation to target the evertor muscles in order to stimulate faster muscle activation should also be considered.

<u>8.3 Aim: To Examine the Effects of Foot Structure and Athletic Taping on Lower Limb Biomechanics</u>

The aim to examine the effects of foot structure and athletic taping on lower limb biomechanics was addressed in Studies Five and Six, where the effects of athletic taping techniques on individuals with neutral, pronated and supinated feet were measured both before and after exercise. The purpose of these studies was to determine whether athletic taping was a suitable preventative technique to be used on individuals with different foot structures during measures applicable to sport. Before conducting Study Five, a thorough review of literature indicated that athletic taping techniques appear to produce a combination of both mechanical and neuromuscular effects, including changes in range of motion (Paris et al., 1995), changes in muscle activity (Franettovich et al., 2008), and changes in postural stability (Leanderson et al., 1996). No previous studies were found to have measured the effects of athletic taping on subjects with neutral, pronated and supinated foot structures.

A common concern with studies involving the use of athletic taping is the reliability of the application of tape. Pilot Study Seven therefore compared the reliability of athletic taping application over two days and found that according to range of motion measures, both the arch taping technique and the ankle taping technique resulted in high reliability over different days. The same taping techniques were then used in Studies Five and Six. Study Five measured the effects of athletic taping on individuals with neutral, pronated and supinated foot structures immediately after application and found that initially, the arch and ankle taping techniques appeared to have positive effects on postural stability and muscle reaction time. Muscle reaction time of the peroneus longus was found to be significantly slower in comparison to the no-tape control in both neutral and pronated groups; however this was thought to be due to the mechanical restriction of the tape preventing normal movement at the joint. This is in line with previous studies where it has been identified that ankle taping results in decreased speed of inversion at the ankle (Karlsson & Andreasson, 1992). Whilst inversion speed was not measured in Study Five, it is thought that the ankle taping technique is actually more likely to be beneficial in the prevention of ankle sprains despite resulting in slower muscle reaction time measures, however this requires further research. The ankle taping technique had no effect on the reaction time of individuals with supinated feet. Postural stability improved with ankle and arch taping on all three foot structures. The proprioceptive taping technique used in this study had no effect on postural stability or muscle reaction time which suggests that proprioceptive taping techniques are only beneficial in improving joint position sense as a measure of neuromuscular control as identified by Matsusaka et al. (2001) and Simoneau (1997).

Study Six developed the concept of measuring the effects of athletic taping on individuals with neutral, pronated and supinated foot structures, by making the methodology more applicable to sport. This method involved the use of an exercise protocol which was developed in Pilot Study Eight which compared three exercise protocols and found that the most effective method in reducing the mechanical effects of tape was a treadmill and mat exercise protocol. This method was used in Study Six where the results of several measurement parameters indicated reduced postural stability across three hopping tasks and reduced muscle reaction time.

The results of Studies Five and Six provide support for the mechanical effects of taping whereby the restrictive nature of the taping causes reduced range of motion which in turn affects the response to lower limb biomechanical tasks. As the proprioceptive taping technique had no effect in Study Five, and the results in Study Six after exercise indicated slower muscle reaction time and poorer postural stability across some parameters, there is limited evidence to support previous suggestions of an underlying neuromuscular effect of taping.

One factor questioning the theory of the mechanical effects of taping is that the supinated group was not affected by the ankle taping condition during measures of muscular reaction time. As both the neutral and pronated groups were shown to have reduced reaction time after the application of tape, and as the application of tape had been shown to have strong reliability over different days during Pilot Study Seven, one would expect the supinated group to also have reduced peroneus longus reaction time. Within the measures of postural stability, ankle taping resulted in increased average velocity in subjects with supinated feet and also appeared to cause more gross body movements in comparison to those with neutral and pronated feet. However perhaps due to the anatomical variation in foot structure in comparison to these with neutral and pronated feet, the athletic taping does not apply the same amount of restriction to the peroneus longus muscle therefore not affecting the reaction time to a tilt platform perturbation.

8.3.1 Clinical Implications

As shown in Studies Five and Six, the use of athletic taping as a preventative aid is not supported on individuals with supinated feet, as it appears to have no effect of muscle reaction time. In addition, during the forward hop task, the increased velocity in the supinated group as a result of the ankle taping is a potential concern, particularly with the observed increase in gross body movement in comparison to the neutral and pronated groups. This may have detrimental effects during sport, particularly on unstable playing surfaces. In neutral and pronated groups where initially the taping resulted in slower reaction times (likely to be due to reduced inversion velocity), the effects diminished after a short period of exercise indicating the need for regular reapplication, or a more permanent technique such as ankle bracing.

8.4 Aim: To Examine the Effects of Asymmetrical Foot Structure on Lower Limb Biomechanics

It was observed during Studies One and Two, that 31% of the subjects used in testing had asymmetrical feet, where one foot was neutral, and the other either pronated or supinated. This was an unexpected finding as most studies in previous literature only measure one limb, and so the findings of asymmetry are rarely reported. Despite having low subjects numbers available for secondary analysis, it was thought that as asymmetry affected a large proportion of the subjects used in Studies One and Two it should not be ignored. Therefore an additional aim was added to the thesis in order to supplement the findings of Studies One and Two, and further explore the effects of foot structure on lower limb biomechanics.

In order to explore the effects of asymmetry further, Study Three measured the effects of limb dominance on subjects with neutral foot structures. No significant differences were found between the dominant and non-dominant limb during measures of postural stability or muscle reaction times. This study enabled further research into the effects of foot structure asymmetry as limb dominance could be dismissed as a potential intervening variable when analysing differences between the contralateral feet of individuals with asymmetry. Study Four measured the effects of foot structure asymmetry, however due to low subject numbers, only descriptive data could be used for analysis. The study indicated that the contralateral limbs of subjects with asymmetrical feet respond differently to both postural stability and muscle reaction time tasks. Whilst these results could not be analysed statistically, the study opens doors for further research on a much larger scale in order to fully understand the implications for individuals with asymmetrical foot structures.

8.4.1 Clinical Implications

In light of the results of Study Three, clinicians can exclude limb dominance as a potential source of variation when comparing contralateral limbs on individuals with symmetrical feet. Clinicians should however be cautious of generalising findings on individuals with asymmetrical feet as in light of Study Four it appears that asymmetrical limbs appear to respond differently to muscular reaction time and postural stability tasks. In terms of injury prevention, it appears that the contralateral limbs of individuals with asymmetrical feet may require different attention in order to reduce the risk of injury during sport.

8.5 Review of Hypotheses

With reference to the original hypotheses highlighted in the introduction of this thesis (Section 1.3), it can be concluded that both hypothesis can be formally accepted as individuals with pronated and supinated foot structures were shown to have reduced neuromuscular control on measures of dynamic postural stability (poorer) and muscular reaction time (slower) in comparison to those with neutral feet. Additionally, athletic taping was shown to affect lower limb neuromuscular control on measures of dynamic postural stability and muscular reaction time in comparison to a no-tape control. Specific hypotheses were detailed in each study.

8.5.1 Hypotheses for Specific Studies

In some studies, the experimental hypotheses were ‘partially accepted’. This occurred when some parameters were identified to be significantly different, and others within the same test were not. In this case, one must reject the null hypotheses (as some differences were identified), however cannot fully accept the experimental hypotheses unless all parameters indicated a significant difference. Therefore, the hypotheses were partially supported. The hypotheses could be re-worded, however for some studies where many parameters were observed, this would result in many hypotheses. For example, in Study one it was hypothesised that subjects with pronated and supinated foot structures would have significantly higher postural stability parameters in comparison to those with neutral foot structures across all seven tasks. If this hypothesis was divided into individual hypotheses for each parameter to be tested and each time frame, it would result in 330 separate hypotheses. A calculation of all individual hypotheses for Studies One-Six resulted in 2067 separate hypotheses.

8.6 Review of Methodology

8.6.1 Validity

An underlying theme in this thesis was to conduct research in a manner in which strong internal and external validity were maintained. In reflection of Pilot Studies One-Eight, and Studies One-Six, it is thought that this balance was generally well maintained. There were a number of areas where internal validity could be improved; for example, the use of light gates in Study One may have had a small effect in maintaining consistent running speeds among subjects, however pilot testing indicated that the running speeds were highly consistent over different days without light gates (Section 3.3.6). Additionally it was thought that in order to maintain external validity, the use of light gates should be avoided. It could also be argued that the external validity of the exercise protocol used in Pilot Study Eight could have been improved. However in order to maintain internal validity which was clearly lacking in a number of studies in the literature which had used more sport specific protocols, it was decided that the treadmill and mat exercise protocol provided the best balance in maintaining experimental control, whilst stressing the tape in a range of planes as done in during sport.

8.6.2 Reliability

A lot of attention in this thesis was placed on the reliability of methods to be used in Studies One-Six. As the testing period for each study extended over several months due to the availability of subjects, and the time required for each testing session, it was crucial to identify that the methods to be used were consistent over different days. Generally the measures of reliability were high for each pilot study indicating that any differences identified in Studies One-Six were a result of the independent variable, rather than due to measurement error.

8.7 Limitations

The main limitation of the work in this thesis was present in Study Four, where due to low subject numbers only descriptive data could be analysed. Were more subjects initially recruited for Studies One and Two, more data would have been available for secondary analysis and statistics may have been calculated. This would have provided a better insight into the effects of asymmetrical foot structures on lower limb biomechanics. Additionally, more subjects would result in increased statistical power. Increased power results in increased probability of correctly rejecting the null hypothesis, or correctly identifying a true significant difference between data sets. Whilst low subject numbers was particularly an issue for Study Four, increased subject numbers across all studies would have resulted in increased power, and therefore strengthened the overall findings of this thesis.

8.8 Contributions to the Literature

This research has contributed to the literature in a number of ways; this is the first in depth study into the effects of foot structure on lower limb biomechanics during methods applicable to sport. This research builds upon previous studies which had measured the effects of foot structure on static postural stability, and instead analyses the effects of foot structure during dynamic activity. In addition this thesis includes the first study to have measured the effects of foot structure on muscular reaction time to a tilt platform perturbation. Whilst aspects of neuromuscular control have been measured with the use of ankle taping previously, existing literature were inconclusive. Additionally, no previous studies had measured the effects of arch taping on dynamic postural stability and muscle reaction time. Therefore this research adds to the literature by measuring both the effects of ankle and arch taping on postural stability and muscle reaction time using subjects with neutral, pronated and supinated feet. A further addition to the literature is the finding of 31% of subjects with asymmetry in measures of navicular drop height. This supports the findings of Shultz and Nguyen (2007) and confirms the need for further research into the effects of asymmetrical foot structures during dynamic movement.

Study Two; the effects of foot structure on muscular reaction time, has already been accepted for publication in the Journal of Athletic Training, and a number of other papers taken from this thesis will be submitted in the near future.

8.9 Recommendations for Future Research

The most pressing area for further research uncovered by the work in this thesis is a large scale study on the effects of foot asymmetry during sport. The procedure in Study Four needs to be repeated on a much larger scale with more subjects so that statistical analysis can be performed in order to fully understand the effects of foot structure asymmetry during sport.

A further area for research would be to measure the effects of neutral, pronated and supinated foot structures on subjects with a history of ankle sprains. All studies within this thesis excluded subjects with a history of ankle sprains in order to remove potential intervening variables which may affect measures of postural stability and muscle reaction time. It has been found that individuals with functional ankle instability have slower peroneus longus reaction times and reduced postural stability in comparison to stable controls (Mitchell et al., 2008a; Mitchell et al., 2008b). No previous studies have been found to have measured postural stability or muscle reaction times on individuals with functional ankle instability and neutral, pronated and supinated foot structures. It is speculated that individuals with functional ankle instability, and pronated or supinated feet, would have poorer neuromuscular responses to both postural stability tasks and tilt platform perturbations indicating that this subject group would be at increased risk of obtaining further ankle sprains. However until this area is further researched this theory remains speculation.

8.10 Conclusion

This thesis has provided a valuable insight into the effects of neutral, pronated and supinated foot structures on lower limb biomechanics. The findings of this thesis indicate that dynamic postural stability is poorer in individuals with pronated and supinated feet, particularly during diagonal and lateral movements. Furthermore, individuals with pronated and supinated feet have slower muscle reaction times to a tilt platform perturbation in comparison to individuals with neutral feet. It was identified that 31% of subjects used in this study had asymmetrical feet. Limb dominance did not affect postural stability or muscle reaction time; however non-statistical analysis indicated that the contralateral feet of individuals with asymmetrical foot structures appear to respond differently to measures of postural stability and muscle reaction time. Arch and ankle taping were found to improve postural stability in the neutral and pronated groups, additionally; the peroneus longus reaction time reduced in these groups with ankle taping, however after exercise these effects reversed.

References

Akhbari, B., Takamjani, I. E., Salavati, M. & Sanjari, M. A. (2007). A 4-week biodex stability exercise program improved ankle musculature onset, peak latency and balance measures in functionally unstable ankles. *Physical Therapy in Sport, 8*, 117-129.

Alexander, C. M., McMullan, M. & Harrison, P. J. (2008). What is the effect of taping along or across a muscle on motorneurone excitability? A study using tricepts surae. *Manual Therapy, 13*, 57-62.

Allen, M. K. & Glasoe, W. M. (2000). Metrecom measurement of navicular drop in subjects with anterior cruciate ligament injury. *Journal of Athletic Training, 35*(4), 403-406.

Allison, G. T., Hooper, D., Martin, L., Tillberg. N. & Woodhouse. (1999). The influence of rigid taping on peroneal latency in normal ankles. *Australian Journal of Physiotherapy, 45*, 195-201.

Alt, W., Lohrer, H. & Gollhofer, A. (1999). Functional properties of adhesive ankle taping: neuromuscular and mechanical effects before and after exercise. *Foot and Ankle International, 20*(4), 238-245.

Ator, R., Gunn, K., McPoil, T. G. & Knecht, H. G. (1991). The effect of adhesive strapping on medial longitudinal arch support before and after exercise. *Journal of Orthopaedic and Sports Physical Therapy, 14*, 18-23.

Ball, N. & Scurr, J. (2010). An assessment of the reliability and standardisation of tests used to elicit reference muscular actions for electromyographical normalisation. *Journal of Electromyography and Kinesiology, 20*, 81-88.

Bartlett, R. (2007). *Introduction to sports biomechanics; analysing human movement pattern.* (2nd ed.). London: Routledge.

Beam, J. W. (2006). Orthopedic *Taping, Wrapping, Bracing & Padding*. Philadelphia: F. A. Davis Company.

Beckett, M. E., Massie, D. L., Bowers, K. D. & Stoll, D. A. (1992). Incidence of hyperpronation in the ACL injured knee: a clinical perspective. *Journal of Athletic Training, 27*(1), 58-62.

Beckman, S. M. & Buchanan, T. S. (1995). Ankle inversion injury and hypermobility: effect on hip and ankle muscle electromyography onset latency. *Archives of Physical and Medical Rehabilitation, 76*, 1138-1143.

Benesch, S., Pütz, W., Rosenbaum, D. & Becker, H. P. (2000). Reliability of peroneal reaction time measurements. *Clinical Biomechanics, 15*, 21-28.

Bennell, K. L. & Goldie, P. A. (1994). The differential effects of external ankle support on postural control. *Journal of Orthopaedic and Sports Physical Therapy, 20*(6), 287-295.

Bennell, K., Matheson, G., Meeuwisse, W. & Brukner, P. (1999). Risk factors for stress fractures. *Sports Medicine, 28*(2), 91-122.

Berg, C. L., Hart, J. M., Palmieri-Smith, R., Cross, K. M & Ingersoll, C. D. (2007). Cryotherapy does not affect peroneal reaction following sudden inversion. *Journal of Sport Rehabilitation, 16*, 285-294.

Birmingham, T. (2000). Test re-test reliability of lower extremity functional instability measures. *Clinical Journal of Sports Medicine, 10*, 264-268.

Bloomfield, J., Polman, R. & O'Donoghue, P. (2007). Physical demands of different positions in FA Premier League soccer. *Journal of Sports Science and Medicine, 6*, 63-70.

Bogey, R., Cerny, K. & Mohammed, O. (2003). Repeatability of wire and surface electrodes in gait. *American Journal of Physical Medicine and Rehabilitation, 82*, 338-344.

Bolgla, L. A. & Malone, T. R. (2004). Plantar fasciitis and the windlass mechanism: a biomechanical link to clinical practice. *Journal of Athletic Training, 39*(1), 77-82.

Bolgla, L. A. & Uhl, T. L. (2007). Reliability of electromyography normalisation methods for evaluating the hip musculature. *Journal of Electromyography and Kinesiology, 17*, 102-111.

Boyle, J. & Negus, V. (1998). Joint position sense in the recurrently sprained ankle. *Australian Journal of Physiotherapy, 44*(3), 159-163.

Branthwaite, H. R., Payton, C. J. & Chockalingam, N. (2004). The effect of simple insoles on three dimensional foot motion during normal walking. *Clinical Biomechanics, 19*, 972-977.

Brantingham J. W., Adams, K. J., Cooley J. R., Globe, D. & Globe, G. (2007). A single-blind pilot study to determine risk and association between navicular drop, calcaneal eversion, and low back pain. *Journal of Manipulative and Physiological Therapeutics 30*(5), 380-385.

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.

Briem, K., Eythörsdöttir, H., Magnúsdóttir, R. G., Pálmarsson, R., Rúnarsdöttir, T. & Sveinsson, T. (2011). Effects of kinesio tape compared with non-elastic sports tape and the untapped ankle during a sudden inversion perturbation in male athletes. *The Journal of Orthopaedic and Sports Physical Therapy, 41*(5), 328-335.

Brody, D. M. (1982). Techniques in the evaluation and treatment of the injured runner. *Orthopaedic Clinics of North America, 13*(3), 541-548.

Brooks, S. C., Potter, B. T. & Rainey, J. B. (1981). Treatment for partial tears of the lateral ligament of the ankle sprain. *British Medical Journal, 282*(6264), 606-607.

Browne, J. & O'Hare, N. (2000). A quality control procedure for force platforms. *Physiological Measurement, 21*, 515-524.

Brunt, D., Andersen, J. C., Huntsman, B., Reinhert, L. B., Thorell, A. C. & Sterling, J. C. (1991). Postural responses to lateral perturbation in healthy subjects and ankle sprain patients. *Medicine and Science in Sport and Exercise, 24*(2), 171-176.

Bruton, A., Conway, J. H. & Holgate, S. T. (2000). Reliability: what is it, and how is it measured? *Physiotherapy*, 86, 2, 94-99.

Buckley, J. G., Anand, V., Scally A. & Elliott, D. B. (2005). Does head extension and flexion increase postural instability in elderly subjects when visual information is kept constant? *Gait and Posture, 21*(1), 59-64.

Burden, A. (2007) Surface electromyography. In C. J. Payton & R. M. Bartlett (Eds.), *Biomechanical evaluation of movement in sport and exercise. The British Association of Sport and Exercise Sciences Guidelines*. (pp. 77-98). United Kingdom: Routledge.

Burns, P. R. & Lowery, N. (2011). Etiology, pathophysiology, and most common injuries of the lower extremity in the athlete. *Clinics in Podiatric Medicine and Surgery, 28*(1), 1-18.

Callaghan, M. J. (1997). Role of ankle taping and bracing in the athlete. *British Journal of Sports Medicine, 31*, 102-108.

Campanini, I., Merlo, A., Degola, P., Merletti, R., Vezzosi, G. & Farina, D. (2007). Effect of electrode location on EMG signal envelope in leg muscles during gait. *Journal of Electromyography and Kinesiology*, 17, 515-526.

Carcia, C. R., Drouin, J. M. & Houglum, P. A. (2006). The influence of a foot orthotic on lower extremity transverse plane kinematics in collegiate female athletes with pes planus. *Journal of Sports Science and Medicine, 5*, 646-655.

Catterall, C., Reilly, T., Atkinson, G. & Coldwells, A. (1993). Analysis of the work rates and heart rates of association football referees. *British Journal of Sports Medicine, 27*(3), 193-196.

Chan, K. M., Yuan, Y., Chien, P. & Tsang, G. (1993). Sports causing most injuries in Hong Kong. *British Journal of Sports Medicine, 27*(4), 263-267.

Chapman, J. P., Chapman, L. J. & Allen, J. J. (1987). The measurement of foot preference. *Neuropsychologia, 25*(3), 579-583.

Chen, C. L., Yeung, K. T., Wang, C. H., Chu, H. T. & Yeh, C. Y. (1999). Anterior ankle-foot orthosis effects on postural stability in hemiplegic patients. *Archives of Physical Medicine and Rehabilitation, 80*, 1587-1592.

Childs, R. A., Olson, B. A., McPoil, T. G. & Cornwall, M. W. (1996). The effect of three treatment techniques in reducing metatarsal head pressures during walking. *The Lower Extremity, 3,* 25-29.

Cobb, S. C., Tis, L. L., Johnson, B. F. & Higbie, E. J. (2004). The effect of forefoot varus on postural stability. *Journal of Orthopaedic and Sports Physical Therapy, 34*, 79-85.

Cordova, M. L., Bernard, L. W., Au, K, K., Demchak, T. J., Stone, M. B. & Sefton, J. M. (2010). Cryotherapy and ankle bracing effects on peroneus longus response during sudden inversion. *Journal of Electromyography and Kinesiology, 20*(2), 348-353.

Cordova, M. L., Cardona, C. V., Ingersoll, C. D. & Sandrey, M. A. (2000). Long term ankle brace use does not affect peroneus longus muscle latency during sudden inversion in normal subjects. *Journal of Athletic Training, 35*(4), 407-411.

Cornwall, M. W. & McPoil, T. G. (1997). Effect of foot orthotics on the initiation of plantar surface loading. *The Foot, 7*, 148-152.

Cote, K. P., Brunet, M. E., Gansneder, B. M. & Shultz, S. J. (2005). Effects of pronated and supinated foot postures on static and dynamic postural stability. *Journal of Athletic Training, 40*(1), 41-46.

Cowan, D. N., Robinson, J. R., Jones, B. H., Polly, D. W. & Hudson Berrey, B. (1994). Consistency of visual assessments of arch height among clinicians. *Foot and Ankle International, 15*(4), 213-217.

D'Août, K., Pataky T. C., De Clercq, D. & Aerts, P. (2009). The effects of habitual footwear use: foot shape and function in native barefoot walkers. *Footwear Science, 1*(2), 81-94.

Dahle, L. K., Mueller, M., Delitto, A. & Diamond, J. E. (1991). Visual assessment of foot type and relationship of foot type to lower extremity injury. *Journal of Orthopaedic and Sports Physical Therapy, 14(*2), 70-74.

Dawe, E. J. C. & Davis, J. (2011). Anatomy and biomechanics of the foot and ankle. *Journal of Orthopaedic Trauma, 25*(4), 279-286.

Dayakidis, M. K. & Boudolos, K. (2006). Ground reaction force data in functional ankle instability during two cutting movement. *Clinical Biomechanics, 21*, 405-411.

De Luca, C. J. (1997). The use of surface electromyography in biomechanics. *Journal of Applied Biomechanics, 13*(2), 135-163.

Del Rossi, G., Fiolkowski, P., Horodyski, M. B., Bishop, M. & Trimble, M. (2004). For how long do temporary techniques maintain the height of the medial arch? *Physical Therapy in Sport, 5*(2), 84-89.

Delahunt, E., O'Driscoll, J. & Moran, K. (2009). Effects of taping and exercise on ankle joint movement in subjects with chronic ankle instability: a preliminary study. *Archives of Physical Medicine and Rehabilitation, 90,* 1418-1422.

Denegar, C. R. & Ball, D. W. (1993). Assessing reliability and precision of measurement: an introduction to intraclass correlation and standard error of measurement. *Journal of Sport Rehabilitation, 2*, 35-42.

Denyer, J. & Mitchell A. (2011). The effects of short term anti-pronation techniques on peak plantar pressures during dynamic movement, British Association of Sport and Exercise Sciences Conference presentation, University of Essex, 6-8th September.

Di Fabio, R. P. (1987). Reliability of computerised surface electromyography for determining the onset of muscle activity. *Physical Therapy, 67*(1), 43-48.

Doyle, R. J., Hsiao-Weclsler, E. T., Ragan, B. G. & Rosengren, K. S. (2007). Generalizability of centre of pressure measures of quiet standing. *Gait and Posture, 25*(2), 166-171.

Dunn, J. E., Link, C. L., Felson, D. T., Crincoli, M. G., Keysor, J. J. & McKinlay, J. B. (2004). Prevalance of foot and ankle conditions in a multi-ethnic community sample of older adults. *American Journal of Epidemiology, 159*(5), 491-498.

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*(2), 73-77.

Eechaute, C., Vaes, P., Duquet, W. & Van Gheluwe, B. (2007). Test-retest reliability of sudden ankle inversion measurements in subjects with healthy ankle joints. *Journal of Athletic Training, 42*(1), 60-65.

Eils, E. & Rosenbaum, D. (2001). A multi-station proprioceptive exercise program in patients with ankle instability. *Medicine and Science in Sports and Exercise, 33*(12), 1991-1998.

Engström, B., Johansson, C. & Törnkvist, H. (1991). Soccer injuries among elite female players. *The American Journal of Sports Medicine, 19*(4), 372-375.

Farina, D., Merletti, R., Nazzaro, M. & Caruso, I. (2001). The effect of joint angle on EMG variables in leg and thigh muscles. *Engineering in Medicine and Biology, 20*(6), 62-71.

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.

Feuerbach, J. W. & Grabiner, M. D. (1993). Effect of the aircast on unilateral postural control: amplitude and frequency variables. *Journal of Sport and Physical Therapy,* 7(3), 149-154.

Fleet, K., Galen, S. & Moore, C. (2009). Duration of strength retention of ankle taping during activities of daily living. *Injury, 40*, 333-336.

Fong, D. T. P., Chan, Y. Y., Mok, K. M., Yung, P. S. H. & Chan, K. M. (2009). Understanding acute ankle ligamentous sprain injury in sports. *Sports Medicine, Arthroscopy, Rehabilitation, Therapy and Technology*, 1, 1-14.

Franco, A. H. (1987). Pes cavus and pes planus, analysis and treatment. *Physical Therapy, 67*(5), 688-694.

Franettovich, M., Chapman, A. & Vicenzino, B. (2008). Tape that increases medial longitudinal arch height also reduces leg muscle activity: a preliminary study. *Medicine and Science in Sports and Exercise, 40*(4), 593-600.

Frankeny, J. R., Jewett, D. L., Hanks, G. A. & Sebastianelli, W. J. (1993). A comparison of ankle taping methods. *Clinical Journal of Sports Medicine, 3*, 20-25.

Freeman, M. A. R., Dean, M. R. E. & Hanham, I. W. F. (1965). The etiology and prevention of functional instability of the foot. *The Journal of Bone and Joint Surgery, 47B*(4), 678–685.

Fumich, R. M., Ellison, A. E., Guerin, G. J. & Grace P. D. (1981). The measured effect of taping on combined foot and ankle motion before and after exercise. *The American Journal of Sports Medicine, 9*(3), 165-170.

Garrick, J. G. (1977). The frequency of injury, mechanism of injury and epidemiology of ankle sprains. *The American Journal of Sports Medicine, 5*(6), 241-242.

Gerdle, B., Karlsson, S., Day, S. & Djupsjöbacka, M. (1999). Acquisition, processing and analysis of the surface electromyogram. In U. Windhorst & H. Johansson (Eds.), *Modern techniques in neuroscience research*. (pp. 705-755). New York: Springer.

Gibney, V. P. (1895). Sprained ankle, a treatment that involves no loss of time, requires no crutches, and is not attended with an ultimate impairment of function. *The New York Medical Journal, 61*,193–197.

Giza, E., Fuller, C., Junge, A. & Dvorak, J. (2003). Mechanisms of foot and ankle injuries in soccer. *The American Journal of Sports Medicine, 31*(4), 550-554.

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.*

Gould, G. M. (1908). *Right-handedness and left-handedness*. Philadelphia, PA: Lippincott.

Gray, H. (1918). *Anatomy of the Human Body.* (20th ed.). Philadelphia: Lea & Febiger. Online Edition (2000); http://www.bartleby.com/107.

Greene, T. A. & Hillman, S. K. (1990). Comparison of support provided by a semi-rigid orthosis and adhesive ankle taping before, during, and after exercise. *The American Journal of Sports Medicine, 18*(5), 498-506.

Gribble, P. A. & Hertel, J. (2004). Effect of hip and ankle muscle fatigue on unipedal postural control. *Journal of Electromyography and Kinesiology, 14*, 641-646.

Grundy, M. Tosh, P. A., McLeish, R. D. & Smidt, L. (1975). An investigation of the centre of pressure under the foot while walking. *The Journal of Bone and Joint Surgery, 57*(1), 98-103.

Gurney, J. K., Kersting, U. G. & Rosenbaum, D. (2009). Dynamic foot function and morphology in elite rugby league athletes of different ethnicity. *Applied Ergonomics, 40*, 554-559.

Gurney, J. K., Kuch, C., Rosenbaum, D. & Kersting, U. G. (2012). The Māori foot exhibits differences in plantar loading and midfoot morphology to the Caucasian foot. *Gait and Posture, 36*(1), 157-159.

Guskiewicz, K. M. & Perrin, D. H. (1996). Research and clinical applications of assessing balance. *Journal of Sport Rehabilitation, 5*, 45-63.

Gutierrez, G. M. & Kaminski, T. (2010), A novel dynamic ankle-supinating device. *Journal of Applied Biomechanics, 26*, 114-121.

Hadley, A., Griffiths, S., Griffiths, L. & Vicenzino, B. (1999). Antipronation taping and temporary orthoses. Effects on tibial rotation position after exercise. *Journal of the American Podiatric Medical Association, 89*(3), 118-123.

Hageman, P. A., Leibowitz, M. & Blanke, D. (1995). Age and gender effects on postural control measures. *Archives of Physical Medicine and Rehabilitation, 76*, 961-965.

Halseth, T., McChesney, J. W., DeBeliso, M. Vaughn, R. & Lien, J. (2004). The effects of kinesio taping on proprioception at the ankle. *Journal of Sports Science and Medicine, 3*, 1-7.

Hargrave, M. D., Carcia, C. R., Gansneder, B. M. and Shultz, S. J. (2003). Subtalar pronation does not influence impact forces or rate of loading during a single-leg landing. *Journal of Athletic Training, 38*(1), 18-23.

Heaton, J. (1998). Secondary Analysis of Qualitative Data: *Social Research Update Issue 22*. Retrieved January, 2012, from www.sru.soc.surrey.ac.uk/SRU22.html.

Hermens, H. J., Freiks, B., Disselhorst-Klug, C. & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology, 10*, 361-374.

Hertel, J. (2000). Functional instability following lateral ankle sprain. *Sports Medicine, 29*(5), 361-371.

Hertel, J. (2002). Functional anatomy, pathomechanics, and pathophysiology of lateral ankle instability. *Journal of Athletic Training, 37*(4), 364-375.

Hertel, J. N., Guskiewicz, K. M., Kahler, D. M., & Perrin, D. H. (1996). Effect of lateral ankle joint anaesthesia on centre of balance, postural sway and joint position sense. *Journal of Sport Rehabilitation, 2*, 111-119.

Hertel, J., Gay, M. R., & Denegar, C. R. (2002). Differences in postural control during single-leg stance among healthy individuals with different foot types. *Journal of Athletic Training, 37*(2), 129-132.

Hills, A. P., Hennig, E. M., McDonald, M. & Bar-Or, O. (2001). Plantar pressure differences between obese and non-obese adults: a biomechanical analysis. *International Journal of Obesity, 25*(11), 1674-1679.

Hintermann, B. & Nigg, B. M. (1998). Pronation in runners: implications for injuries. *Sports Medicine, 26*(3), 169-176.

Hodges, P. W. & Bui, B. H. (1996). A comparison of computer-based methods for the determination of onset of muscle contraction using electromyography. *Electroencephalography and Clinical Neurophysiology, 101*, 511-519.

Hoffman, M., Schrader, J., Applegate, T. & Koceja, D. (1998). Unilateral postural control of the functionally dominant and non-dominant extremities of healthy subjects. *Journal of Athletic Training, 33*(4), 319-322.

Holmes, C. F., Wilcox, D. & Fletcher, J. P. (2002) Effect of a modified, low-Dye medial longitudinal arch taping procedure on the subtalar joint neutral position before and after light exercise. *Journal of Orthopedic Sports Physical Therapy, 32*(5), 194-201.

Hootman, J. M., Dick, R. & Agel, J. (2007). Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives. *Journal of Athletic Training, 42*(2), 311-319.

Hopkins, J. T., Brown, T. N., Christensen, L. & Palmieri-Smith, R. M. (2009). Deficits in peroneal latency and electromechanical delay in patients with functional ankle instability. *Journal of Orthopaedic Research, 27*(12), 1541-1546.

Hopkins, J. T., Hunter, I. & McLoda, T. (2006). Effects of ankle joint cooling on peroneal short latency response. *Journal of Sports Science and Medicine, 5*, 333-339.

Hopkins, J. T., McLoda, T. & McCaw, S. (2007). Muscle activation following sudden ankle inversion during standing and walking. *European Journal of Applied Physiology, 99*, 371-378.

Horak, F. B. (1987). Clinical measurement of postural control in adults. *Physical Therapy, 67*(12), 1881-1885.

Hufschmidt, A., Dichgans, J., Mauritz, K. H. & Hufschmidt, M. (1980). Some methods and parameters of body sway quantification and their neurological applications. *European Archives of Psychiatry and Clinical Neuroscience, 228*(2), 135-150.

Hunt, A. E. & Smith, R. M. (2004). Mechanics and control of the flat versus normal foot during the stance phase of walking. *Clinical Biomechanics, 19*, 391-397.

Hunt, G. C., Sneed, T., Hamann, H. & Chisam, S. (2004). Biomechanical and histological considerations for development of plantar fasciitis and evaluation of arch taping as a treatment option to control associated plantar heel pain: a single-subject design. *The Foot, 14*, 147-153.

Hunt, G. C., Stowell, T., Alnwick, G. M. & Evans, S. (2007). Arch taping as a symptomatic treatment in patients with Sever's disease: A multiple case series. *The Foot, 17(4), 178-183.*

Hunter, M. C. & Hoffman, M. A. (2001). Postural control: visual and cognitive manipulations. *Gait and Posture, 13*, 41-48.

Igbigbi, P. S. & Msamati, B. C. (2002). The footprint ratio as a predictor of pes planus: a study of indigenous Malawians. *Journal of Foot and Ankle Surgery, 41*(6), 394-397.

Janin, M. & Dupui, P. (2009). The effects of unilateral medial arch support stimulation on plantar pressure and centre of pressure adjustment in young gymnasts. *Neuroscience Letters, 461*, 245-248.

Johnson, C. H. & Christensen, J. C. (1999). Biomechanics of the first ray part 1: the effects of peroneus longus function: a three-dimensional kinematic study on a cadaver model. *Journal of Foot and Ankle Surgery, 38*(5), 313-321.

Jones, A. M. & Doust, J. H. (1996). A 1% treadmill grade most accurately reflects the energetic cost of outdoor running. *Journal of Sports Sciences, 14*, 321-327.

Kaminski, T. W. & Gerlach, T. M. (2001). The effect of tape and neoprene ankle supports on ankle joint position sense. *Physical Therapy in Sport, 2*, 132-140.

Kannus, P. & Renström, P. (1991). Treatment for acute tears of the lateral ligaments of the ankle. *The Journal of Bone and Joint Surgery, 73*(2), 305-312.

Karatsolis, K., Nikolopoulos, C. S., Papadopoulos, E. S., Vagenas, G. T., Terzis, E. & Athanasopoulos, S. (2009). Eversion and inversion muscle group peak torque in hyperpronated and normal individuals. *The Foot, 19*(1), 29-35.

Karlsson, J. & Andreasson, G. O. (1992). The effect of external ankle support in chronic lateral ankle joint instability. *The American Journal of Sports Medicine, 20(*3), 257-261.

Karlsson, J., Swärd, L. & Andréasson, G.O. (1993). The effect of taping on ankle stability: practical implications, *Sports Medicine, 16*(3), 210-215.

Kaufman, K. R., Brodine, S. K., Shaffer, R. A., Johnson, C. W. & Cullison, T. R. (1999). The effect of foot structure and range of motion on musculoskeletal overuse injuries. *The American Journal of Sports Medicine, 27*(5), 585-593.

Kaye, R. A. & Jahss, M. H. (1991). Tibialis posterior: a review of anatomy and biomechanics in relation to support of the medial longitudinal arch. *Foot and Ankle International, 11*(4), 244-247.

Kell, C. & van Duursen, R. W. M. (2005). Posture and balance. In M. Trew, & T. Everett (Eds.), *Human movement: an introductory text*. (pp. 231-253). United Kingdom: Elsevier.

Kendall, F. P., McCreary, E. K. & Provance, P. G. (1993). *Muscles, Testing and Function.* (4th ed.). United States of American: Williams & Wilkins.

Kent, M. (2006). *Oxford Dictionary of Sports Science and Medicine*. (3rd ed.). Oxford: Oxford University Press.

Knutson, L. M., Soderberg, G. L., Ballantyne, B. T. & Clarke, W. R. (1994). A study of various normalisation procedures for within day electromyographic data. *Journal of Electromyography and Kinesiology, 4*(1), 47-59.

Kollmitzer, J., Ebenbichler, G. R & Kopf, A. (1999). Reliability of surface electromyography measurements. *Clinical Neuropathy, 110*, 725-734.

Konrad, P. (2005). The ABC of EMG; a practical introduction to kinesiological electromyography, 1, Noraxon Inc.

Konradsen, L. & Ravn, J. B. (1991). Prolonged peroneal reaction time in ankle instability. *International Journal of Sports Medicine, 12*(3), 290-292.

Konradsen, L. (2002). Sensorimotor control of the uninjured and injured human ankle. *Journal of Electromyography and Kinesiology, 12*, 199-203.

Konradsen, L., Ravn, J. B & Sorensen, A. I. (1993). Proprioception at the ankle: the effect of anaesthetic blockade of ligament receptors. *The Journal of Bone and Joint Surgery, 75B*(3), 433-436.

Konradsen, L., Voigt, M. & Hojsgaard, C. (1997). Ankle inversion injuries, the role of the dynamic defence mechanism. *The American Journal of Sports Medicine, 25*(1), 54-58.

Korpelainen, R., Orava, S., Karpakka, J., Siira, P. & Hulkko, A. (2001). Risk factors for recurrent stress fractures in athletes. *The American Journal of Sports Medicine, 29*(3), 304-310.

Kumai, T., Takakura, Y., Rufai, A., Milz, S. & Benjamin, M. (2002). The functional anatomy of the human anterior talofibular ligament in relation to ankle sprains. *Journal of Anatomy, 200(5), 457-465.*

Landorf, K. B., Radford, J. A., Keenan, A. M. & Redmind, A. C. (2005). Effectiveness of low-dye taping for the short term management of plantar fasciitis. *Journal of the American Podiatric Medical Association, 95*(6), 525-529.

Langdon, J.H., Bruckner, J. & Baker, H.H., (1991). Pedal mechanics and bipedalism in early hominids. In B. Senut, & Y. Coppens (Eds.), *Origine(s) de le Bipedie chez les Hominides.* (pp. 159-167). Paris: Editions du CNRS.

Lange, B., Chipchase, L. & Evans, A. (2004). The effect of low-Dye taping on plantar pressures, during gait on subjects with navicular drop exceeding 10mm. *Journal of Orthopedic and Sports Physical Therapy, 34*(4), 201-209.

Laughman, R. K., Carr, T. A., Chao, E. Y., Youdas, J. W. & Sim, F. H. (1980). Three-dimensional kinematics of the taped ankle before and after exercise. *The American Journal of Sports Medicine, 8*(6), 425-431.

Laurens, J., Awai, L., Bockisch, C. J., Hegemann, S., van Hedal, H. J. A., Dietz, V. & Straumann, D. (2010). Visual contribution to postural stability: interaction between target fixation or tracking and static or dynamic large-field stimulus. *Gait and Posture, 31*(1), 37-41.

Le Clair, K. & Riach, C. (1996). Postural stability measures: what to measure and for how long. *Clinical Biomechanics, 11*(3), 176-178.

Leanderson, J., Ekstam, S. & Salomonsson, C. (1996). Taping of the ankle – the effect on postural sway during perturbation before and after a training session. *Knee Surgery, Sports Traumatology, Arthroscopy, 4*(1), 53-56.

Leavey, V. J., Sandrey, M. A. & Dahmer, G. (2010). Comparative effects of 6-week balance, gluteus medius strength, and combined programs on dynamic postural control. *Journal of Sports Rehabilitation, 19*(3), 268-287.

Ledoux, W. R. & Hillstrom, H. J. (2002). The distributed plantar vertical force of neutrally aligned and pes planus feet. *Gait and Posture*, *15*, 1-9.

Ledoux, W. R., Shofer, J. B., Ahroni, J. H., Smith, D. G., Sangeorzan, B. J. & Boyko, E. J. (2003). Biomechanical differences among pes cavus, neutrally aligned, and pes planus feet in subjects with diabetes. *Foot and Ankle International, 24*(11), 845-850

Lee, A. (1981). Methods of impact absorption when landing from a jump. *Engineering in Medicine, 10*, 207-211.

Lephart, S. M., Pincivero, D. M., Giraido, J. L. & Fu, F. H. (1997). The role of proprioception in the management and rehabilitation of athletic injuries. *The American Journal of Sports Medicine, 25*, 130-137.

Linford, C. W., Hopkins, J. T., Schulthies, S. S., Freland B. Draper, D. O. & Hunter, I. (2006). Effects of neuromuscular training on reaction time and electromechanical delay of the peroneus longus muscle. *Archives of Physical Medicine and Rehabilitation. 87*, 395-401.

Lohkamp, A., Craven, S., Walker-Johnson, C. & Greig, M. (2009). The influence of ankle taping on changes in postural stability during soccer-specific activity. *Journal of Sport Rehabilitation, 18*, 482-492.

Lohrer, H., Alt, W. & Gollhofer, A. (1999). Neuromuscular properties and functional aspects of taped ankles. *The American Journal of Sports Medicine, 27*(1), 69-75.

Loos, T. & Boelens, P. (1984). The effect of ankle-tape on lower limb muscle activity. *International Journal of Sports Medicine, 5*, 45-46.

Louwerens, J. W. K., van Linge, B., de Klerk, L. W. L., Mulder, P. G. H. & Snijders, C. J. (1995). A quantified electromyographic study of 10 controls and 25 patients with chronic ankle instability. *Acta Orthopaedica, 66*(6), 517-523.

Lutter, L. D. (1980). Foot-related knee problems in the long distance runner. *Foot and Ankle International, 1*(2), 112-116.

Lynch, S. A., Eklund, U., Gottlieb, D., Renström, P. A. F. H. & Beynnon, B. (1996). Electromyographic latency changes in the ankle musculature during inversion moments. *The American Journal of Sports Medicine, 24*(3), 363-369.

Macgregor, K., Gerlach, S., Mellor, R. & Hodges, P. W. (2005). Cutaneous stimulation from patella tape causes a differential increase in vasti muscle activity in people with patellofemoral pain. *Journal of Orthopaedic Research, 23(2)*, 351-358.

Madigan, M. L., & Pidcoe, P. E. (2003). Changes in landing biomechanics during a fatiguing landing activity. *Journal of Electromyography and Kinesiology, 13*, 491-498.

Maffulli, N., Longo, U. G., Petrillo, S. & Denaro, V. (2012). Lateral ankle instability, *Orthopedics and Trauma, 26*(1), 20-24.

Manfroy, P. P., Ashton-Miller, J. A. & Wojtys, E. M. (1997). The effect of exercise, prewrap and athletic tape on the maximal active and passive ankle resistance to ankle inversion. *The American Journal of Sports Medicine, 25*(2), 156-163.

Manoli, A. & Graham, B. (2005). The subtle cavus foot, “the underpronator,” a review. *Foot and Ankle International, 26*(3), 256-263.

Martin, N. & Harter, R. A. (1993). Comparison of inversion restraint provided by ankle prophylactic devices before and after exercise. *Journal of Athletic Training, 28*(4), 324-329.

Martini, F. H. (2006). *Fundamentals of Anatomy & Physiology*. (7th ed.). United States of America: Pearson Education Inc.

Maslen, B. A. & Ackland, T. R. (1994). Radiographic study of skin displacement errors in the foot and ankle during standing. *Clinical Biomechanics, 9*, 291-296.

Matsusaka, N., Yokoyama, S., Tsurusaki, T., Inokuchi, S. & Okita, M. (2001). Effect of ankle disk training combined with tactile stimulation to the leg and foot on functional instability of the ankle. *The American Orthopaedic Society for Sports Medicine, 29*(1), 25-30.

McCluskey, G. M., Blackburn, T. A. & Lewis, T. (1976). A treatment for ankle sprains. *The American Journal of Sports Medicine, 4*(4), 158-161.

McDowell, I. (2006). *Measuring Health: A Guide to Rating Scales and Questionnaires*. (3rd ed.). (pp. 39-41). Oxford University Press.

McKenzie, D. C., Clement, D. B. & Taunton, J. E. (1985). Running shoes, orthotics and injuries. *Journal of Sports Medicine, 2*(5), 334-347.

McPoil, T. G. & Cornwall, M. W. (1996). Relationship between three static angles of rearfoot and the pattern of rearfoot motion during walking. *Journal of Orthopaedic and Sports Physical Therapy, 23*(6), 370-375.

McPoil, T. G., Cornwall, M. W., Medoff, L., Vicenzino, B., Forsberg, K. & Hilz, D. (2008). Arch height change during sit-to-stand: an alternative for the navicular drop test. *Journal of Foot and Ankle Research, 1,* 3.

Meana, M., Alegre, L. M., Elvira, J. L. L. & Aguado, X. (2008). Kinematics of ankle taping after a training session. *International Journal of Sports Medicine, 29*(1), 70-76.

Mei-Dan, O. Kahn, G., Zeev, A., Rubin, A., Constantini, N., Even, A. (2005). The medial longitudinal arch as a possible risk factor for ankle sprains. *Foot and Ankle International, 26*(2), 180-183.

Menadue, C., Raymond, J., Kilbreath, S. L., Refshauge, K. M. & Adams, R. (2006). Reliability of two goniometric methods of measuring active inversion and eversion range of motion at the ankle. Musculoskeletal Disorders, 7, 60.

Menz, H. B. (1998). Alternative techniques for the clinical assessment of foot pronation. *Journal of American Podiatric Medicine Association, 88*(3), 119-129.

Merletti, R. (1996). Standards for reporting EMG data. *Journal of Electromyography and Kinesiology, 6*(4), III-IV.

Mesin, L., Merletti, R. & Rainoldi, A. (2009). Surface EMG: the issue of electrode location. *Journal of Electromyography and Kinesiology, 19*, 719-726.

Micheli, L. & Fehlandt, A. F. (1992). Overuse injuries to tendons and apophyses in children and adolescents. *Clinics in Sports Medicine, 11*(4), 713-726.

Michelson, J. D., Durant, D. M. & McFarland E. (2002). The injury risk associated with pes planus in athletes. *Foot Ankle International, 23*(7), 629-633.

Mitchell, A., Dyson, R., Hale, T. & Abraham, C. (2008a). Biomechanics of ankle instability. part 1: reaction time to simulated ankle sprain. *Medicine and Science in Sports and Exercise, 40*(8), 1515-1521.

Mitchell, A., Dyson, R., Hale, T. & Abraham, C. (2008b). Biomechanics of ankle instability. part 2: postural sway – reaction time relationship. *Medicine and Science in Sports and Exercise, 40*(8), 1522-1528.

Moghadam, M., Ashayeri,H., Salavati, M., Sarafzadeh, J., Taghipoor, K. D., Saeedi, A. et al. (2011). Reliability of centre of pressure measures of postural stability in healthy older adults: effects of postural task difficulty and cognitive load. *Gait and Posture, 33*(4), 651-655.

Morey-Klapsing, G., Arampatzis, A. & Brüggemann, G. P. (2004). Choosing EMG parameters: comparison of different onset determination algorithms and EMG integrals in a joint stability study. *Clinical Biomechanics, 19*, 196-201.

Morrissey, D. (2000). Proprioceptive shoulder taping. Journal of Body Work and Movement Therapies, 4(3), 189-194.

Mueller, M. J., Host, J. V. & Norton, B. J. (1993). Navicular drop as a composite measure of excessive pronation. *Journal of the American Podiatric Medical Association, 83*(4), 198-202.

Munro, B. (2000). *Statistical Methods for Health Care Research,* (4th ed.). Philadelphia: JB Lippincott.

Nachbauer, W. & Nigg, B. M. (1992). Effects of arch height of the foot on ground reaction forces in running. *Medicine and Science in Sport and Exercise, 24*(11), 1264-1269.

Nashner, L. M., Block, F. O. & Wall, C. (1982). Adaptation to altered support and visual conditions during stance: patients with vestibular deficits. Journal of Neuroscience, 2, 536-544.

Nguyen, A. D. & Shultz, S. J. (2009). Identifying relationships among lower extremity alignment characteristics. *Journal of Athletic Training, 44*(5), 511-518.

Nicolopoulos, C. S., Scott, B. W. & Giannoudis, P. V. (2000). Biomechanical basis of foot orthotic prescription. *Current Orthopedics, 14,* 464-469.

Nielsen, R. G., Rathleff, M. S., Simonsen, O. H. & Langberg, H. (2009). Determination of normal values for navicular drop during walking: a new model correcting for foot length and gender. *Journal of Foot and Ankle Research, 2*, 12.

Nishikawa, T., Kurosaka, M., Yoshiya, S., Lundin, T. M. & Grabiner, M. D. (2002). Effects of prophylactic ankle supports on pronation during gait. *International Orthopaedics, 26*, 381-385.

Norcross, M. F., Blackburn, J. T. & Goerger, B. M. (2009). Reliability and interpretation of single leg stance and maximum voluntary isometric contraction methods of electromyography normalization. *Journal of Electromyography and Kinesiology*, *20*(3), 420-425.

O’Sullivan, K., Kennedy, N., O’Neill, E. & Mhainin, U. N. (2008). The effect of low-dye taping on rearfoot motion and plantar pressure during the stance phase of gait. *BMC Musculoskeletal Disorders, 9*, 111.

Osborne, M. D., Chou, L. S., Laskoski, E. R., Smith, J. & Kaufman, K. R., The effect of ankle disk training on muscle reaction time in subjects with a history of ankle sprain. *The American Journal of Sports Medicine, 29*(5), 627-632.

Ozer, D., Senbursa, G., Baltaci, G. & Hayran, M. (2009). The effect on neuromuscular stability, performance, multi-joint coordination and proprioception of barefoot, taping or preventative bracing. *The Foot, 19*(4), 205-210.

Palmieri, R. M., Ingersoll, C. D., Stone, M. B. & Krause, B. A. (2002). Centre-of-pressure parameters used in the assessment of postural control. *Journal of Sports Rehabilitation, 11*, 51-66.

Papadopoulos, E. S., Nicolopoulos, C., Baldoukas, A., Anderson, E. G. & Athanasopoulos, S. (2005). The effect of different ankle brace-skin interface application pressures on the electromyographic peroneus longus reaction time. *The Foot, 14*, 175-179.

Paris, D. L., Vardaxis, V. & Kokkaliaris, J. (1995). Ankle ranges of motion during extended activity periods while taped and braced. *Journal of Athletic Training, 30*(3), 223-228.

Pavol, M. J. (2005). Detecting and understanding differences in postural sway. *Journal of Neurophysiology, 23*, 20-21.

Pease, W. S. & Elinski, A. (2003). Surface and wire electromyographic: recording during fatiguing exercise. *Electromyography and Clinical Neurophysiology, 43*(5), 267-271.

Pederson, T. S., Ricard, M. D., Merrill, G., Schulthies, S. S. & Allsen, P. E. (1997). The effects of spatting and ankle taping on inversion before and after exercise. *Journal of Athletic Training, 32*(1), 29-33.

Perrin, D. H. (2005). *Athletic Taping and Bracing*. (2nd ed.). Leeds: Human Kinetics.

Peterka, R. J. & Loughlin, P. J. (2004). Dynamic regulation of sensorimotor integration in human postural control. *Journal of Neurophysiology, 91*, 410-423.

Peterka, R. J. (2002). Sensorimotor integration in human postural control*. Journal of Neurophysiology, 88*, 1097-1118.

Peters, M. (1988). Footedness: asymmetries in foot preference and skill and neurophysiological assessment of foot movement. *Psychological Bulletin, 103*(2), 179-192.

Picciano, A. M., Rowlands, M. S. & Worrell, T. (1993). Reliability of open and closed kinetic chain subtalar joint neutral positions and navicular drop test. *Journal of Orthopaedic and Sports Physical Therapy, 18*(4), 553-558.

Pope, M. H., Renström, P., Donnermeyer, D. & Morgenstern, S. (1987). A comparison of ankle taping methods. *Medicine and Science in Sport and Exercise, 19*(2), 143-147.

Previc, F. H. (1991). A general theory concerning the prenatal origins of cerebral lateralisation in humans. *Psychological Review, 98*(3), 299-334.

Radford, J. A., Landorf, K. B., Buchbinder, R. & Cook, C. (2006). Effectiveness of low-Dye taping for the short-term treatment of plantar heel pain: a randomized trial. *BMC Musculoskeletal Disorders, 7*, 64.

Rainoldi, A., Melchiorri, G. & Caruso, I. (2004). A method for positioning electrodes during surface EMG recordings in lower limb muscles. *Journal of Neuroscience Methods, 134*, 37-43.

Rarick, G. L., Bigley, G., Karst, R. & Malina, R. M. (1962). The measurable support of the ankle joint by conventional methods of taping. *Journal of Bone and Joint Surgery, 44*(6), 1183-1190.

Rash, G. S. (2003). *Electromyography Fundamentals; Gait and Clinical Movement Analysis Society*. Retrieved January, 2010, from www.gcmas.org.

Razeghi, M. & Batt, M. E. (2002). Foot type classification: a critical review of current methods. *Gait and Posture, 15*, 282-291.

Refshauge, K. M., Kilbreath, S. L. & Raymond, J. (2000). The effect of recurrent ankle inversion sprain and taping on proprioception at the ankle. *Medicine and Science in Sport and Exercise, 32*(1), 10-15.

Refshauge, K. M., Raymond, J., Kilbreath, S. L., Pengal, L. & Heijnen, I. (2009). The effect of ankle taping on detection of inversion-eversion movements in participants with recurrent ankle sprain. *The American Journal of Sports Medicine, 37*(2), 371-375.

Ricard, M. D., Schulties, S. S. & Saret, J. J. (2000), Effects of high-top and low-top shoes on ankle inversion. *Journal of Athletic Training, 35*(1), 38-43.

Ricard, M. D., Sherwood, S. M., Schulthies, S. S. & Knight, K. L. (2000). Effects of tape and exercise on dynamic ankle inversion. *Journal of Athletic Training, 35*(1), 31-37.

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*(4), 240-251.

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., Schmitz, R. J., Gale, M. G., & McCaw, S. T. (2002). Effect of ankle taping and bracing on vertical ground reaction forces during drop landings before and after treadmill jogging. *Journal of Orthopaedic and Sports Physical Therapy, 32*, 628-635.

Robbins, S., Waked, E. & Rappelt, R. (1995). Ankle taping improves proprioception before and after exercise in young men. *British Journal of Sports Medicine, 29*(4), 242-247.

Robertson, D. G. E., Caldwell, G. E., Hamill, J., Kamen, G. & Whittlesey, S. N. (2004). *Research methods in biomechanics*. (1st ed.). United States of America: Human Kinetics.

Rocchi, M. B. L., Sisti, D., Ditroilo, M., Calavalle, A. & Panebianco, R. (2005). The misuse of the confidence ellipse in evaluating statokinesigram. *Italian Journal of Sport Science, 12*, 169-172.

Rodgers, M. M. (1988). Dynamic biomechanics of the normal foot and ankle during walking and running. *Physical Therapy, 90*(10), 1822-1830.

Root, M. L. (1973). Biomechanical examination of the foot. *Journal of American Podiatry Association, 63*(1), 28-29.

Rosenbaum, D. & Becker, H. P. (1997). Plantar pressure distribution measurements. Technical background and clinical applications. *The Journal of Foot and Ankle Surgery, 3*, 1-14.

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*(6), 332-338.

Rothermel, S. A., Hale, S. A. & Denegar, C. R. (2004). Effect of active foot positioning on the outcome of a balance training program. *Physical Therapy in Sport, 5*,(2), 98-103.

Russo, S. J. & Chipchase, L. S. (2001). The effect of low-Dye taping on peak plantar pressures of normal feet during gait. *Australian Journal of Physiotherapy, 47*(4), 239-244.

Sacco, I. C. N., Gomes, A. A., Otuzi, M. E., Pripas, D. & Onodera, A. N. (2009). A method for better positioning bipolar electrodes for lower limb EMG recordings during dynamic contractions. *Journal of Neuroscience Methods, 180*, 133-137.

Sachithanandam, V. & Joseph, B. (1995). The influence of footwear on the prevalence of the flat foot. *The Journal of Bone and Joint Surgery*, *77*, 254-257.

Sadeghi, H., Allard, P., Prince, F. & Labelle H. (2000). Symmetry and limb dominance in able-bodied gait: a review. *Gait and Posture, 12*(1), 34-35.

Safran, M, R., Benedetti, R. S., Bartolozzi, A. R. & Mandelbaum, B. R. (1999). Lateral ankle sprains: a comprehensive review part 1: etiology, pathoanatomy, histopathogenesis and diagnosis. *Medicine and Science in Sports and Exercise, 31*(7), S429-S437.

Salavati, M., Hadian , M. R., Mazaheri, M., Negahban, H., Ebrahimi, I., Talebian, S., et al. (2008). Test-retest reliability of centre of pressure measures of postural stability during quiet standing in a group with musculoskeletal disorders consisting of back pain, anterior cruciate ligament injury and functional ankle instability. *Gait and Posture, 29*, 460-464.

Saltzman, C. L., Nawoczenski, D. A. & Talbot, K. D. (1995). Measurement of the medial longitudinal arch. *Archives of Physical Medicine and Rehabilitation, 76*, 45-49.

Sawkins, K., Refshauge, K., Kilbreath, S. & Raymond, J. (2007). The placebo effect of ankle taping in ankle instability. *Medicine and Science in Sports and Exercise, 39*(5), 781-787.

Schmidt, R. A. & Lee, T. D. (2005). *Motor Control and Learning; A Behavioural Emphasis*. (4th ed.). United States of America: Human Kinetics.

Schneider, P. L., Crouter, S. E. & Bassett jr. D. R. (2004). Pedometer measures of free-living physical activity: comparison of 13 models. *Medicine and Science in Sport and Exercise, 36*(2), 331–335.

Scranton, P. E., Pedegana, L. R. & Whitesel, J. P. (1982). Alterations in support phase forces using supportive devices. *The American Journal of Sports Medicine, 10*(1), 6-11.

Sell, K. E., Verity, T. M., Worrell, T. W., Pease, B. J. & Wigglesworth, J. (1994). Two measurement techniques for assessing subtalar joint position: a reliability study. *Journal of Orthopaedic and Sports Physical Therapy, 19*(3), 162-167.

Sell, T. (2012). An examination, correlation and comparison of static and dynamic measures of postural stability in healthy, physically active adults. *Physical Therapy in Sport, 13*(2), 80-86.

Sforza, C., Michielon, G., Fragnito, N. & Ferrario, V. F. (1998). Foot asymmetry in healthy adults: elliptic fourier analysis of standardised footprints. *Journal of Orthopaedic Research, 16,* 758-765.

Shapiro, M. S., Kabo, J. M., Mitchell, P. W., Loren, G. & Tsenter, M. (1994). Ankle sprain prophylaxis: an analysis of the stabilising effects of braces and tape. *The American Journal of Sports Medicine, 22(*1), 78-82.

Shima, N., Maeda, A. & Hirohashi, K. (2005). Delayed latency of peroneal reflex to sudden inversion with ankle taping or bracing. *International Journal of Sports Medicine, 26*, 476-480.

Shultz, S. J. & Nguyen, A. D. (2007). Bilateral asymmetries in clinical measures of lower-extremity anatomic characteristics. *Clinical Journal of Sports Medicine, 17*(5), 357-361.

Simoneau, G. G. (1997). Changes in ankle joint proprioception resulting from strips of athletic tape applied over the skin. *Journal of Athletic Training, 32*(2), 141-147.

Sinnatamby, C. S. (1999), *Last's anatomy; regional and applied*. (10th ed.). London: Elsevier Churchill Livingstone.

Skelton, D. (2001). Effects of physical activity on postural stability. *Age and Ageing, 30*(S4), 33-39.

Smith, M., Brooker. S., Vicenzino, B & McPoil, T. (2004). Use of anti-pronation taping to assess suitability of orthotic prescription: case report. *Australian Journal of Physiotherapy, 50*(2), 111-113.

Soboroff, S., Pappius, E. & Komaroff, A. (1984). Benefits, risks and costs of alternative approaches to the evaluation and treatment of severe ankle sprain. *Clinical Orthopedics and Related Research, 183*, 160-168.

Soderberg, G. L. & Knutson, L. M. (2000). A guide for use and interpretation of kinesiologic electromyographical data. *Physical Therapy, 80*(5), 485-498.

Spanos, S. Brunswic, M. & Billis, E. (2008). The effect of taping on the proprioception of the ankle in a non-weight bearing position, amongst injured athletes. *The Foot, 18*, 25-33.

Staheli, L. T., Chew, D. E. & Corbett, M. (1987). The longitudinal arch. A survey of eight hundred and eighty-two feet in normal children and adults. *The Journal of Bone and Joint Surgery, 69*, 426-428.

Staudenmann, D., Roeleveld, K., Stegeman, D. F. & van Dieën, J. H. (2009). Methodological aspects of SEMG recordings for force estimation – A tutorial and review. *Journal of Electromyography and Kinesiology*, *20*(3), 375-387.

Stergiou, N. & Bates, B. T. (1997). The relationship between subtalar and knee joint function as a possible mechanism for running injuries. *Gait and Posture, 6*(3), 177-185.

Stevens, D. L. & Tomlinson, G. E. (1971). Measurement of human postural sway. *Proceedings of the Royal Society of Medicine, 64*(6), 653-655.

Suda, E. Y., Amorium, C. F. & Sacco, I, C. N (2009). Influence of ankle functional instability on the ankle electromyography during landing after volleyball blocking. *Journal of Electromyography and Kinesiology, 19*, e84-e93.

Tiberio, D. (1988). Pathomechanics of structural foot deformities, *Physical Therapy, 68*(12), 1840-1849.

Tobin, S. & Robinson, G. (2000). The effect of McConnell's vastus lateralis inhibition taping technique on vastus lateralis and vastus medialis obliquus activity. *Physiotherapy, 86*(4), 173-183.

Tropp, H. (2002). Commentary: functional ankle instability revisited. *Journal of Athletic Training, 37*(4), 512-515.

Tsai, L. C., Yu, B., Mercer, V. S. & Gross, M. T. (2006). Comparison of different structural foot types for measures of standing postural control. *Journal of Orthopaedic and Sports Physical Therapy, 36*(12), 942-953.

Vaes, P., Duquet, W. & Van Gheluwe, B. (2002). Peroneal reaction times and eversion motor response in healthy and unstable ankles. *Journal of Athletic Training, 37*(4), 475-480.

Venturni, C., Andre, A., Aguilar, B. P. & Giacomelli, B. (2006). Reliability of two evaluation methods of active range of motion in the ankle of healthy individuals. *Acta Fisiatrica, 13*(1), 39-43.

Verhagen, E. A. L. M., van der Beek, A. J. & van Mechelen, W. (2001). The effect of tape, braces and shoes on ankle range of motion. *Sports Medicine, 31*(9), 667-677.

Vicenzino, B. (2004). Foot orthotics in the treatment of lower limb conditions: a musculoskeletal physiotherapy perspective. *Manual Therapy, 9*(4), 185-196.

Vicenzino, B., Feilding, J., Howard, R., Moore, R. & Smith, S. (1997). An investigation of the anti-pronation effect of two taping methods after application and exercise. *Gait and Posture, 5*(1), 1-5.

Vicenzino, B., Franettovich, M., McPoil, T., Russell, T. & Skardoon, G. (2005). Initial effects of anti-pronation tape on the medial longitudinal arch during walking and running. *British Journal of Sports Medicine, 39(*12), 939-943.

Vicenzino, B., Griffiths, S. R., Griffiths, L. A. & Hadley, A. (2000). Effect of antipronation tape and temporary orthotic on vertical navicular height before and after exercise. *Journal of Orthopedic Sports Physical Therapy*, 30(6), 333-339.

Vicenzino, B., McPoil, T. & Buckland, S. (2007). Plantar foot pressures after the augmented lowDye taping technique. *Journal of Athletic Training, 42*(3), 374- 380.

Vicenzino, B., McPoil, T. G., Russell, T. & Peisker, S. (2006). Anti-pronation tape changes foot posture but not plantar ground contact during gait. *The Foot, 16*, 91-97.

Vincent, W. J. & Weir, J. P. (2012). *Statistics in Kinesiology*. (4th ed.). United States of America: Human Kinetics.

Vinicombe, A., Raspovic, A. & Menz, H. (2001). Reliability of navicular displacement measurement as a clinical indicator of foot posture. *Journal of American Podiatric Medical Association, 91*(5), 262-268.

Wall, K. B., Swankik, K. A. & Swankik, C. B. (2005). Augmented low-dye arch taping effects on muscle activity and ground reaction forces in people with pes planus. *Journal of Athletic Training, 40*(2), S98

Waterman, B. R., Owens, B. D., Davey. S., Zacchilli, M. A. & Belmont, P. J. (2010). The epidemiology of ankle sprains in the United States, *The Journal of Bone and Joint Surgery, 92*(13), 2279-2284.

Weir, J. P. (2005). Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. *Journal of Strength and Conditioning Research, 19*(1), 231-240.

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. *Sports Medicine, 36*(5), 392-410.

Wikstrom, E. A., Tillman, M. D., Kline, K. J. & Borsa, P. A. (2006). Gender and limb differences in dynamic postural stability during landing. *Clinical Journal of Sports Medicine, 16*(4), 311-315.

Wikstrom, E. A., Tillman, M. D., Schenker, S. M. & Borsa, P. A. (2008). Jump landing direction influences dynamic postural stability scores. *Journal of Science and Medicine in Sport, 11*(1), 106-111.

Wikstrom, E. A., Tillman, M. D., Smith, A. N. & Borsa, P. A. (2005). A new force-plate technology measure of dynamic postural stability: the dynamic postural stability index. *Journal of Athletic Training, 40*(4), 305-309.

Wilkerson, G. B. (2002). Biomechanical and neuromuscular effects of ankle taping and bracing. *Journal of Athletic Training, 37*(4), 436-445.

Williams, C. & Wragg, C. (2003). *Data Analysis and Research for Sport and Exercise Sciences: A Student Guide*. (1st ed.). London: Routledge.

Williams, D. S., McClay, I. S. & Hamill, J. (2001). Arch structure and injury patterns in runners. *Clinical Biomechanics, 16(1)*, 341-347.

Wilson, F., Byrne, A. & Gissane, C. (2011). A prospective study of injury and activity in elite soccer referees and assistant referees. *Irish Medical Journal, 104*(10), 295-297.

Winter, D. A. & Scott, S. H. (1991). Technique for interpretation of electromyography for concentric and eccentric contractions in gait. *Journal of Electromyography and Kinesiology, 1*(4), 263-269.

Winter, D. A. (1995). Human balance and posture control during standing and walking. *Gait and Posture, 3*(4), 193-214.

Wright, I. C., Neptune, R. R., van den Bogert, A. J. & Nigg, B. M. (2000). The influence of foot positioning on ankle sprains. *Journal of Biomechanics, 33*, 513-519.,

Yang, J. F. & Winter, D. A. (1984). Electromyographic amplitude normalisation methods: improving their sensitivity as diagnostic tools in gait analysis. *Archives of Physical Medicine and Rehabilitation, 65*(9), 517-521.

Yi, C. H., Brunt, D., Kim, H. D. & Fiolkowski, P. (2003). Effect of ankle taping and exercise on EMG and kinetics during landing. *Journal of Physical Therapy Science, 15*, 81-85.

Zazulak, B. T., Ponce, P. L., Straub, S. J., Medvecky, M. J., Avedisian, L. & Hewett, T. E. (2005). Gender comparison of hip muscle activity during single leg landing. *Journal of Orthopaedic and Sports Physical Therapy, 35*, 292-299.

Zhu, H., Wertsch, J. J., Harris, G. F. & Alba, H. M. (1995). Walking cadence effect on plantar pressures. *Archives of Physical Medicine and Rehabilitation, 76*(11), 1000-1005.

Zifchock, R.A., Davis, I., Hillstrom, H. & Song, J. (2006). The effect of gender, age and lateral dominance on arch height and stiffness. *Foot and Ankle International, 27*(5), 367-372.

Appendices

Appendix One: Subject Briefing Document

UNIVERSITY OF HERTFORDSHIRE
SCHOOL OF LIFE SCIENCES

Subject Briefing Document

Researcher: Joanna Denyer

Title of study: The Effects of Foot Structure and Athletic Taping on Lower Limb Biomechanics

Introduction

Sports injuries such as ankle sprains and stress fractures can be related to slight abnormalities in the structures of the foot and ankle, in particular abnormalities of the arch of the foot. Arch taping is widely used by athletes with abnormal foot structures to help support the arch, reducing stress on the surrounding structures and therefore reducing risk of injury. Ankle taping is also commonly used by athletes as extra support, for those who have existing ankle injuries, and for those with healthy ankles who use taping as a preventative measure.

Currently, there is inconclusive research in to the effects of taping and foot structure on lower limb biomechanics. This project aims to examine the effects of foot and ankle taping on postural stability, and muscular reaction time. It is important that these effects are established to determine whether foot and ankle taping actually has an effect on athletic performance and on reducing risk of injury.

Am I eligible to take part in this study?

You are eligible to participate in this study if you are a student at UH aged between 18-30 years. You must participate in a team sport for at least 2 hours a week. All participants will be personal contacts, there will be 60 volunteers used in this study.

What is involved?

The main testing protocol is identified below. Alternatively you may be required to participate in pilot testing of components of the procedure as directed by the researcher. If used for pilot testing, you will not be required to complete the whole of the main testing protocol.

For the main testing protocol, you will be required on three days for approximately 1 hour each day. You will be required to meet in G105 wearing sports shorts and t-shirt.

- Your height, mass and navicular drop height will be measured
- You will randomly select order of conditions (No-tape, Arch tape, Ankle tape, Proprioceptive tape)
- You will practice movements until you feel comfortable and perform them with correct technique (forward hop, diagonal hop, lateral hop, 30 cm drop jump, 45° cut, forward run and back pedal)
- The taping condition will be applied
- Surface electrodes will be placed on the your skin (3 muscles on each leg)
- Postural stability: you will perform the seven movements across a force platform, each movement will be performed three times on the dominant leg
- Muscle reaction time: you will stand on the tilt platform, and the tilt mechanism will be randomly initiated six times

- Subjects will perform exercise protocol – 15 minutes of moderate exercise on treadmill followed by mat based exercises.
- Postural stability and Muscle reaction time measures will be repeated.
- The taping technique will be removed
- This procedure will be repeated with each condition during the four testing sessions

<u>When should I refuse to take part?</u>

You should refuse to take part on this study if any of the following apply:
Recent lower limb injury
Myositis ossificans
Poor circulation
General illness
Acute trauma to lower limb
Soft tissue inflammation
Skin infection / condition
Allergy to tape
Allergy to tape adhesive
Allergy to tape adhesive removal fluid / alcohol wipes
Are under the influence of alcohol or any other psycho-active substance
Lower limb surgery
Regular use of orthotics, taping or bracing
Participation in gymnastics, dance, martial arts
Do not wish to take part

<u>What are the adverse effects?</u>

There are no known adverse effects to this study. All risks have been assessed and protocols of safe working will be followed to ensure that no participants will be harmed in this study. A health screen will be used to ensure that all volunteers are able to be used as subjects in this study. The lead tester is first aid trained; any questions regarding the effects of this study should be directed to the researcher.

<u>Consent</u>

Before the testing will commence, a consent form (LEC2) must be read and signed by each participant. This indicates you have read and understood the subject briefing. There will be an opportunity to ask any further questions before the consent form is signed. Participation in this study is voluntary, and you have the right to withdraw at any time, without prejudice or having to give reason. If after consenting to participate, you withdraw your consent, any information already obtained will be removed from the study and destroyed.

<u>Personal Data</u>

Some personal data (height, mass, age) is collected in this study so that subject characteristics can be described. All data collected in this study will be treated with confidentiality, and will be kept on a password protected laptop. Any personal data will be kept in a locked cabinet at the residence of the researcher. Once the examination process is complete, personal data will be destroyed. Participant's names will not be used, and there will be no way of identifying individual subjects in the write up of this study.

Appendix Two: Consent Form

UNIVERSITY OF HERTFORDSHIRE
SCHOOL OF LIFE SCIENCES

Consent Form

I, the undersigned, agree to take part in:

Approved Protocol Number:.....................LSGP1/09/065...

Title of Study: The Effects of Foot Structure and Athletic Taping on Lower Limb Biomechanics

to be carried out by

Name of Investigator(s)...........Joanna Denyer..

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(s) of investigator(s) ....Joanna Denyer..

Signature(s) of investigator(s).................................... **Date** ....

UNIVERSITY OF HERTFORDSHIRE
SCHOOL OF LIFE SCIENCES

<u>Health Screen Document</u>

Researcher: Joanna Denyer

Title of study: The Effects of Foot Structure and Athletic Taping on Lower Limb Biomechanics

Name of participant:

It is important when having volunteered as subject for this study, and having read the briefing sheet for subjects that you answer the following questions. Please do not answer any questions if you consider them intrusive.

1) Do you suffer from high blood pressure, or any heart problems?

Yes	No

2) Do you often get dizzy, or do you know that you have low blood pressure?

Yes	No

3) When and what did you last eat?

4) Are you under the influence of alcohol or any other psycho-active substance?

Yes	No

5) Have you had a cold or flu in the last two weeks?

Yes	No

6) Are you suffering from any musculo-skeletal injury?

Yes	No

7) Are you currently taking any medication (over the counter, or prescription)?

Yes	No

8) Have you ever been told that you should not exercise?

Yes	No

9) Do you feel fully fit, and eager to act as subject?

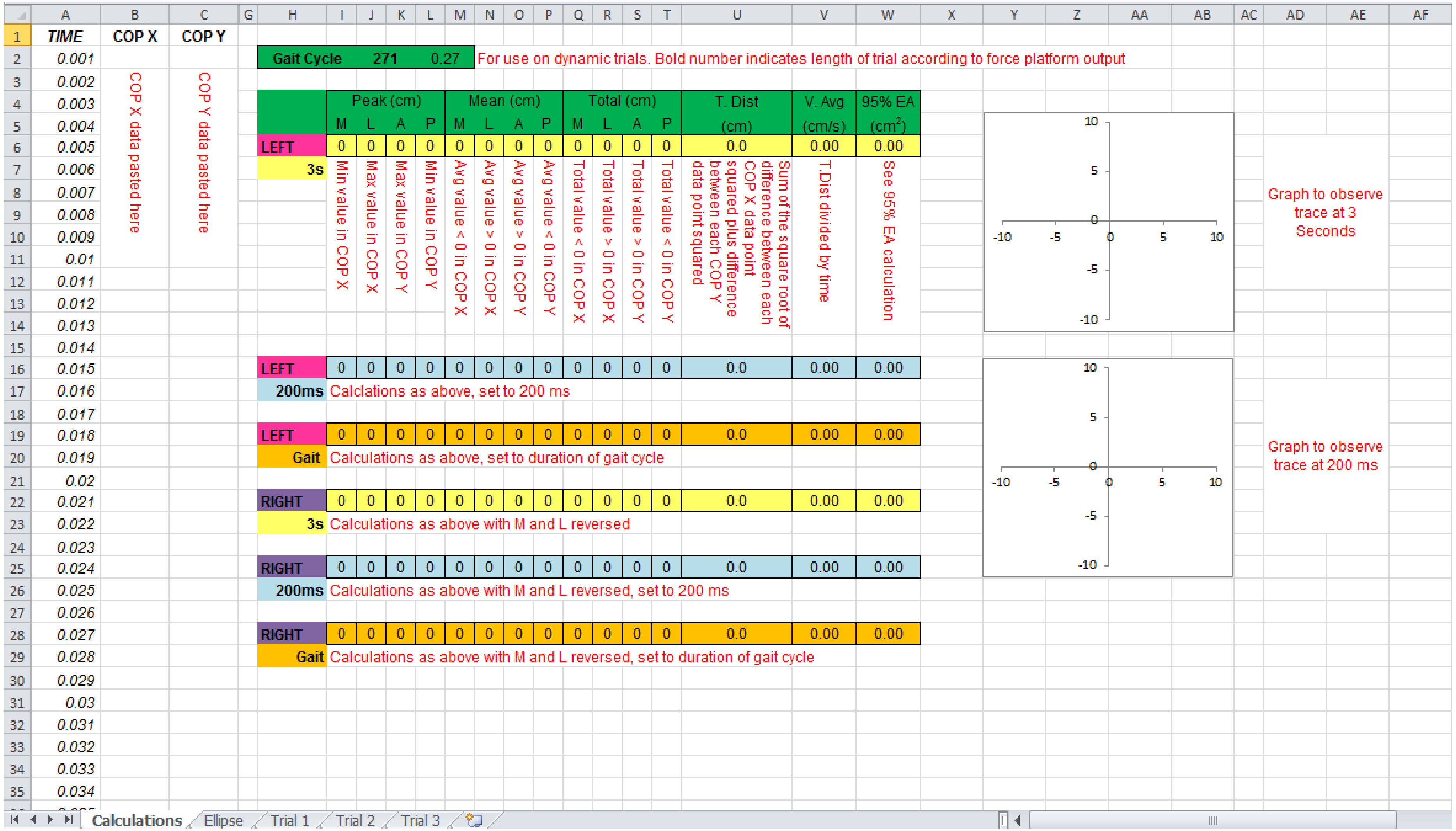
Yes	No

10) Is there any reason, not stated above, why you cannot take part as a subject in this practical?

Yes	No

Signature of subject: .... **Date:** ....

Checked by (Name): .... **Date:** ....



	TIME	COP X	COP Y
1	TIME	COP X	COP Y
2	0.001		
3	0.002	COP X data pasted here	COP Y data pasted here
4	0.003		
5	0.004		
6	0.005		
7	0.006		
8	0.007		
9	0.008		
10	0.009		
11	0.01		
12	0.011		
13	0.012		
14	0.013		
15	0.014		
16	0.015		
17	0.016		
18	0.017		
19	0.018		
20	0.019		
21	0.02		
22	0.021		
23	0.022		
24	0.023		
25	0.024		
26	0.025		
27	0.026		
28	0.027		
29	0.028		
30	0.029		
31	0.03		
32	0.031		
33	0.032		
34	0.033		
35	0.034		

Gait Cycle 271 0.27 For use on dynamic trials. Bold number indicates length of trial according to force platform output

	Peak (cm)				Mean (cm)				Total (cm)				T. Dist (cm)	V. Avg (cm/s)	95% EA (cm^2)
	M	L	A	P	M	L	A	P	M	L	A	P			
LEFT	0	0	0	0	0	0	0	0	0	0	0	0	0.0	0.00	0.00
3s	Min value in COP X	Max value in COP X	Max value in COP Y	Min value in COP Y	Avg value < 0 in COP X	Avg value > 0 in COP X	Avg value > 0 in COP Y	Avg value < 0 in COP Y	Total value < 0 in COP X	Total value > 0 in COP X	Total value > 0 in COP Y	Total value < 0 in COP Y	Sum of the square root of difference between each COP X data point squared plus difference between each COP Y data point squared	T.Dist divided by time	See 95% EA calculation
LEFT	0	0	0	0	0	0	0	0	0	0	0	0	0.0	0.00	0.00
200ms	Calclations as above, set to 200 ms														
LEFT	0	0	0	0	0	0	0	0	0	0	0	0	0.0	0.00	0.00
Gait	Calculations as above, set to duration of gait cycle														
RIGHT	0	0	0	0	0	0	0	0	0	0	0	0	0.0	0.00	0.00
3s	Calculations as above with M and L reversed														
RIGHT	0	0	0	0	0	0	0	0	0	0	0	0	0.0	0.00	0.00
200ms	Calculations as above with M and L reversed, set to 200 ms														
RIGHT	0	0	0	0	0	0	0	0	0	0	0	0	0.0	0.00	0.00
Gait	Calculations as above with M and L reversed, set to duration of gait cycle														

Graph to observe trace at 3 Seconds

Graph to observe trace at 200 ms

Calculations Ellipse Trial 1 Trial 2 Trial 3

Excel Template Used To Calculate Centre of Pressure Parameters

	A	B	C	D	E	F	G	H	I
1	Time	COP X Relative	COP Y Relative	X^2	y^2	X^2*Y^2			
2	0.001			=B2*B2	=C2*C2	=D2*E2			
3	0.002	COP X data pasted here	COP Y data pasted here	=B3*B3	=C3*C3	=D3*E3		**95% EA at Three Seconds**	
4	0.003			=B4*B4	=C4*C4	=D4*E4		Std Dev COP X	=STDEV(B2:B3001)
5	0.004			=B5*B5	=C5*C5	=D5*E5		StdDev^2 COP X	=I4^2
6	0.005			=B6*B6	=C6*C6	=D6*E6		Std Dev COP Y	=STDEV(C2:C3001)
7	0.006			=B7*B7	=C7*C7	=D7*E7		StdDev^2 COP Y	=I6^2
8	0.007			=B8*B8	=C8*C8	=D8*E8		X + Y	=I5+I7
9	0.008			=B9*B9	=C9*C9	=D9*E9		Square	=I8^2
10	0.009			=B10*B10	=C10*C10	=D10*E10		Multiply	=I5*I7
11	0.01			=B11*B11	=C11*C11	=D11*E11		Correl	=CORREL(B2:B3001,C2:C3001)
12	0.011			=B12*B12	=C12*C12	=D12*E12		Correl^2	=I11^2
13	0.012			=B13*B13	=C13*C13	=D13*E13		Sum	=SUM(F2:F3001)
14	0.013			=B14*B14	=C14*C14	=D14*E14		Square Root	=SQRT(I13)
15	0.014			=B15*B15	=C15*C15	=D15*E15		CoV XY	=(1/3000)*I14
16	0.015			=B16*B16	=C16*C16	=D16*E16		CoV XY^2	=I15^2
17	0.016			=B17*B17	=C17*C17	=D17*E17		minus	=I10-I12
18	0.017			=B18*B18	=C18*C18	=D18*E18		x4	=I17*4
19	0.018			=B19*B19	=C19*C19	=D19*E19		subtract	=I9-I18
20	0.019			=B20*B20	=C20*C20	=D20*E20		D=	=SQRT(I19)
21	0.02			=B21*B21	=C21*C21	=D21*E21		Sum	=I5+I7+I20
22	0.021			=B22*B22	=C22*C22	=D22*E22		F= 3	=3*I21
23	0.022			=B23*B23	=C23*C23	=D23*E23		S major	=SQRT(I22)
24	0.023			=B24*B24	=C24*C24	=D24*E24		S minor	=SQRT(3*(I8-I20))
25	0.024			=B25*B25	=C25*C25	=D25*E25		**Area**	=(2*3.1415963*3)*SQRT(I10-I12)
26	0.025			=B26*B26	=C26*C26	=D26*E26			
27	0.026			=B27*B27	=C27*C27	=D27*E27			
28	0.027			=B28*B28	=C28*C28	=D28*E28			
29	0.028			=B29*B29	=C29*C29	=D29*E29			
30	0.029			=B30*B30	=C30*C30	=D30*E30			
31	0.03			=B31*B31	=C31*C31	=D31*E31			
32	0.031			=B32*B32	=C32*C32	=D32*E32			
33	0.032			=B33*B33	=C33*C33	=D33*E33		Calculated with the assistance of Neil Willmore (Laboratory Technician) using guideance from AMTI Bioanalysis 2.2, Watertown MA.	
34	0.033			=B34*B34	=C34*C34	=D34*E34			
35	0.034			=B35*B35	=C35*C35	=D35*E35			
36	0.035			=B36*B36	=C36*C36	=D36*E36			
37	0.036			=B37*B37	=C37*C37	=D37*E37			
38	0.037			=B38*B38	=C38*C38	=D38*E38			
39	0.038			=B39*B39	=C39*C39	=D39*E39			
40	0.039			=B40*B40	=C40*C40	=D40*E40			
41	0.04			=B41*B41	=C41*C41	=D41*E41			

Calculations | Ellipse | Trial 1 | Trial 2 | Trial 3

Excel Template Used To Calculate 95% Ellipse Area, (Showing Formulas)

Appendix Five: Definition of COP Parameters

Definition of Centre of Pressure Parameters

Variable		Definition
Peak (cm)		Peak value in each direction
	Medial	Maximum value in X-axis (> 0)
	Lateral	Minimum value in X-axis (< 0)
	Anterior	Maximum value in Y-axis (> 0)
	Posterior	Minimum value in Y-axis (< 0)
Average (cm)		Average value in each direction
	Medial	Average value in X-axis (> 0)
	Lateral	Average value in X-axis (< 0)
	Anterior	Average value in Y-axis (> 0)
	Posterior	Average value in Y-axis (< 0)
Total (cm)		Total of values in each direction
	Medial	Average value in X-axis (> 0)
	Lateral	Average value in X-axis (< 0)
	Anterior	Average value in Y-axis (> 0)
	Posterior	Average value in Y-axis (< 0)
Total Distance (cm)		Total length (L) of COP path $$L = \sum_{i=2}^{n} \sqrt{(x_i - x_{i-1})^2 + (y_i - y_{i-1})^2}$$
Average Velocity (cm/s)		Total distance travelled by COP over time; the total length of the COP path (L) is calculated. This number is then divided by the total number of frames (n) and the total change in time (Δt). $$\text{Vavg} = \frac{L}{n * \Delta t}$$
95% Ellipse Area (cm^2)		Area that with 95% probability contains the centre of the points of postural sway. $$Area = (2 \times 3.1415963 \times F)\sqrt{\sigma_x^2 \sigma_y^2 - \sigma_{xy}^2}$$

Test Retest Reliability of Hopping Tasks at 200 ms – Raw Data

		Forward Hop					Diagonal Hop					Lateral Hop				
		Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM
Peak (cm)	M	2.1(1.0)	2.0(0.9)	.68	.77	0.41	1.7(1.8)	1.0(0.9)	.37	.22	1.27	1.6(0.7)	1.4(0.7)	.41	.33	0.60
	L	1.3(0.5)	1.4(0.4)	.94	.19	0.40	3.5(1.4)	3.5(2.7)	.85	.74	1.01	3.6(3.3)	3.6(1.9)	.93	.86	0.96
	A	3.9(2.1)	4.6(2.0)	.27	.75	1.01	6.4(2.9)	7.0(3.6)	.58	.82	1.32	8.4(1.3)	8.6(1.0)	.54	.67	0.62
	P	4.6(3.6)	3.6(3.5)	.37	.76	1.68	3.4(3.1)	3.1(5.1)	.05	.94	0.96	1.8(1.1)	1.1(1.5)	.13	.72	0.70
Mean (cm)	M	0.6(0.2)	0.9(0.5)	.08	.85	0.15	0.5(0.4)	0.6(0.4)	.89	.66	0.22	0.7(0.5)	0.8(0.6)	.36	.88	0.18
	L	0.5(0.2)	0.5(0.1)	.49	.35	0.12	1.2(0.4)	1.1(0.9)	.50	.85	0.25	1.0(0.4)	1.3(0.9)	.23	.78	0.34
	A	2.0(1.4)	2.5(1.0)	.22	.90	0.38	2.8(1.4)	3.0(0.5)	.96	.89	0.35	4.0(1.1)	4.2(0.9)	.51	.72	0.52
	P	2.3(2.3)	2.2(1.6)	.41	.84	0.77	1.7(1.4)	1.5(1.9)	.46	.81	0.69	0.7(0.6)	0.6(0.8)	.78	.93	0.17
Total (cm)	M	5.9(0.9)	5.2(1.7)	.25	.67	0.78	4.9(3.2)	5.0(2.0)	.89	.34	2.11	5.6(3.3)	3.5(1.7)	.08	.38	2.18
	L	4.2(1.4)	5.6(3.3)	.14	.83	1.04	6.6(2.1)	7.5(2.8)	.42	.67	1.38	7.2(5.6)	6.4(2.5)	.70	.33	3.45
	A	7.5(2.7)	10.3(4.4)	.17	.19	3.42	11.2(4.8)	13.3(3.7)	.29	.73	2.24	13.7(4.7)	12.7(3.6)	.26	.84	1.60
	P	10.8(7.3)	10.6(8.7)	.82	.96	1.56	8.9(6.2)	9.2(7.0)	.71	.92	1.85	6.4(3.6)	5.1(3.6)	.17	.78	1.67
T.Dist (cm)		22.9(6.8)	25.7(13.5)	.34	.97	1.73	25.4(9.1)	28.2(9.1)	.56	.79	4.01	26.4(12.8)	22.2(8.7)	.09	.90	3.32
V.Avg (cm/s)		114.5(34.0)	128.3(67.3)	.34	.97	8.76	126.8(45.2)	140.7(45.3)	.57	.79	20.09	131.8(63.9)	110.8(43.4)	.09	.91	16.57
95% EA (cm^2)		23.7(19.0)	26.0(14.9)	.89	.92	4.82	40.1(22.2)	50.2(28.9)	.50	.63	15.26	41.3(27.7)	47.2(17.6)	.49	.58	14.69

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement; M: medial; L: lateral; A: anterior; P: posterior; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). ICC > .70 indicating high correlation highlighted.

Test Retest Reliability of Forward Run, 45° Cut and Backpedal

		Forward Run					45° Cut					Backpedal				
		Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM
Peak (cm)	M	3.6(1.7)	3.4(2.2)	.05	.53	0.59	7.5(4.2)	13.3(6.2)	.05	.66	3.43	11.0(7.3)	9.8(4.6)	.75	.02	5.79
	L	2.5(1.2)	7.3(4.3)	.05	.54	0.79	17.8(13.8)	18.3(11.5)	.48	.89	4.08	9.0(4.8)	6.8(5.7)	.47	.16	4.71
	A	10.4(4.6)	8.8(4.7)	.46	.21	4.09	11.3(6.4)	11.3(10.8)	.99	.20	7.54	16.0(10.3)	9.7(6.6)	.20	.28	7.55
	P	8.4(5.9)	8.0(4.7)	.73	.21	4.61	11.6(15.2)	6.2(2.2)	.42	.84	4.19	16.5(16.6)	17.5(8.5)	.89	.21	11.19
Mean (cm)	M	2.3(1.8)	3.0(2.2)	.08	.55	0.50	7.7(5.8)	9.3(8.1)	.34	.52	0.00	10.8(5.9)	6.5(4.9)	.05	.53	0.57
	L	2.4(2.3)	4.7(2.2)	.05	.74	1.25	13.0(11.2)	12.8(6.8)	.99	.30	7.53	7.5(0.7)	6.3(5.5)	.40	.54	0.00
	A	5.1(2.8)	4.6(3.8)	.05	.56	0.88	4.0(2.0)	3.4(2.4)	.38	.15	1.94	5.3(7.5)	4.0(4.2)	.50	.01	3.20
	P	3.4(1.9)	3.5(1.5)	.36	.70	0.91	6.3(10.2)	2.4(0.5)	.62	.53	2.25	14.6(14.2)	14.0(7.7)	.52	.16	9.87
Total (cm)	M	12.1(4.4)	14.6(6.1)	.10	.80	2.44	35.3(25.3)	41.0(33.5)	.19	.99	2.71	8.8(4.1)	9.7(5.9)	.34	.55	0.41
	L	11.3(4.8)	14.7(8.0)	.05	.82	2.81	33.7(25.6)	35.3(29.6)	.58	.98	3.86	11.2(3.3)	7.3(3.5)	.09	.13	3.53
	A	17.2(4.6)	13.4(4.8)	.05	.59	3.17	17.2(5.9)	15.8(4.4)	.17	.96	0.97	18.2(9.0)	18.0(8.1)	.95	.78	3.86
	P	26.5(9.5)	22.3(4.1)	.12	.47	5.38	26.5(6.3)	20.8(5.8)	.05	.75	3.29	8.5(4.9)	10.0(4.1)	.58	.08	4.22
T.Dist (cm)		54.3(17.0)	51.0(16.5)	.36	.67	9.46	93.3(52.8)	95.7(63.2)	.70	.99	6.44	37.0(14.7)	36.2(14.8)	.74	.92	3.99
V.Avg(cm/s)		288.3(143.5)	289.8(124.2)	.88	.85	50.59	488.4(248.1)	311.0(255)	.55	.99	18.21	131.8(19.2)	174.0(40.4)	.05	.62	22.87
95% EA (cm^2)		49.4(25.3)	61.6(37.1)	.12	.87	11.59	137.0(108.9)	151.8(150.4)	.16	1.0	5.03	31.2(13.2)	35.2(23.4)	.30	.58	4.71

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement; M: medial; L: lateral; A: anterior; P: posterior; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). ICC > .70 indicating high correlation highlighted.

Test Retest Reliability of Drop Jump; at 200 ms and Three Seconds

		Drop Jump (200 ms)					Drop Jump (Three Seconds)				
		Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM
Peak (cm)	M	1.5(0.4)	1.6(0.6)	.69	.58	0.32	1.9(0.8)	2.0(0.4)	.67	.24	0.52
	L	1.3(0.9)	1.3(0.8)	.50	.43	0.63	2.0(0.7)	1.5(0.4)	.05	.34	0.49
	A	9.1(2.9)	9.1(2.1)	.59	.84	0.98	9.2(2.6)	9.3(2.2)	.84	.81	1.04
	P	2.7(1.3)	2.3(1.7)	.22	.46	1.10	3.6(1.3)	2.3(0.5)	.05	.10	1.10
Mean (cm)	M	0.6(0.2)	0.5(0.2)	.43	.18	0.19	0.6(0.3)	0.7(0.2)	.42	.04	0.22
	L	0.4(0.2)	0.5(0.3)	.55	.25	0.23	0.7(0.2)	0.7(0.2)	.25	.65	0.11
	A	3.8(1.3)	4.4(0.8)	.05	.82	0.47	1.9(1.0)	1.5(0.5)	.22	.37	0.63
	P	1.5(0.7)	1.4(0.9)	.27	.42	0.57	1.0(0.3)	0.8(0.2)	.05	.74	0.13
Total (cm)	M	7.5(2.0)	7.9(4.1)	.71	.55	2.14	25.5(4.8)	23.2(2.9)	.05	.76	1.98
	L	6.8(2.8)	6.4(3.4)	.40	.35	2.49	24.8(4.1)	21.9(2.8)	.05	.81	1.63
	A	19.4(4.2)	16.9(1.8)	.06	.55	2.28	37.7(2.7)	39.2(4)	.11	.74	1.73
	P	14.1(3.0)	12.3(2.8)	.05	.44	2.23	29.0(2.3)	29.6(2.5)	.39	.60	1.49
T. Dist (cm)		38.9(7.8)	35.3(6.9)	.06	.47	5.40	92.6(8.4)	91.4(7.3)	.65	.50	5.45
V.Avg (cm/s)		194.5(38.9)	176.3(34.4)	.06	.47	26.95	30.9(2.8)	30.5(2.4)	.64	.50	1.82
95% EA (cm^2)		20.2(7.9)	20.7(10.3)	.46	.39	6.97	23.1(6.7)	24.2(9.7)	.58	.74	4.18

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement; M: medial; L: lateral; A: anterior; P: posterior; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). ICC > .70 indicating high correlation highlighted.

Test Retest Reliability of Hopping Tasks at Three Seconds

		Forward Hop					Diagonal Hop					Lateral Hop				
		Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM	Test	Retest	*P*	ICC	SEM
Peak (cm)	M	2.2(0.7)	2.4(1.0)	.40	.51	0.57	1.8(0.6)	1.9(0.2)	.40	.29	0.37	2.2(0.7)	1.9(0.5)	.34	.36	0.47
	L	1.7(1.5)	2.2(0.8)	.05	.51	0.19	3.6(1.3)	3.6(2.1)	.90	.77	0.81	2.8(1.5)	3.7(1.9)	.62	.16	1.56
	A	4.8(1.3)	5.0(1.8)	.79	.67	0.92	6.7(2.5)	7.3(2.5)	.40	.74	1.23	8.5(1.3)	8.6(0.9)	.68	.65	0.65
	P	6.8(2.0)	6.3(4.2)	.47	.97	0.79	5.5(4.0)	4.2(4.3)	.15	.85	1.56	3.8(2.0)	3.3(2.0)	.35	.75	0.96
Mean (cm)	M	0.6(0.3)	0.8(0.3)	.05	.63	0.15	0.8(0.2)	0.8(0.1)	.44	.03	0.18	0.8(0.3)	0.7(0.2)	.05	.52	0.07
	L	0.6(0.5)	0.6(0.4)	.89	.31	0.27	0.9(0.2)	0.8(0.2)	.31	.46	0.15	0.8(0.6)	0.8(0.3)	.96	.38	0.34
	A	1.5(0.4)	1.3(0.6)	.33	.72	0.35	1.3(0.4)	1.5(0.5)	.46	.50	0.30	1.5(0.4)	1.3(0.2)	.15	.57	0.22
	P	1.0(0.9)	1.1(0.3)	.90	.41	0.24	1.2(0.5)	1.3(0.5)	.81	.54	0.32	1.5(0.9)	1.2(0.4)	.27	.68	0.40
Total (cm)	M	18.7(3.3)	19.9(3.6)	.28	.62	1.74	17.5(5.0)	17.9(2.3)	.78	.66	2.18	17.8(3.3)	16.4(2.0)	.22	.42	2.07
	L	17.4(5.8)	18.9(3.7)	.12	.84	1.25	19.6(2.9)	20.2(3.8)	.42	.89	1.07	20.5(5.8)	19.8(3.7)	.48	.96	0.97
	A	23.1(5.1)	26.4(5.6)	.07	.70	2.58	26.5(5.4)	28.1(4.7)	.31	.66	2.91	29.5(5.1)	28.4(5.3)	.38	.82	2.15
	P	24.1(5.5)	23.8(6.7)	.76	.92	1.97	23.3(7.1)	22.8(7.7)	.54	.96	1.39	21.8(5.5)	20.2(6.3)	.21	.85	2.25
T.Dist (cm)		66.2(11.7)	70.4(14.3)	.11	.96	2.34	66.1(13.7)	70.7(12.6)	.33	.57	8.51	68.0(11.7)	67.3(13.9)	.90	.38	9.83
V.Avg (cm/s)		22.1(4.9)	23.5(4.8)	.11	.96	0.78	23.1(4.2)	23.6(4.2)	.51	.89	1.34	23.9(4.9)	22.4(4.6)	.07	.92	1.34
95% EA (cm^2)		22.5(27.1)	30.7(18.1)	.16	.67	9.19	32.2(15.8)	31.8(13.9)	.94	.67	8.27	42.1(27.1)	33.3(22.8)	.34	.53	16.9

Values are mean (SD). *P:* p-value of the paired t-test on test retest differences; ICC: intraclass correlation coefficient; SEM: standard error of measurement; M: medial; L: lateral; A: anterior; P: posterior; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). ICC > .70 indicating high correlation highlighted.

Results of Hopping Tasks; 200 ms

		Forward Hop			Diagonal Hop			Lateral Hop		
		Neutral	Pronated	Supinated	Neutral	Pronated	Supinated	Neutral	Pronated	Supinated
Peak (cm)	A	4.2(1.5)	5.8(1.9)	5.2(1.7)	6.8(1.6)	7.6(2.3)	6.0(2.2)	8.8(2.3)	9.6(1.1)	9.0(2.3)
	L	0.8(0.3)	1.4(0.9)	1.6(1.6)	2.3(1.5)	2.5(1.4)	1.4(0.9)	2.6(1.7)	2.4(1.0)	1.6(0.8)
	P	2.0(1.2)	4.9(5.6)	1.8(0.6)	1.8(0.9)	1.7(0.8)	1.6(0.6)	1.4(0.7)	2.1(1.2)	0.9(0.6)
	M	2.5(1.2)	2.0(1.0)	2.9(1.8)	1.1(0.8)	1.1(0.7)	1.8(1.1)	1.1(0.6)	1.2(0.9)	2.0(0.9)
Mean (cm)	A	2.4(0.8)	3.0(0.6)	2.9(0.7)	3.7(1.0)	3.7(1.0)	3.4(1.5)	4.6(0.9)	5.1(1.1)	4.0(0.5)
	L	0.6(0.4)	0.6(0.3)	0.7(0.4)	0.7(0.4)	1.0(0.4)	0.7(0.4)	0.9(0.5)	0.9(0.4)	0.6(0.3)
	P	1.7(1.4)	2.3(2.3)	1.3(1.3)	1.0(0.7)	1.3(0.8)	1.8(1.4)	0.9(0.9)	0.9(0.4)	0.6(0.4)
	M	1.0(0.4)	0.7(0.3)	1.1(0.6)	0.7(0.5)	0.7(0.5)	0.9(0.5)	0.8(0.4)	0.7(0.5)	1.1(0.5)
Total (cm)	A	8.4(2.3)	8.4(4.0)	9.3(3.4)	9.8(2.9)	10.3(2.1)	11.2(3.6)	11.7(3.7)	11.4(1.8)	12.2(3.1)
	L	3.3(1.3)	4.4(2.9)	4.4(2.7)	4.5(1.9)	4.9(1.2)	4.7(1.5)	5.6(2.4)	5.6(1.8)	5.6(1.6)
	P	7.8(4.2)	10.7(9.0)	6.4(2.2)	4.1(2.5)	7.1(5.4)	5.4(2.0)	3.6(0.7)	4.9(3.3)	5.8(2.4)
	M	4.7(1.5)	5.2(2.0)	5.4(1.5)	3.4(1.7)	4.7(2.3)	5.2(1.5)	3.4(1.5)	3.6(1.6)	4.5(1.6)
T. Dist (cm)		20.8(6.3)	23.3(11.8)	19.1(4.9)	16.9(4.8)	20.8(6.1)	21.2(6.1)	20.4(6.9)	19.9(5.5)	22.9(6.3)
V.Avg (cm/s)		98.8(25.7)	84.6(16.6)	95.6(24.4)	80.0(18.2)	96.0(20.6)	105.9(30.5)	89.7(16.4)	99.4(27.3)	114.5(31.3)
95% EA (cm^2)		20.8(7.2)	21.3(7.3)	25.2(9.3)	28.2(13.4)	33.3(7.6)	29.4(11.1)	42.1(15.3)	41.7(14.7)	33.2(7.8)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). Significant increase in comparison to neutral group ($P < .05$) indicated.

Results of Forward Run, 45° Cut and Backpedal Tasks

		Forward Run			45° Cut			Backpedal		
		Neutral	Pronated	Supinated	Neutral	Pronated	Supinated	Neutral	Pronated	Supinated
Peak (cm)	A	11.2(5.0)	11.9(4.8)	13.3(4.5)	11.0(4.8)	12.1(5.2)	9.0(4.6)	10.0(3.8)	7.1(3.2)	5.6(3.3)
	L	5.1(4.1)	4.8(2.8)	3.5(3.4)	14.3(8.5)	11.9(4.7)	11.9(8.4)	5.9(2.9)	4.9(2.9)	7.1(2.8)
	P	11.0(5.7)	8.9(5.5)	6.2(4.6)	22.9(10.7)	15.2(7.7)	14.0(7.5)	5.9(1.7)	5.4(1.9)	7.4(3.4)
	M	5.7(2.9)	6.2(4.8)	4.2(1.9)	14.2(5.0)	10.9(6.5)	10.4(6.9)	8.5(5.3)	5.6(2.1)	9.1(5.7)
Mean (cm)	A	7.1(4.2)	6.7(2.7)	7.0(2.4)	3.1(1.7)	6.9(4.7)	3.8(3.2)	4.0(2.3)	2.9(1.7)	2.2(2.1)
	L	2.0(2.0)	2.5(2.2)	2.1(1.5)	12.4(7.3)	8.0(6.2)	8.3(6.0)	2.8(1.8)	1.8(1.2)	2.7(1.5)
	P	5.7(4.3)	3.7(2.4)	2.5(1.4)	16.5(7.5)	9.7(7.9)	10.8(10.0)	3.7(2.9)	2.5(1.6)	5.3(3.1)
	M	3.1(1.9)	4.0(4.4)	1.5(0.8)	6.6(4.1)	8.8(5.2)	1.7(1.2)	6.2(4.4)	5.2(3.8)	7.7(3.9)
Total (cm)	A	8.6(3.4)	8.9(4.0)	11.9(4.6)	24.3(21.7)	16.3(11.7)	15.9(8.5)	17.4(4.6)	15.8(5.8)	15.2(3.5)
	L	8.4(2.9)	6.9(2.4)	7.8(5.2)	18.8(11.6)	11.6(4.9)	11.6(7.3)	9.0(3.4)	9.5(3.5)	8.9(2.5)
	P	25.2(7.2)	22.9(8.3)	22.8(12.5)	42.1(25.2)	30.6(8.3)	30.1(17.7)	11.9(3.5)	11.7(5.5)	8.7(3.0)
	M	9.4(4.1)	9.2(3.0)	9.9(4.8)	18.0(8.0)	16.5(7.6)	14.1(7.3)	7.4(3.1)	7.5(4.4)	7.0(2.4)
T. Dist (cm)		42.4(7.0)	36.4(9.9)	43.3(19.3)	83.8(55.3)	59.8(21.4)	57.0(29.2)	36.7(8.7)	34.4(14.3)	31.4(7.1)
V.Avg (cm/s)		211.5(58.0)	190.9(75.5)	180.7(56.6)	214.5(50.9)	213.9(56.2)	229.8(112)	136.6(35.5)	114.4(26.7)	120.2(32.3)
95% EA (cm^2)		71.3(27.9)	56.3(12.8)	49.8(23.8)	104.1(48.7)	103.7(27.6)	68.8(39.3)	35.1(15.5)	30.3(20.1)	35.5(17.5)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). Significant increase in comparison to neutral group (P < .05) indicated.

Results of Drop Jump; 200 ms and Three Seconds

		Drop Jump 3 Seconds			Drop Jump 200 ms		
		Neutral	Pronated	Supinated	Neutral	Pronated	Supinated
Peak (cm)	A	8.5(2.1)	7.8(1.0)	9.0(2.6)	8.3(2.3)	7.7(0.9)	8.7(2.7)
	L	2.3(0.5)	2.6(1.6)	2.2(0.9)	2.1(0.6)	1.7(0.8)	1.5(1.1)
	P	4.2(1.3)	5.5(2.3)	4.2(0.9)	3.1(0.7)	3.8(2.0)	2.7(1.2)
	M	2.2(0.8)	2.9(1.4)	3.0(1.2)	1.6(1.1)	2.7(1.7)	2.3(1.0)
Mean (cm)	A	1.5(0.4)	2.0(0.5)	2.0(0.6)	3.2(0.8)	3.4(0.7)	3.9(0.9)
	L	0.8(0.2)	0.9(0.6)	1.0(0.3)	0.8(0.3)	0.7(0.3)	0.5(0.2)
	P	1.3(0.4)	1.5(0.6)	1.4(0.2)	1.6(0.4)	1.9(0.9)	1.5(0.6)
	M	0.8(0.2)	0.9(0.3)	1.1(0.4)	0.6(0.2)	0.7(0.3)	0.9(0.6)
Total (cm)	A	36.1(5.5)	36.6(3.9)	37.5(5.7)	18.3(5.6)	17.0(4.0)	18.0(5.0)
	L	21.3(2.7)	20.4(2.7)	21.0(4.2)	6.9(2.1)	6.2(1.6)	7.4(3.6)
	P	29.4(5.2)	29.2(5.1)	29.1(4.9)	12.6(4.8)	13.7(5.3)	13.1(4.4)
	M	20.8(2.5)	21.1(3.4)	22.1(2.7)	6.6(1.9)	6.8(1.7)	7.8(1.7)
T. Dist (cm)		88.4(9.5)	87.5(14.3)	87.5(12.8)	36.3(11)	36.7(9.7)	37.5(10.6)
V.Avg (cm/s)		28.9(3.8)	29.2(4.8)	29.2(4.3)	173.6(47.5)	183.5(48.6)	187.3(52.8)
95% EA (cm^2)		29.0(6.2)	29.4(3.5)	44.6(11.7)	22.6(7.7)	30.4(11.4)	20.9(7.9)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). Significant increase in comparison to neutral group (*P* < .05) indicated.

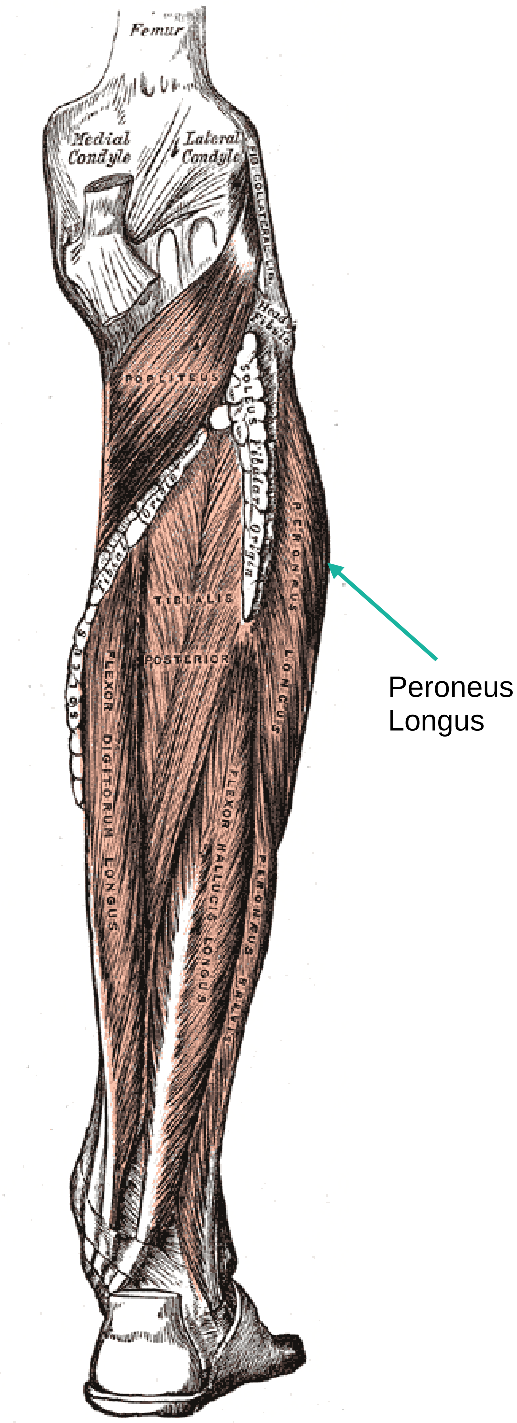
Results of Hopping Tasks; Three Seconds.

		Forward Hop			Diagonal Hop			Lateral Hop		
		Neutral	Pronated	Supinated	Neutral	Pronated	Supinated	Neutral	Pronated	Supinated
Peak (cm)	A	5.4(1.9)	6.2(1.5)	5.6(1.4)	6.9(1.5)	7.7(2.0)	6.4(1.5)	8.6(2.0)	9.6(1.1)	9.1(2.1)
	L	1.8(0.5)	2.0(0.7)	2.1(1.4)	2.8(1.2)	2.8(1.1)	2.2(0.5)	3.1(1.3)	2.7(0.8)	2.3(0.4)
	P	4.8(4.0)	5.8(5.0)	3.0(0.9)	2.7(0.5)	3.5(1.9)	2.8(0.6)	3.1(0.8)	3.1(0.6)	2.9(0.9)
	M	3.1(1.4)	2.3(0.8)	3.1(1.6)	2.1(1.0)	2.0(1.1)	2.3(0.7)	1.9(0.4)	2.0(0.5)	2.4(0.8)
Mean (cm)	A	1.6(0.7)	1.7(0.6)	1.9(0.5)	1.5(0.4)	1.7(0.5)	1.6(0.3)	1.7(0.5)	1.8(0.5)	1.5(0.3)
	L	0.7(0.2)	0.8(0.2)	0.6(0.2)	0.8(0.2)	0.8(0.2)	0.9(0.3)	0.8(0.1)	0.9(0.3)	0.8(0.2)
	P	1.4(0.4)	1.4(0.6)	1.2(0.4)	1.0(0.3)	1.4(0.4)	1.1(0.4)	1.2(0.4)	1.3(0.3)	1.3(0.3)
	M	0.7(0.2)	0.8(0.2)	0.7(0.2)	0.8(0.1)	0.7(0.2)	0.9(0.3)	0.8(0.2)	0.8(0.2)	0.8(0.3)
Total (cm)	A	26.5(2.9)	26.2(4.4)	28.0(5.1)	27.9(3.1)	28.0(2.7)	28.5(3.1)	28.8(3.1)	29.1(3.0)	30.2(3.9)
	L	17.6(2.0)	18.1(4.3)	18.2(3.6)	18.7(2.7)	19.8(3.8)	19.6(3.0)	20.4(2.2)	19.6(3.6)	19.5(1.8)
	P	25.6(6.0)	19.2(2.8)	22.3(4.8)	21.3(3.9)	22.5(7.3)	21.1(3.3)	20.4(3.3)	19.0(4.4)	21.3(4.3)
	M	19.2(2.6)	19.0(3.3)	19.6(2.0)	18.1(1.6)	19.0(3.7)	19.4(1.2)	18.5(2.2)	18.4(3.6)	19.2(2.4)
T. Dist (cm)		72.5(6.9)	64.8(8.5)	69.0(10.6)	68.4(7.7)	68.8(9.5)	71.4(8.8)	69.8(5.8)	68.3(10.7)	70.8(8.0)
V.Avg (cm/s)		23.8(2.7)	23.2(5.3)	23.9(4.4)	22.8(2.5)	23.9(4.4)	23.8(2.9)	23.3(1.9)	23.7(4.7)	24.3(3.4)
95% EA (cm^2)		29.4(13.3)	27.3(5.5)	27.5(11.3)	28.1(5.4)	28.2(8.3)	30.4(10.5)	35.0(6.6)	38.7(12.1)	37.9(11.1)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

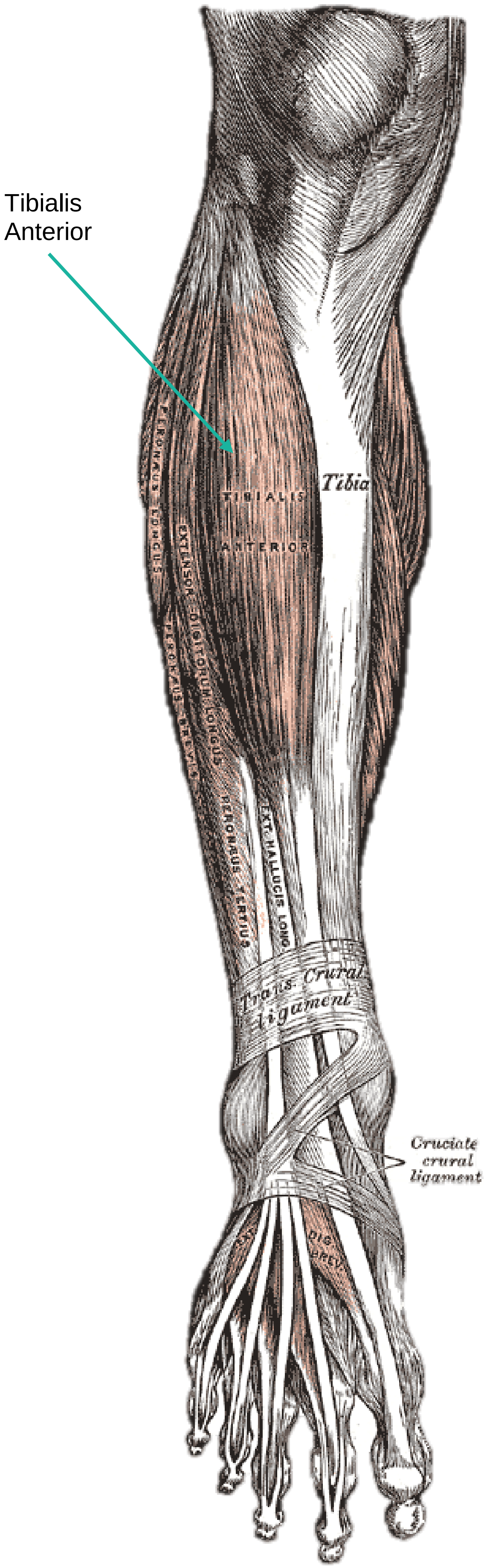
Extract from Gray (1918).

“The **Peroneus longus** is situated at the upper part of the lateral side of the leg, and is the more superficial of the peroneal muscles. It arises from the head and upper two-thirds of the lateral surface of the body of the fibula, from the deep surface of the fascia, and from the intermuscular septa between it and the muscles on the front and back of the leg; occasionally also by a few fibres from the lateral condyle of the tibia. Between its attachments to the head and to the body of the fibula there is a gap through which the common peroneal nerve passes to the front of the leg. It ends in a long tendon, which runs behind the lateral malleolus, in a groove common to it and the tendon of the peroneus brevis, behind which it lies; the groove is converted into a canal by the superior peroneal retinaculum, and the tendons in it are contained in a common mucous sheath. The tendon then extends obliquely forward across the lateral side of the calcaneus, below the trochlear process, and the tendon of the peroneus brevis, and under cover of the inferior peroneal retinaculum. It crosses the lateral side of the cuboid, and then runs on the under surface of that bone in a groove which is converted into a canal by the long plantar ligament; the tendon then crosses the sole of the foot obliquely, and is inserted into the lateral side of the base of the first metatarsal bone and the lateral side of the first cuneiform. Occasionally it sends a slip to the base of the second metatarsal bone. The tendon changes its direction at two points: first, behind the lateral malleolus; secondly, on the cuboid bone; in both of these situations the tendon is thickened, and, in the latter, a sesamoid fibrocartilage (sometimes a bone), is usually developed in its substance”.



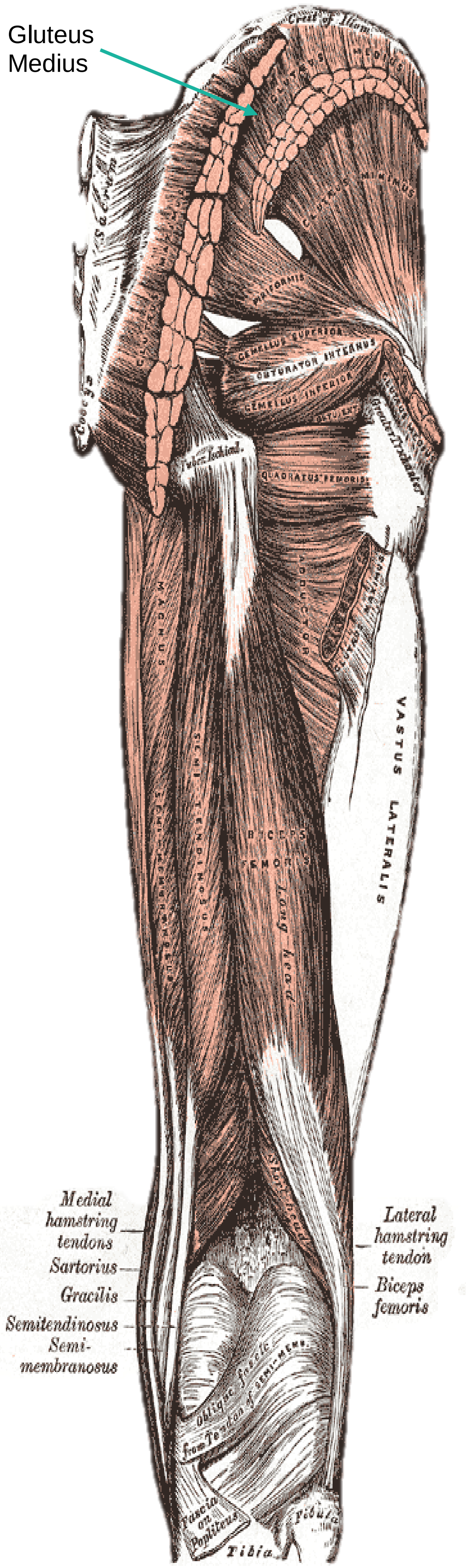
Extract from Gray (1918).

"The **Tibialis anterior** is situated on the lateral side of the tibia; it is thick and fleshy above, tendinous below. It arises from the lateral condyle and upper half or two-thirds of the lateral surface of the body of the tibia; from the adjoining part of the interosseous membrane; from the deep surface of the fascia; and from the intermuscular septum between it and the extensor digitorum longus. The fibres run vertically downward, and end in a tendon, which is apparent on the anterior surface of the muscle at the lower third of the leg. After passing through the most medial compartments of the transverse and cruciate crural ligaments, it is inserted into the medial and under surface of the first cuneiform bone, and the base of the first metatarsal bone. This muscle overlaps the anterior tibial vessels and deep peroneal nerve in the upper part of the leg".

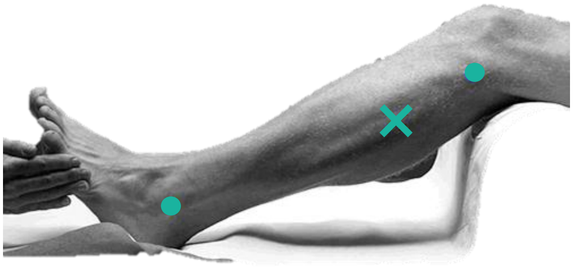
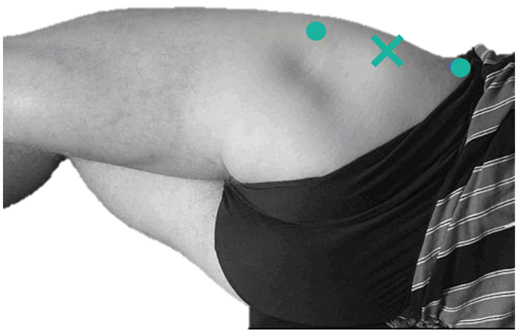


Extract from Gray (1918).

“The **Gluteus medius** is a broad, thick, radiating muscle, situated on the outer surface of the pelvis. Its posterior third is covered by the gluteus maximus, its anterior two-thirds by the gluteal aponeurosis, which separates it from the superficial fascia and integument. It arises from the outer surface of the ilium between the iliac crest and posterior gluteal line above, and the anterior gluteal line below; it also arises from the gluteal aponeurosis covering its outer surface. The fibres converge to a strong flattened tendon, which is inserted into the oblique ridge which runs downward and forward on the lateral surface of the greater trochanter. A bursa separates the tendon of the muscle from the surface of the trochanter over which it glides”.



SENIAM guidelines for Electrode Placement

Muscle	Electrode Placement	
Peroneus Longus	Electrodes need to be placed at 25% of 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	Electrodes need to be placed at 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% of the line from the iliac crest to the greater trochanter; in the direction of the between the two points.	

Adapted from SENIAM: www.seniam.org; indicating anatomical landmarks and positioning of electrodes.

	TIME	GM	TA	PL	Tilt	Change
4357	4.356	0.0015	0.0015	0.0015	23	4.356
4358	4.357	0.0015	0.0015	0.0015	23	4.357
4359	4.358	0.0015	0.0015	0.0015	23	4.358
4360	4.359	0.0022	0.0015	0.0015	23	4.359
4361	4.36	0.0022	0.0015	0.0015	23	4.36
4362	4.361	0.0022	0.0015	0.0015	23	4.361
4363	4.362	0.0022	0.0015	0.0037	23	4.362
4364	4.363	0.0022	0.0015	0.0045	23	4.363
4365	4.364	0.003	0.0015	0.0052	23	4.364
4366	4.365	0.003	0.0015	0.0052	23	4.365
4367	4.366	0.003	0.0015	0.0052	23	4.366
4368	4.367	0.0037	0.0015	0.0067	23	4.367
4369	4.368	0.0037	0.0015	0.0082	23	4.368
4370	4.369	0.003	0.0015	0.0097	23	4.369
4371	4.37	0.003	0.0022	0.0142	23	4.37
4372	4.371	0.003	0.0022	0.0157	23	4.371
4373	4.372	0.003	0.0022	0.015	23	4.372
4374	4.373	0.003	0.0022	0.015	23	4.373
4375	4.374	0.003	0.0022	0.0172	23	4.374
4376	4.375	0.0022	0.0022	0.018	23	4.375
4377	4.376	0.0015	0.003	0.0187	23	4.376
4378	4.377	0.0015	0.003	0.018	23	4.377
4379	4.378	0.0015	0.003	0.0202	23	4.378
4380	4.379	0.0022	0.0037	0.0202	23	4.379
4381	4.38	0.0045	0.0112	0.0277	23	4.38
4382	4.381	0.006	0.015	0.0337	23	4.381
4383	4.382	0.0075	0.015	0.0337	23	4.382
4384	4.383	0.0097	0.015	0.0337	23	4.383
4385	4.384	0.0105	0.015	0.0412	23	4.384
4386	4.385	0.0105	0.015	0.0615	23	4.385
4387	4.386	0.0105	0.0157	0.0757	23	4.386
4388	4.387	0.0112	0.0165	0.0885	23	4.387
4389	4.388	0.0112	0.0165	0.0907	23	4.388
4390	4.389	0.015	0.0165	0.0907	23	4.389
4391	4.39	0.0217	0.0187	0.09	23	4.39
4392	4.391	0.033	0.0277	0.0945	23	4.391
4393	4.392	0.0472	0.0315	0.0945	23	4.392
4394	4.393	0.0547	0.0315	0.1012	23	4.393
4395	4.394	0.0607	0.0315	0.108	23	4.394

Sheet1 Sheet2 Sheet3

Tilt Onset

4.317	Time of row where tilt onset occurs
#A4318	Hyperlink to tilt onset
4318	Cell where value in I changes indicating tilt onset

Baseline (mV)

GM	TA	PL	
0.0014	0.0015	0.0033	Average of cells 150ms prior to tilt onset
0.0021	0.0018	0.0053	3*SD of cells 150ms prior to tilt onset

Conditional formatting set to highlight cells after tilt exceeding avg + 3*SD

Muscular Contraction Onset (ms)

GM	TA	PL	
E4381	F4380	G4370	Cell reference for cells after tilt onset exceeding avg +3*SD of baseline for 25 consecutive cells
4381	4380	4370	As above

Reaction Time (ms)

GM	TA	PL	
63	62	52	Muscular contraction onset - tilt onset

Early Burst

25 Consecutive cells above 3SD of baseline

Excel Template Used To Calculate Reaction Time

The Effects of Foot Structure on Muscular Reaction Time

Joanna Denyer, Naomi Hewitt and Dr Andrew Mitchell

Sport, Health & Exercise Research Group, University of Hertfordshire, Hatfield, AL10 9AB, United Kingdom.

INTRODUCTION

Medial longitudinal arch abnormalities are frequently associated with lower limb injuries;[1-2] the most prevalent of which are lateral ankle sprains, with both low arches[1] and high arches[2] shown to be a risk factor. In the UK as many as 302,000 patients attend Emergency Departments with ankle sprains each year.[3] There is a clear demand for further research in this area both in terms of treatment, and in providing suggestions to reduce the rate of ankle sprains.

The measurement of muscle reaction time to a tilt platform perturbation is a well-established method of analysing lower limb neuromuscular control.[4] It is designed to simulate the mechanics of an inversion ankle sprain, therefore stressing the dynamic defence mechanism.[4] Typically, electromyography (EMG) measures include the peroneus longus,[4] and the tibialis anterior[4] as these muscles have a direct role in the dynamic defence mechanism. The gluteus medius has also been included to identify the use of hip strategy during perturbation.

No previous study has assessed the effects of pronated and supinated foot structures on muscle reaction time.

PURPOSE

To determine whether pronated and supinated foot structures contribute to neuromuscular deficits as measured by muscle reaction time to a simulated ankle sprain mechanism.

METHODS

Thirty volunteers participated in this study and were categorised into three groups which were dependant on navicular drop height measures. All measurements were taken by the same researcher as shown to be reliable during pilot testing (ICC = .97, SEM = .5mm). The average of three measurements was used to categorise subjects where ≥10mm indicated a pronated foot, 5-9mm indicated those with neutral arches and ≤4mm indicated a supinated foot. Subject characteristics were as follows; neutral group age: 20.2 ± 2.1years, height: 169.3 ± 8.8cm, weight: 70 ± 11.2kg; pronated group age: 21.4 ± 1.5years, height: 169.8 ± 10.5cm, weight: 70.4 ± 14.6kg; supinated group age: 20.8 ± 1.6years, height: 169.3 ± 6.2cm, weight: 69.5 ± 10.7kg.

The tilt platform, as previously used by Mitchell et al.[4] comprised of two independently moveable platforms, each moving from a neutral position to 30° inversion and 20° of plantar flexion when released. A diagram of the tilt platform used is shown in Figure 1.

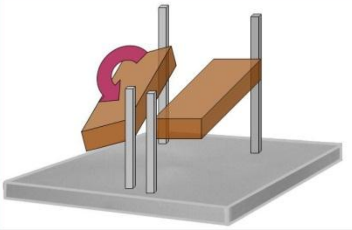


Figure 1. Tilt Platform

Subjects were asked to stand on the tilt platform in a relaxed stance, looking directly ahead Subjects were informed that between 6 and 8 tilts would occur and either ankle may be tested at any one time. Subjects were not told when the perturbations would occur, and they were initiated at variable intervals to reduce anticipatory effects. An average of three tilts on the dominant side was used for analysis. Electrodes (Biometrics SX230) were placed on the tibialis anterior, the peroneus longus, and the gluteus medius.

DATA REDUCTION AND STATISTICAL ANALYSIS

Raw EMG data was processed with an RMS filter using a 10ms moving window. Data was then reduced using a custom made Excel template (Microsoft Office). Muscle reaction time was defined as the time between the onset of the tilt mechanism, to the onset of the EMG signal when it reached a level of 3SD above the baseline for 25ms consecutively. The baseline value was the average value recorded over 150ms immediately prior to the onset of the tilt mechanism.

As data were normally distributed (Shapiro-Wilk $P > .05$) a separate one-way analysis of variance was performed for each dependant variable (peroneus longus, tibialis anterior, gluteus medius) with foot structure (neutral, pronated and supinated) as the independent variable. The a priori alpha level was set at .05.

Table 1. Reaction Time Measurements (ms)

Muscle	Neutral Group	Pronated Group	Supinated Group
Peroneus Longus	39.6 ± 5.1*	49.7 ± 9.5*	47.2 ± 5.8*
Gluteus Medius	52.0 ± 10.2	54.0 ± 10.9	47.8 ± 7.2
Tibialis Anterior	43.6 ± 8.3	45.7 ± 6.4	49.2 ± 4.3

Results are mean± SD; *Indicates significant difference to neutral group ($P < .05$)

RESULTS

Results are shown in Table 1. Across the three groups, the peroneus longus had a reaction time range of 39-49ms, the tibialis anterior ranged from 43-49ms, and the gluteus medius ranged from 47-54ms.

Analysis of the peroneus longus indicated a significant main effect of foot structure $F(2, 27) = 5.5$, $P = .01$, $\eta p2 = .29$, observed power = .82. Post hoc testing revealed a significant difference between the neutral and both the pronated group ($P = .01$) and the supinated group ($P = .04$), with both groups showing a slower peroneal reaction time in comparison to the neutral group. No significant differences were identified within the tibialis anterior $F(2, 27) = 1.9$, $P = .17$, $\eta p2 = .12$, observed power = .35 or the gluteus medius $F(2, 27) = 1.1$, $P = .35$, $\eta p2 = .07$, observed power = .22.

CONCLUSIONS

The results of this study suggest that those with pronated and supinated foot structures have slower muscle reaction time than those with neutral feet. In light of the results of this study, future research should address whether those with pronated or supinated foot structure are more at risk of obtaining a lateral ankle sprain than those with neutral foot structures. Researchers investigating other aspects of muscle reaction time should now control for foot type as differences among participants may interfere with results.

REFERENCES

1. Mei-Dan, O. Kahn, G., Zeev, A., Rubin, A., Constantini, N., Even, A. (2005). The medial longitudinal arch as a possible risk factor for ankle sprains. *Foot and Ankle International, 26*(2), 180-183.
2. Williams, D. S., McClay, I. S. & Hamill, J. (2001). Arch structure and injury patterns in runners. *Clinical Biomechanics, 16*, 341-347.
3. 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 based practice for severe ankle sprains. *Emergency Medicine Journal, 20*, 508-510.
4. Mitchell, A., Dyson, R., Hale, T. & Abraham, C. (2008). Biomechanics of ankle instability. part 1: reaction time to simulated ankle sprain. *Medicine and Science in Sports and Exercise, 40*(8), 1515-1521.

University of Hertfordshire

Award winning poster at University of Hertfordshire Life Sciences Research Day 2012

Navicular Drop Height Measures; Raw Data

Navicular Drop Height (mm)		Foot Type
Right Foot	Left Foot	
8	10	Neutral & Pronated
11	9	Neutral & Pronated
8	15	Neutral & Pronated
9	10	Neutral & Pronated
11	8	Neutral & Pronated
9	11	Neutral & Pronated
5	1	Neutral & Supinated
7	4	Neutral & Supinated
6	3	Neutral & Supinated
2	5	Neutral & Supinated
14	14	Pronated
11	10	Pronated
10	11	Pronated
11	10	Pronated
12	12	Pronated
10	11	Pronated
13	13	Pronated
9	9	Neutral
5	5	Neutral
6	5	Neutral
8	6	Neutral
9	9	Neutral
7	6	Neutral
6	9	Neutral
3	1	Supinated
2	4	Supinated
4	2	Supinated
3	3	Supinated
2	2	Supinated
1	2	Supinated
0	0	Supinated
2	2	Supinated

THE EFFECTS OF SHORT TERM ANTI-PRONATION TECHNIQUES ON PEAK PLANTAR PRESSURES DURING DYNAMIC MOVEMENTS

Joanna Denyer and Dr Andrew Mitchell

Sport, Health & Exercise Research Group, University of Hertfordshire, Hatfield, AL10 9AB, United Kingdom.

INTRODUCTION

Over-pronation is one of the most common lower limb disorders, and has been related to a variety of injuries including ankle sprains, knee injuries, stress fractures and lower back pain (Kaufman et al., 1999; Williams et al., 2001). Anti-pronation techniques (APT) such as taping and shop-bought orthotics are commonly used to reduce symptoms associated with over-pronation; they are often used as an indicator to see if expensive custom made orthotics are necessary to purchase (Smith, Brooker, Vicenzino & McPoil, 2004).

Previous research on APT assess the distribution of force across the plantar aspect of the foot when standing (Ator, Gunn, McPoil & Knecht, 1991; Russo & Chipchase, 2001), however past research is inconclusive as to which technique is most effective. Additionally, few previous studies have measured the effects of APT during sport specific movements, making current findings difficult to apply to athletes.

PURPOSE

The purpose of this study was to compare the effects of three anti-pronation techniques; augmented low-Dye taping (ALD), double X-arch taping (DXA) and shop bought orthotics over two sports specific movements; a forward run, and a lateral change of direction.

METHODS

Twelve active female participants (age 21 years ±0.79, height 168.59cm ±7.16 and mass 70.71kg ±14.93) with over-pronated feet (navicular drop height ≥10 mm) volunteered to participate in this study. Ethical approval was granted from the Institutional Ethics Committee. The control and three interventions (ALD, DXA and orthotic) were were applied to each participant on the dominant leg in a randomised order. Both taping techniques are shown in figures 1 and 2, and the orthotic (Regular Orthaheel Orthotics, Vasyli International) is shown in figure 3.

Peak plantar pressures (PPP) were measured on an RS Scan (RS Scan International, Belgium) during two sports specific movements; a forward run, and lateral change of direction. The forward run involved a 8m run through timing gates across the foot scan. The lateral change of direction involved an 8m run up through timing gates towards the foot scan, then planting the dominant foot on the foot scan before side stepping to the laterally for 4m through a second set of timing gates. Three measurements were taken before applying the next APT.

Figure 1. ALD Taping

Figure 2. DXA Taping

Figure 3. Regular Orthaheel Orthotics

RESULTS

During the forward run movement, the ALD resulted in a significant reduction in total medial PPP in comparison to the control (p=0.033). During the lateral change of direction movement, both taping techniques resulted in significantly lower total medial PPP than the control (ALD p=0.005; DXA p=0.010). Figures 4 and 5 show total shifts in medial and lateral PPP; the steeper the trend line, the more effective the anti-pronating technique.

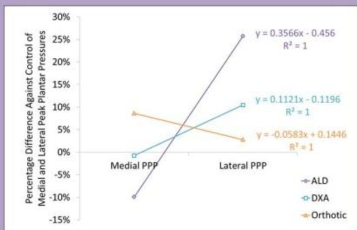


Figure 4. Total shifts in PPP showing trend line equations for each APT during a forward run.

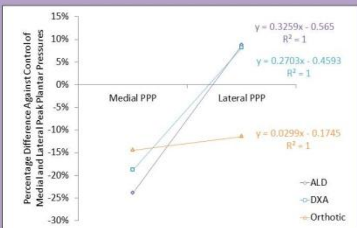


Figure 5. Total shifts in PPP showing trend line equations for each APT during a lateral change of direction.

CONCLUSIONS

These results suggest that the ALD taping is the most effective technique in reducing over-pronation as indicated by a reduction in total medial PPP, in two dynamic movements; a forward run and a lateral change of direction. The DXA taping also proved to be moderately effective, however the orthotic was not shown to be effective in either dynamic movement.

The mechanism by which anti-pronation taping produces a lateral shift is explained by Russo and Chipchase (2001); it is thought that the tape prevents the medial longitudinal arch from lowering towards the ground as normal during gait, which in turn causes an overcompensation of pressure on the lateral side.

The results of this study are consistent with previous research which has also shown a decrease in medial PPP with anti-pronation taping during gait (Russo & Chipchase, 2001), however further research is required to understand the function of anti-pronation taping in dynamic movements.

PRACTICAL APPLICATION

Taping should be considered rather than orthotics as an immediate treatment for symptoms associated with over-pronation or as a preventative measure for asymptomatic athletes with over pronated feet.

REFERENCES

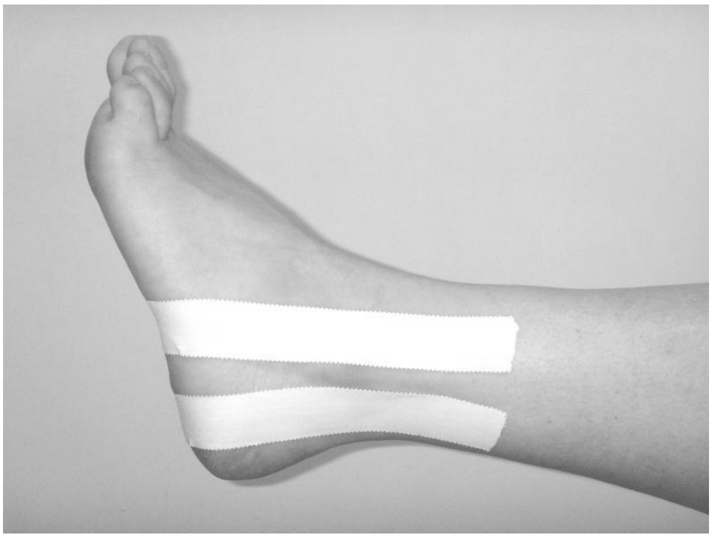
1. Kaufman, K. R., Brodine, S. K., Shaffer, R. A., Johnson, C. W. & Cullison, T. R. (1999). The effect of foot structure and range of motion on musculoskeletal overuse injuries. *American Journal of Sports Medicine, 27*(5), 585-593.
2. Williams, D. S., McClay, I. S. & Hamill, J. (2001). Arch structure and injury patterns in runners. *Clinical Biomechanics, 16*, 341-347.
3. Smith, M., Brooker. S., Vicenzino, B & McPoil, T. (2004). Use of anti-pronation taping to assess suitability of orthotic prescription: case report. *Australian Journal of Physiotherapy, 50*(2), 111-113.
4. Ator, R., Gunn, K., McPoil, T. G. & Knecht, H. G. (1991). The effect of adhesive strapping on medial longitudinal arch support before and after exercise. *Journal of Orthopedic and Sports Physical Therapy, 14*, 18-23.
5. Russo, S. J. & Chipchase, L. S. (2001). The effect of low-Dye taping on peak plantar pressures of normal feet during gait. *Australian Journal of Physiotherapy, 47*(4), 239-244.

University of Hertfordshire

Poster presented at British Association of Sport and Exercise Sciences (BASES) Conference, University of Essex, 6-8th September 2011.


The Modified Low-Dye Arch Taping Technique


The Closed Basket Weave Ankle Taping Technique


The Proprioceptive Taping Technique

Results of Forward Hop Task; Neutral Group

		Neutral (200 ms)				Neutral (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	2.4(1.7)	2.7(1.5)	2.5(1.3)	2.1(1.2)	2.9(1.4)	3.1(1.3)	2.7(1.1)	2.5(0.9)
	L	1.5(1.2)	1.4(1.2)	2.6(2.7)	0.8(0.6)	2.1(0.9)	2.1(0.9)	3.1(2.4)	1.6(0.5)
	A	5.0(1.0)	4.6(1.9)	4.5(1.6)	5.3(1.2)	5.5(0.9)	5.3(1.4)	5.3(1.0)	5.6(1.2)
	P	2.8(2.1)	4.9(3.6)	6.0(4.0)	2.6(2.1)	4.0(1.7)	5.4(3.1)	6.6(3.6)	3.7(1.6)
Mean (cm)	M	1.1(0.7)	0.9(0.3)	0.8(0.4)	0.7(0.2)	0.8(0.2)	0.7(0.1)	0.7(0.2)	0.8(0.1)
	L	0.6(0.3)	0.6(0.5)	0.7(0.3)	0.3(0.2)	0.9(0.3)	0.8(0.1)	0.7(0.1)	0.9(0.2)
	A	2.5(0.7)	2.5(0.7)	2.1(0.9)	3.0(0.8)	1.3(0.5)	1.6(0.8)	1.4(0.5)	1.7(0.5)
	P	1.3(0.9)	2.0(1.4)	2.3(1.6)	1.4(1.0)	1.3(0.5)	1.1(0.2)	1.2(0.5)	1.4(0.3)
Total (cm)	M	5.1(1.9)	5.5(1.4)	6.3(1.2)	5.0(2.2)	16.7(3.4)	16.3(2.5)	16.9(4.0)	16.2(1.2)
	L	4.6(2.1)	4.7(3.2)	6.8(4.4)	3.9(1.5)	19.4(1.2)	18.1(2.5)	19.3(1.5)	18.2(2.1)
	A	8.7(2.5)	8.6(1.3)	10.5(4.9)	8.6(2.2)	28.3(3.8)	25.3(4.2)	24.9(4.8)	26.6(3.6)
	P	7.2(3.0)	8.9(4.5)	11.7(4.0)	7.6(3.6)	20.4(3.1)	21.3(5.7)	21.9(3.0)	20.8(3.6)
T. Dist (cm)		21.1(3.9)	23.7(8.4)	27.7(9.4)	20.1(6.3)	67.4(7.4)	64.3(10.1)	65.8(7.0)	65.3(8.2)
V.Avg (cm/s)		105.7(19.3)	106.2(29.1)	115.1(29.5)	100.7(31.6)	22.5(2.5)	21.4(3.4)	21.9(2.3)	21.8(2.7)
95% EA (cm^2)		19.0(10.0)	37.8(46.4)	30.0(21.5)	17.0(10.1)	31.1(12.4)	30.2(9.0)	24.7(8.3)	37.7(11.8)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape.

Results of Forward Hop Task; Pronated Group

		Pronated (200 ms)				Pronated (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	2.0(1.2)	2.3(1.2)	2.6(1.8)	2.8(1.4)	2.4(1.0)	2.5(1.1)	2.8(1.6)	3.0(1.3)
	L	1.3(0.4)	1.2(0.9)	1.8(2.0)	1.2(0.7)	1.8(0.5)	1.9(0.7)	2.3(2.0)	1.8(0.7)
	A	4.6(1.6)	3.7(1.5)	3.7(1.9)	4.9(1.7)	5.2(1.3)	4.6(1.2)	4.7(1.6)	5.4(1.2)
	P	3.8(2.7)	5.0(4.1)	7.1(5.5)	4.4(3.8)	4.6(2.6)	6.0(3.3)	8.0(5.0)	5.2(3.2)
Mean (cm)	M	0.7(0.4)	0.8(0.5)	0.7(0.4)	0.8(0.3)	0.7(0.2)	0.7(0.1)	0.7(0.3)	0.8(0.2)
	L	0.5(0.2)	0.5(0.3)	0.8(1.2)	0.6(0.4)	0.7(0.1)	0.7(0.1)	1.2(1.2)	0.9(0.2)
	A	2.5(0.9)	1.8(0.6)*	1.8(0.9)*	2.6(1.2)	1.6(0.7)	1.4(0.4)	1.5(0.6)	1.6(0.5)
	P	1.7(1.1)	2.1(1.6)	2.6(0.9)	2.0(1.2)	1.2(0.3)	1.3(0.4)	1.6(0.4)	1.2(0.3)
Total (cm)	M	5.3(1.6)	4.9(1.9)	6.8(4.6)	5.9(3.2)	15.6(1.3)	16.6(4.0)	18.4(7.6)	17.4(4.3)
	L	3.9(1.6)	3.6(1.7)	4.8(3.5)	3.9(2.0)	17.9(2.8)	18.7(4.2)	21.1(7.5)	19.0(3.8)
	A	9.0(2.8)	7.3(3.5)	7.4(3.4)	8.6(3.4)	27.2(5.8)	24.7(7.1)	27.0(9.9)	27.6(5.9)
	P	8.2(2.6)	9.3(4.1)	12.0(5.6)	8.9(4.9)	20.8(4.0)	23.2(5.0)	25.6(5.4)	21.3(5.4)
T. Dist (cm)		21.4(5.3)	20.1(6.1)	25.3(10.4)	21.9(9.4)	68.5(11.2)	65.6(16.2)	69.1(14.9)	65.9(11.9)
V.Avg (cm/s)		96.0(14.7)	94.5(22.6)	109.4(31.2)	93.1(29.2)	22.8(3.7)	21.9(5.4)	24.0(7.4)	22.5(4.8)
95% EA (cm^2)		17.8(9.0)	19.0(11.3)	25.9(29.3)	26.5(11.9)	30.3(15.0)	25.9(15.0)	32.4(11.2)	27.2(10.3)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape. * Indicates significant difference in postural stability in comparison to no-tape control (P < .016).

Results of Forward Hop Task; Supinated Group

		Supinated (200 ms)				Supinated (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	2.2(1.8)	3.4(1.5)	2.4(1.3)	2.2(1.9)	2.6(1.5)	3.5(1.3)	2.7(1.2)	2.8(1.3)
	L	1.4(0.6)	0.9(0.6)	1.2(0.5)	1.0(0.6)	1.9(0.4)	1.6(0.4)	1.5(0.4)	1.9(0.4)
	A	5.4(1.8)	4.6(1.9)	4.6(2.6)	5.2(2.2)	5.9(1.5)	5.4(1.4)	5.3(2.1)	5.8(1.8)
	P	3.4(5.7)	3.1(4.7)	5.0(4.6)	3.0(5.0)	4.4(5.3)	3.9(4.4)	5.7(4.1)	4.3(4.6)
Mean (cm)	M	0.7(0.4)	0.9(0.4)	1.0(0.5)	0.7(0.5)	0.9(0.2)	0.8(0.2)	0.7(0.2)	0.8(0.2)
	L	0.6(0.2)	0.4(0.3)	0.5(0.2)	0.3(0.2)	0.8(0.2)	0.8(0.2)	0.8(0.1)	0.8(0.2)
	A	2.7(1.0)	2.4(0.8)	1.9(1.0)*	2.6(1.1)	1.6(0.4)	1.5(0.2)	1.4(0.3)	1.5(0.4)
	P	1.3(2.2)	1.3(1.6)	1.8(1.6)	1.3(1.9)	1.1(0.4)	1.1(0.2)	1.3(0.5)	1.1(0.5)
Total (cm)	M	5.0(1.8)	6.3(1.2)	5.9(1.8)	4.8(1.3)	18.1(2.0)	17.6(2.0)	16.7(1.7)	17.0(2.0)
	L	3.5(2.2)	3.1(1.3)	4.3(1.1)	3.0(1.6)	19.4(3.7)	18.9(3.8)	19.3(2.5)	19.0(3.0)
	A	7.4(2.9)	6.9(2.3)	8.1(2.9)	6.7(2.4)	25.4(4.1)	24.8(4.1)	26.4(2.5)	26.1(2.0)
	P	7.5(7.1)	7.4(5.5)	9.8(4.9)	6.6(5.4)	19.8(6.4)	19.3(5.2)	21.5(5.0)	20.3(6.1)
T. Dist (cm)		18.7(6.8)	19.0(4.1)	22.5(4.7)	17.0(4.8)	65.4(11.0)	63.8(9.1)	66.7(6.9)	65.2(8.5)
V.Avg (cm/s)		74.9(10.9)	95.2(20.7)	112.3(23.7)*	85.1(23.8)	21.8(3.7)	21.3(3.0)	22.2(2.3)	21.7(2.8)
95% EA (cm^2)		26.5(19.8)	20.2(9.9)	17.3(10.7)	15.5(8.3)	31.0(11.2)	25.8(4.9)	26.9(9.4)	29.2(13.3)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape. * Indicates significant difference in postural stability in comparison to no-tape control ($P < .016$).

Results of Diagonal Hop Task; Neutral Group

		Neutral (200 ms)				Neutral (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	1.4(1.5)	1.2(0.8)	1.8(2.0)	0.9(0.3)	2.2(1.2)	1.9(0.7)	2.4(1.8)	1.8(0.2)
	L	4.0(2.0)	3.0(2.2)	3.6(1.7)	2.9(0.8)	4.3(1.7)	3.3(1.9)	3.7(1.6)	3.1(0.7)
	A	8.5(2.2)	6.1(2.8)	6.2(3.2)	7.1(1.3)	8.5(2.2)	6.5(2.3)	6.3(3.1)	7.2(1.2)
	P	1.9(0.9)	3.8(3.0)	4.3(3.8)	2.2(1.3)	3.6(1.0)	4.6(3.2)	5.2(3.1)	3.9(1.4)
Mean (cm)	M	0.6(0.5)	0.4(0.3)	0.7(0.4)	0.6(0.5)	0.8(0.2)	0.7(0.1)	0.7(0.2)	0.8(0.1)
	L	1.1(0.6)	0.8(0.4)	0.9(0.3)	1.1(0.2)	0.9(0.3)	0.8(0.1)	0.7(0.1)	0.9(0.2)
	A	3.8(0.9)	3.0(1.3)	3.3(0.9)	4.4(0.4)	1.3(0.5)	1.6(0.8)	1.4(0.5)	1.7(0.5)
	P	0.8(0.7)	1.6(1.2)	2.0(1.8)	1.1(0.9)	1.3(0.5)	1.1(0.2)	1.2(0.5)	1.4(0.3)
Total (cm)	M	3.6(2.3)	3.0(1.9)	3.6(3.1)	3.0(0.7)	16.7(3.4)	16.3(2.5)	16.9(4.0)	16.2(1.2)
	L	6.0(1.9)	5.5(1.8)	6.4(1.7)	5.0(1.4)	19.4(1.2)	18.1(2.5)	19.3(1.5)	18.2(2.1)
	A	13.1(3.2)	9.6(3.1)	9.2(3.4)	10.5(3.5)	28.3(3.8)	25.3(4.2)	24.9(4.8)	26.6(3.6)
	P	5.1(1.8)	6.7(4.0)	8.3(3.4)	5.8(1.9)	20.4(3.1)	21.3(5.7)	21.9(3.0)	20.8(3.6)
T. Dist (cm)		22.4(5.0)	19.9(5.0)	22.1(3.5)	19.9(3.9)	67.4(7.4)	64.3(10.1)	65.8(7.0)	65.3(8.2)
V.Avg (cm/s)		112.2(25.1)	99.7(24.8)	110.5(17.5)	99.6(19.6)	22.5(2.5)	21.4(3.4)	21.9(2.3)	21.8(2.7)
95% EA (cm^2)		41.7(16.5)	32.2(13.4)	40.7(18.6)	27.4(9.2)	31.1(12.4)	30.2(9.0)	24.7(8.3)	37.7(11.8)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape.

Results of Diagonal Hop Task; Pronated Group

		Pronated (200 ms)				Pronated (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	1.1(0.5)	1.0(0.4)	2.1(1.2)	1.2(0.6)	1.8(0.6)	1.7(0.3)	2.0(0.7)	2.0(0.4)
	L	2.8(0.5)	3.2(0.8)	4.4(2.6)	3.1(1.4)	2.8(0.4)	2.9(0.7)	3.5(1.5)	2.8(1.3)
	A	6.7(1.3)	6.1(3.2)	5.2(3.7)	7.1(1.5)	6.9(1.5)	6.3(2.7)	5.4(3.0)	7.2(1.3)
	P	2.8(1.8)	4.2(4.2)	5.5(4.6)	2.1(2.3)	3.4(1.5)	3.9(1.9)	6.0(3.8)	3.1(1.9)
Mean (cm)	M	0.5(0.3)	0.5(0.3)	0.8(0.4)	0.8(0.3)	0.7(0.2)	0.7(0.1)	0.7(0.3)	0.8(0.2)
	L	0.8(0.1)	1.2(0.5)	2.0(1.3)	1.0(0.5)	0.7(0.1)	0.7(0.1)	1.2(1.2)	0.9(0.2)
	A	3.0(0.4)	2.9(1.4)	2.3(1.5)	3.4(0.6)	1.6(0.7)	1.4(0.4)	1.5(0.6)	1.6(0.5)
	P	1.1(0.8)	1.8(1.1)	1.9(1.3)	0.8(1.0)	1.2(0.3)	1.3(0.4)	1.6(0.4)	1.2(0.3)
Total (cm)	M	3.8(2.1)	3.7(2.2)	4.9(3.4)	3.9(2.9)	15.6(1.3)	16.6(4.0)	18.4(7.6)	17.4(4.3)
	L	6.1(1.2)	5.8(2.0)	7.4(3.7)	5.9(1.6)	17.9(2.8)	18.7(4.2)	21.1(7.5)	19.0(3.8)
	A	11.4(3.5)	10.1(5.8)	10.1(5.4)	11.1(4.7)	27.2(5.8)	24.7(7.1)	27.0(9.9)	27.6(5.9)
	P	7.9(2.1)	8.6(4.0)	9.5(4.0)	6.9(4.4)	20.8(4.0)	23.2(5.0)	25.6(5.4)	21.3(5.4)
T. Dist (cm)		24.6(6.8)	21.1(10.5)	25.4(9.5)	22.9(10.2)	68.5(11.2)	65.6(16.2)	69.1(14.9)	65.9(11.9)
V.Avg (cm/s)		109.7(24.6)	93.0(25.9)	99.9(21.7)	81.3(17.0)	22.8(3.7)	21.9(5.4)	24.0(7.4)	22.5(4.8)
95% EA (cm^2)		24.4(13.3)	26.6(16.0)	32.8(23.3)	25.4(10.9)	30.3(15.0)	25.9(15.0)	32.4(11.2)	27.2(10.3)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape.

Results of Diagonal Hop Task; Supinated Group

		Supinated (200 ms)				Supinated (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	1.3(1.4)	1.1(0.9)	1.0(0.5)	1.2(0.9)	2.2(1.0)	1.8(0.6)	1.7(0.5)	1.8(0.5)
	L	3.0(1.6)	2.3(1.2)	3.1(1.2)	2.6(1.3)	3.3(1.2)	2.7(0.7)	3.3(1.1)	3.0(0.9)
	A	6.5(2.9)	6.6(3.1)	6.6(2.3)	7.4(2.7)	6.9(2.4)	6.8(2.8)	6.9(1.8)	7.8(1.9)
	P	2.7(4.1)	2.4(4.2)	2.7(3.4)	2.2(3.9)	3.5(3.8)	3.6(3.7)	4.1(3.0)	3.7(3.3)
Mean (cm)	M	0.7(0.6)	0.6(0.4)	0.6(0.3)	0.6(0.6)	0.9(0.2)	0.8(0.2)	0.7(0.2)	0.8(0.2)
	L	0.9(0.4)	0.7(0.3)	1.0(0.4)	1.1(0.6)	0.8(0.2)	0.8(0.2)	0.8(0.1)	0.8(0.2)
	A	2.8(1.1)	2.7(1.1)	2.7(0.7)	3.1(1.3)	1.6(0.4)	1.5(0.2)	1.4(0.3)	1.5(0.4)
	P	1.1(1.4)	1.0(1.4)	1.1(1.2)	1.0(1.5)	1.1(0.4)	1.1(0.2)	1.3(0.5)	1.1(0.5)
Total (cm)	M	3.6(1.2)	3.3(1.2)	2.6(0.7)	2.7(0.5)	18.1(2.0)	17.6(2.0)	16.7(1.7)	17.0(2.0)
	L	5.2(2.3)	5.1(1.9)	5.8(1.4)	4.9(1.2)	19.4(3.7)	18.9(3.8)	19.3(2.5)	19.0(3.0)
	A	9.4(3.3)	8.9(2.6)	10.2(1.7)	9.7(1.8)	25.4(4.1)	24.8(4.1)	26.4(2.5)	26.1(2.0)
	P	5.7(5.3)	5.5(4.8)	6.4(5.2)	5.3(4.5)	19.8(6.4)	19.3(5.2)	21.5(5.0)	20.3(6.1)
T. Dist (cm)		19.2(6.5)	18.3(4.9)	20.3(4.8)	18.1(3.9)	65.4(11.0)	63.8(9.1)	66.7(6.9)	65.2(8.5)
V.Avg (cm/s)		86.4(23.4)	83.5(16.5)	92.1(10.5)	90.7(19.4)	21.8(3.7)	21.3(3.0)	22.2(2.3)	21.7(2.8)
95% EA (cm^2)		36.9(14.8)	22.8(17.0)*	30.6(17.0)	31.1(19.5)	31.0(11.2)	25.8(4.9)	26.9(9.4)	29.2(13.3)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape. * Indicates significant difference in postural stability in comparison to no-tape control (P < .016).

Results of Lateral Hop Task; Neutral Group

		Neutral (200 ms)				Neutral (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	1.6(0.8)	1.8(1.1)	1.5(1.4)	1.4(0.9)	2.2(0.6)	2.4(0.7)	2.3(1.0)	2.1(0.6)
	L	3.2(1.5)	2.2(1.2)	3.4(1.1)	3.1(1.1)	3.6(1.6)	2.7(0.8)	3.6(0.9)	3.6(1.1)
	A	8.9(1.2)	9.3(1.8)	9.4(1.7)	9.4(1.5)	8.9(1.2)	9.3(1.8)	9.4(1.7)	9.5(1.5)
	P	1.9(1.0)	1.7(0.8)	1.7(0.8)	2.2(1.1)	4.0(1.3)	3.7(2.1)	4.1(1.8)	3.5(1.2)
Mean (cm)	M	1.1(0.8)	0.9(0.4)	0.8(0.7)	0.8(0.6)	0.8(0.2)	0.9(0.1)	1.0(0.4)	0.8(0.3)
	L	1.1(0.4)	0.9(0.5)	1.2(0.4)	1.1(0.4)	0.9(0.1)	0.9(0.2)	0.9(0.2)	0.8(0.3)
	A	5.0(0.8)	4.4(0.8)	4.2(0.7)	5.1(0.8)	1.7(0.5)	1.8(0.5)	1.5(0.3)	1.9(1.0)
	P	0.7(0.4)	0.9(0.8)	0.9(0.5)	1.1(0.7)	1.3(0.3)	1.5(0.6)	1.4(0.4)	1.2(0.4)
Total (cm)	M	3.3(1.1)	3.7(1.6)	3.8(2.4)	4.4(2.2)	17.1(2.4)	17.3(2.6)	17.3(3.2)	18.0(2.5)
	L	5.7(1.5)	4.5(1.3)	5.8(1.6)	6.1(1.6)	18.2(1.6)	17.5(1.7)	19.2(2.4)	19.4(1.6)
	A	11.7(2.7)	11.4(3.0)	12.8(2.8)	13.8(3.9)	28.7(3.9)	28.1(4.9)	29.7(4.5)	28.9(4.2)
	P	4.4(1.3)	4.1(1.3)	4.8(1.9)	4.5(1.4)	20.4(3.9)	19.3(5.1)	20.6(4.9)	20.7(4.8)
T. Dist (cm)		20.3(3.7)	19.1(3.5)	22.0(5.7)	23.0(5.8)	67.8(9.0)	65.5(9.8)	69.1(11.0)	69.0(9.2)
V.Avg (cm/s)		101.7(18.3)	95.6(17.4)	110.1(28.3)	108.0(27.9)	22.6(3.0)	21.8(3.3)	23.0(3.7)	23.0(3.1)
95% EA (cm^2)		46.3(19.7)	38.9(18.2)	45.9(15.4)	53.5(30.7)	36.7(6.5)	42.1(16.2)	39.9(14.8)	32.1(10.7)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape.

Results of Lateral Hop Task; Pronated Group

		Pronated (200 ms)				Pronated (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	0.9(0.4)	1.7(2.2)	1.4(0.9)	1.2(0.7)	1.8(0.3)	2.4(2.2)	1.9(1.1)	2.1(0.6)
	L	3.4(1.2)	3.8(1.7)	9.1(19.9)	3.4(0.8)	3.5(1.2)	3.9(1.6)	9.2(19.8)	3.5(0.8)
	A	9.2(2.3)	8.8(1.3)	8.8(2.4)	8.6(1.8)	9.2(2.3)	8.8(1.3)	8.9(2.4)	8.6(1.8)
	P	1.6(0.7)	2.4(0.7)	3.9(4.5)	2.4(0.8)	3.1(1.2)	3.7(1.2)	5.4(4.0)	4.0(1.6)
Mean (cm)	M	0.6(0.5)	1.0(2.0)	0.5(0.6)	0.7(0.6)	0.9(0.3)	0.8(0.3)	0.9(0.7)	0.9(0.4)
	L	1.2(0.4)	1.5(0.7)	2.1(1.4)	1.0(0.3)	0.9(0.4)	0.9(0.2)	1.4(1.3)	0.8(0.2)
	A	4.5(1.0)	4.2(0.7)	3.8(0.9)	4.4(0.3)	1.4(0.3)	1.6(0.3)	1.5(0.4)	1.6(0.3)
	P	0.8(0.6)	0.9(0.7)	1.7(1.5)	1.1(0.4)	1.4(0.7)	1.5(0.5)	1.7(0.9)	1.6(0.5)
Total (cm)	M	4.0(0.6)	4.6(3.8)	9.2(19.5)	4.5(2.4)	16.5(3.1)	18.1(4.4)	22.5(17.1)	18.8(3.8)
	L	7.1(1.4)	7.3(4.1)	11.9(20.8)	7.1(1.4)	19.0(4.2)	20.7(5.9)	25.6(18.3)	20.8(3.7)
	A	13.0(3.5)	12.3(3.8)	13.8(4.3)	12.2(4.3)	28.0(4.7)	29.2(5.6)	31.1(7.5)	30.4(8.6)
	P	5.1(1.4)	5.3(2.6)	6.9(4.9)	5.3(2.3)	18.8(4.0)	19.9(4.7)	23.0(5.9)	22.2(7.8)
T. Dist (cm)		21.5(4.9)	23.5(10.8)	23.3(7.4)	22.1(7.4)	67.3(13.4)	71.2(15.5)	71.5(20.6)	72.4(19.5)
V.Avg (cm/s)		107.4(24.6)	99.9(29.1)	115.8(33.6)	103.0(33.3)	22.4(4.4)	23.7(5.2)	28.1(10.7)	24.1(6.5)
95% EA (cm^2)		43.1(14.0)	41.4(20.2)	38.3(21.3)	39.2(7.2)	34.4(15.6)	38.4(15.5)	33.3(22.5)	39.5(19.5)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape.

Results of Lateral Hop Task; Supinated Group

		Supinated (200 ms)				Supinated (Three Seconds)			
		No-Tape	Arch Tape	Ankle Tape	Prop Tape	No-Tape	Arch Tape	Ankle Tape	Prop Tape
Peak (cm)	M	1.4(0.8)	1.2(0.7)	0.9(0.6)*	1.2(0.7)	2.2(0.7)	2.0(0.5)	1.7(0.4)	1.9(0.4)
	L	2.4(0.8)	2.7(1.5)	2.6(0.8)	2.6(0.8)	2.7(0.7)	2.9(1.4)	2.7(0.7)	2.7(0.7)
	A	8.1(0.9)	9.5(1.3)	9.5(2.0)	9.0(1.5)	8.2(1.0)	9.5(1.3)	9.5(2.0)	9.0(1.5)
	P	1.3(0.6)	1.5(0.6)	1.5(1.1)	1.2(0.6)	3.1(1.2)	2.8(0.9)	3.0(0.7)	3.2(0.6)
Mean (cm)	M	0.8(0.5)	0.8(0.5)	0.7(0.4)	0.8(0.4)	0.9(0.3)	0.8(0.1)	0.8(0.2)	0.8(0.2)
	L	0.9(0.2)	0.9(0.6)	1.0(0.3)	1.0(0.3)	0.8(0.2)	0.8(0.2)	0.7(0.1)	0.7(0.2)
	A	3.9(0.7)	4.2(0.8)	4.1(1.0)	4.1(0.7)	1.6(0.5)	1.8(0.5)	1.5(0.6)	1.5(0.4)
	P	0.6(0.3)	0.7(0.6)	0.6(0.3)	0.5(0.4)	1.4(0.7)	1.3(0.3)	1.3(0.4)	1.3(0.3)
Total (cm)	M	3.2(1.6)	3.5(1.4)	3.3(1.4)	3.1(1.3)	17.3(2.0)	18.0(4.1)	17.2(2.5)	16.9(1.6)
	L	4.9(1.3)	5.7(2.2)	5.3(1.5)	5.1(1.7)	18.7(1.8)	19.2(5.1)	18.7(2.8)	18.5(2.0)
	A	10.6(2.9)	11.5(2.3)	12.4(3.6)	11.5(2.9)	26.8(4.2)	28.2(4.4)	28.0(3.2)	27.6(2.9)
	P	4.0(2.4)	4.1(1.7)	4.5(2.1)	4.1(2.5)	19.0(4.7)	18.9(4.7)	18.9(3.1)	18.9(3.6)
T. Dist (cm)		18.1(6.0)	19.9(4.7)	20.6(5.5)	19.2(6.0)	64.7(10.0)	66.7(13.7)	65.8(7.8)	65.0(7.4)
V.Avg (cm/s)		85.4(18.4)	100.1(23.1)	97.5(15.1)	93.6(25.0)	21.6(3.3)	25.3(7.0)	21.9(2.6)	21.7(2.5)
95% EA (cm^2)		36.4(11.7)	36.8(8.3)	31.0(7.8)	36.5(17.1)	31.9(9.9)	33.9(7.4)	28.9(8.4)	31.2(9.8)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2); Prop Tape: proprioceptive tape. * Indicates significant difference in postural stability in comparison to no-tape control (P < .016).

Muscle Reaction Times for Each Taping Condition

Muscle	Taping Technique	Neutral Group	Pronated Group	Supinated Group
Peroneus Longus (ms)	No-Tape	39.6 (5.1)	49.7 (9.5)	47.2 (5.8)
	Ankle Tape	51.1 (11.3)*	58.8 (9.9)*	48.3 (5.3)
	Arch Tape	44.7 (15.6)	51.7 (7.7)	45.0 (7.3)
	Proprioceptive Tape	44.9 (11.1)	52.0 (8.4)	42.5 (4.2)
Tibialis Anterior (ms)	No-Tape	43.6 (8.3)	45.7 (6.4)	49.2 (4.3)
	Ankle Tape	46.7 (7.7)	49.7 (6.0)	46.1 (5.9)
	Arch Tape	43.7 (7.0)	52.3 (8.3)	49.8 (3.6)
	Proprioceptive Tape	51.6 (10.8)	52.8 (4.9)	46.9 (4.8)
Gluteus Medius (ms)	No-Tape	52.0 (10.2)	54.0 (10.9)	47.8 (7.2)
	Ankle Tape	58.0 (6.7)	55.9 (11.9)	47.6 (7.5)
	Arch Tape	48.6 (9.3)	54.9 (10.7)	48.6 (7.0)
	Proprioceptive Tape	51.8 (15.8)	53.5 (13.0)	53.7 (8.7)

Values are mean (SD). *Indicates significantly slower reaction time in comparison to the no-tape control ($P < .016$).

Appendix Seventeen: Rate of Perceived Excursion Scale

Rate of Perceived Excursion Scale used in Study Six

Rating	Description
6	
7	Very, very light
8	
9	Very light
10	
11	Fairly light
12	
13	Somewhat hard
14	
15	Hard
16	
17	Very hard
18	
19	Very, very hard
20	Maximal Exertion

Muscle Reaction Times for Each Taping Condition Pre and Post-Exercise

Muscle	Taping Technique	Neutral Group		Pronated Group		Supinated Group	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peroneus Longus (ms)	No-Tape	52.4(8.9)	59.3(11.4)*	54.8(9.9)	63.5(5.9)*	58.8(7.0)	63.1(13.3)
	Ankle Tape	65.8(8.5)	59.9(15.2)*	68.2(9.1)	61.9(7.6)*	65.8(7.8)	60.0(7.1)
	Arch Tape	56.8(13.3)	58.8(11.1)	57.0(3.7)	58.6(9.6)	63.4(13.8)	64.2(8.7)
Tibialis Anterior (ms)	No-Tape	55.7(13.1)	61.6(16.9)	64.7(15.3)	67.1(14.5)	59.0(6.1)	62.3(21.0)
	Ankle Tape	55.2(9.1)	58.2(16.9)	55.9(20.6)	54.9(20.0)	53.4(16.9)	57.0(13.0)
	Arch Tape	58.3(17.8)	59.2(15.5)	57.9(14.4)	60.4(17.2)	56.7(7.5)	60.2(8.3)
Gluteus Medius (ms)	No-Tape	60.5(13.0)	69.9(8.5)	63.5(6.9)	68.5(14.9)	64.7(11.6)	64.8(18.9)
	Ankle Tape	63.6(19.5)	68.5(15.0)	62.3(14.4)	72.1(9.0)	58.8(10.5)	65.5(10.5)
	Arch Tape	72.3(7.4)	65.9(12.0)	74.8(12.8)	68.3(18.7)	65.3(6.8)	68.4(11.0)

Values are mean (SD). *Indicates significant difference in comparison to the pre-exercise condition ($P < .05$).

Results of Forward Hop Task at 200 ms; Neutral Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	2.8(1.5)	2.1(1.4)	2.1(0.8)	2.0(1.2)	2.1(1.1)	2.4(1.3)
	L	1.0(0.7)	1.2(0.8)	0.9(0.5)	0.9(0.4)	1.0(0.7)	1.2(0.8)
	A	4.6(1.8)	4.6(1.5)	4.1(1.4)	4.0(1.6)	4.0(1.7)	4.8(1.7)
	P	3.8(4.4)	3.5(3.6)	6.1(4.6)	5.8(5.9)	5.4(5.4)	3.4(3.9)
Mean (cm)	M	1.0(0.4)	0.8(0.5)	0.8(0.4)	0.8(0.4)	0.8(0.5)	1.0(0.5)
	L	0.6(0.4)	0.5(0.3)	0.5(0.4)	0.5(0.2)	0.4(0.3)	0.6(0.3)
	A	2.4(0.8)	2.2(0.8)	2.3(0.7)	1.9(0.8)	2.5(1.1)	2.3(0.9)
	P	1.7(1.3)	1.4(1.0)	2.7(1.4)	2.2(1.8)	2.1(2.0)	1.7(1.8)
Total (cm)	M	5.3(2.1)	4.9(1.5)	4.9(1.6)	4.8(1.4)	4.6(1.0)	5.5(1.9)
	L	3.4(1.4)	4.0(1.6)	3.5(1.3)	3.5(1.2)	3.5(1.5)	3.9(1.8)
	A	8.5(2.2)	9.4(2.5)	8.0(3.0)	7.9(2.5)	6.4(2.0)	8.6(2.1)
	P	9.4(5.8)	9.8(5.0)	10.7(6.0)	10.8(6.7)	10.0(6.7)	8.5(4.0)
T. Dist (cm)		21.5(6.7)	23.0(6.5)	22.1(6.8)	22.3(6.1)	20.0(7.3)	21.4(5.3)
V.Avg (cm/s)		107.7(33.4)	114.9(32.6)	110.6(34.2)	111.3(30.7)	99.9(36.3)	107.1(26.3)
95% EA (cm^2)		22.3(8.4)	20.8(7.7)	21.9(12.8)	24.3(15.6)	18.0(9.3)	24.8(13.4)*

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). * Indicates significant difference in postural stability in comparison to pre-exercise condition ($P < .05$).

Results of Forward Hop Task at 200 ms; Pronated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	2.1(1.0)	2.0(1.2)	2.5(1.3)	2.2(0.8)	2.9(1.4)	2.8(1.2)
	L	1.2(0.6)	1.1(0.8)	1.2(0.6)	1.2(0.6)	1.2(0.7)	1.1(0.5)
	A	6.1(1.8)	5.4(1.8)	5.1(2.1)	5.7(2.2)	4.8(1.1)	5.8(1.1)
	P	3.9(4.6)	4.1(4.4)	6.4(5.0)	3.6(3.7)	3.2(2.9)	3.0(3.5)
Mean (cm)	M	0.7(0.3)	0.8(0.5)	0.9(0.4)	0.9(0.4)	1.2(0.8)	1.2(0.5)
	L	0.6(0.2)	0.5(0.3)	0.8(0.3)	0.5(0.3)	0.7(0.3)	0.6(0.2)
	A	3.1(0.6)	2.8(0.9)	2.6(1.0)	3.1(1.2)	2.3(0.5)	3.0(1.0)*
	P	1.9(2.0)	1.6(1.4)	2.6(1.6)	1.8(1.7)	1.4(1.1)	1.4(1.1)
Total (cm)	M	5.1(2.0)	5.7(1.2)	5.3(1.9)	5.0(1.1)	6.0(1.8)	5.9(1.9)
	L	3.8(2.3)	4.3(1.7)	3.8(1.3)	3.9(1.4)	3.6(1.6)	3.7(1.5)
	A	8.9(3.7)	8.8(3.5)	7.5(2.7)	8.7(3.2)	10.2(2.7)	11.1(5.0)
	P	9.6(8.2)	9.3(6.9)	12.6(5.8)	8.6(3.8)	9.9(4.7)	10.3(8.2)
T. Dist (cm)		22.4(11.4)	22.6(8.6)	23.8(5.8)	21.3(3.7)	24.1(7.7)	25.3(13.3)
V.Avg (cm/s)		97.0(31.7)	101.8(22.7)	119.0(29.0)	106.5(18.3)	120.7(38.4)	107.1(23.8)
95% EA (cm^2)		23.9(10.2)	22.3(10.1)	24.8(10.4)	17.8(8.9)	26.3(12.6)	32.5(24.7)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). * Indicates significant difference in postural stability in comparison to pre-exercise condition (P < .05).

Results of Forward Hop Task at 200 ms; Supinated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	3.4(1.7)	3.5(2.4)	2.7(1.8)	2.8(1.7)	2.8(1.4)	3.2(1.8)
	L	1.1(0.7)	1.5(1.1)	1.2(0.8)	1.0(0.6)	1.0(0.6)	0.9(0.7)
	A	5.2(1.7)	5.8(3.0)	4.2(2.1)	4.3(1.9)	4.9(1.9)	3.9(1.5)
	P	3.0(4.2)	4.1(5.9)	5.5(5.3)	4.8(5.7)	5.0(5.0)	4.8(5.1)
Mean (cm)	M	1.3(0.6)	1.4(1.2)	0.9(0.5)	0.9(0.5)	0.9(0.4)	1.1(0.6)
	L	0.6(0.3)	0.7(0.3)	0.5(0.2)	0.4(0.2)	0.4(0.3)	0.3(0.3)
	A	2.9(0.7)	2.8(1.5)	2.4(1.1)	1.9(0.7)	2.4(0.7)	1.7(0.8)
	P	1.2(1.3)	1.6(1.8)	2.3(1.4)	1.8(1.9)	2.2(1.9)	2.2(1.7)
Total (cm)	M	5.8(1.3)	6.2(2.0)	5.6(1.7)	6.1(2.1)	6.5(1.50)	6.4(2.3)
	L	3.6(2.4)	4.7(2.7)	4.1(2.5)	4.4(2.1)	4.7(2.1)	4.8(2.8)
	A	9.1(3.4)	9.4(2.2)	6.8(2.3)	8.4(2.2)	7.5(2.3)	8.1(2.0)
	P	8.5(7.3)	10.5(8.6)	11.0(7.7)	10.7(7.7)	10.9(6.3)	9.8(7.0)
T. Dist (cm)		21.7(10.7)	24.9(11.5)	22.2(10.2)	23.9(9.7)	23.6(7.6)	23.3(9.6)
V.Avg (cm/s)		92.2(20.7)	98.9(18.9)	97.1(30.0)	106.6(30.6)	108.2(25.2)	107.2(39.3)
95% EA (cm^2)		28.4(13.5)	26.2(13.9)	22.3(11.3)	23.7(12.0)	29.2(15.5)	26.6(9.0)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Forward Hop Task at Three Seconds; Neutral Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	3.1(1.3)	2.6(1.1)	2.3(0.7)	2.2(1.1)	2.3(0.9)	2.7(1.1)
	L	1.9(0.5)	1.8(0.6)	1.8(0.6)	1.5(0.3)	1.7(0.6)	1.7(0.6)
	A	5.6(2.0)	5.2(1.3)	5.2(1.3)	4.9(1.2)	4.8(1.3)	5.2(1.5)
	P	3.3(1.0)	3.5(1.7)	7.0(4.1)	6.6(5.4)	6.1(4.9)	3.6(1.2)
Mean (cm)	M	0.7(0.2)	0.8(0.2)	0.8(0.2)	0.7(0.2)	0.7(0.2)	0.8(0.3)
	L	0.7(0.2)	0.7(0.2)	0.7(0.2)	0.6(0.2)	0.7(0.1)	0.7(0.2)
	A	1.7(0.7)	1.6(0.5)	1.5(0.5)	1.4(0.6)	1.6(0.7)	1.6(0.5)
	P	1.4(0.5)	1.3(0.4)	1.4(0.3)	1.2(0.3)	1.1(0.3)	1.3(0.4)
Total (cm)	M	19.4(2.7)	19.1(2.7)	18.6(1.8)	19.3(2.2)	19.3(2.1)	20.8(2.4)
	L	17.8(2.1)	18.1(2.9)	17.5(1.4)	17.9(2.5)	17.7(2.0)	19.0(2.7)
	A	27.5(3.8)	27.4(2.8)	26.0(5.9)	26.0(4.3)	24.2(3.0)	26.9(4.1)
	P	26.2(6.1)	25.5(5.0)	27.4(5.9)	26.4(6.1)	25.5(5.9)	24.9(2.9)
T. Dist (cm)		72.4(8.9)	71.8(8.2)	71.5(9.2)	71.5(7.8)	69.0(5.7)	72.7(6.2)
V.Avg (cm/s)		24.1(3.0)	23.9(2.7)	23.8(3.1)	23.8(2.6)	23.0(1.9)	24.2(2.1)
95% EA (cm^2)		30.8(13.8)	28.6(11.9)	32.2(11.2)	26.0(11.4)	25.8(10.4)	29.0(13.8)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Forward Hop Task at Three Seconds; Pronated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	2.4(0.8)	2.5(0.8)	2.7(1.2)	2.4(0.8)	3.1(1.1)	2.9(1.0)
	L	1.7(0.4)	1.8(0.5)	1.9(0.6)	1.8(0.6)	1.8(0.4)	1.9(0.6)
	A	6.3(1.5)	5.7(1.6)	6.0(1.8)	6.2(1.9)	5.5(0.8)	6.2(1.0)
	P	4.9(4.1)	5.4(3.6)	7.2(4.5)	4.6(3.1)	4.4(2.3)	4.7(3.0)
Mean (cm)	M	0.7(0.1)	0.8(0.2)	0.8(0.2)	0.7(0.2)	0.9(0.5)	0.8(0.2)
	L	0.7(0.1)	0.8(0.2)	0.8(0.3)	0.7(0.2)	0.7(0.2)	0.7(0.2)
	A	1.6(0.6)	1.6(0.4)	1.8(0.7)	2.2(1.2)	1.6(0.5)	1.9(0.6)
	P	1.2(0.4)	1.5(0.4)	1.3(0.4)	1.5(0.6)	1.3(0.3)	1.6(0.6)
Total (cm)	M	18.6(3.1)	19.7(1.8)	19.7(2.9)	19.9(2.2)	20.1(3.5)	19.7(2.8)
	L	17.4(3.6)	18.2(3.1)	18.3(2.6)	18.6(2.9)	17.5(2.9)	17.4(3.2)
	A	26.7(4.1)	27.7(3.7)	28.1(4.7)	28.9(3.6)	28.7(4.7)	31.3(9.1)
	P	24.4(10.4)	24.1(8.2)	30.1(6.1)	25.8(4.7)	25.8(6.6)	26.7(10.4)
T. Dist (cm)		68.6(13.7)	71.4(11.8)	76.8(9.6)	74.3(6.7)	73.4(12.7)	70.7(9.8)
V.Avg (cm/s)		23.1(5.2)	23.8(3.9)	25.6(3.2)	24.8(2.2)	24.5(4.2)	25.4(6.6)
95% EA (cm^2)		27.3(5.5)	33.3(11.4)	33.7(10.4)	35.1(19.0)	28.0(5.9)	37.9(18.4)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Forward Hop Task at Three Seconds; Supinated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	3.5(1.6)	3.6(2.2)	2.9(1.6)	3.0(1.6)	3.0(1.1)	3.2(1.7)
	L	1.7(0.4)	1.9(0.8)	1.6(0.6)	1.7(0.5)	1.7(0.4)	1.6(0.4)
	A	5.7(1.5)	6.4(2.4)	5.0(1.5)	5.3(1.3)	5.5(1.5)	4.4(1.4)
	P	4.0(3.8)	5.7(5.0)	6.1(4.9)	6.0(5.1)	6.0(4.3)	5.6(4.6)
Mean (cm)	M	0.7(0.2)	0.9(0.5)	0.7(0.1)	0.6(0.1)	0.7(0.1)	0.6(0.2)
	L	0.6(0.1)	0.7(0.2)	0.6(0.1)	0.7(0.2)	0.6(0.1)	0.6(0.1)
	A	1.9(0.6)	2.1(0.9)	1.6(0.4)	1.5(0.5)	1.6(0.5)	1.4(0.6)
	P	1.2(0.4)	1.6(0.5)*	1.3(0.2)	1.2(0.4)	1.3(0.2)	1.2(0.2)
Total (cm)	M	19.7(1.9)	20.8(2.2)	19.9(2.9)	20.1(3.1)	21.0(2.4)	21.3(4.5)
	L	17.1(3.4)	18.3(3.7)	18.0(3.7)	18.5(3.2)	19.0(3.2)	19.0(4.9)
	A	27.1(3.9)	28.4(1.4)	25.4(3.4)	25.7(4.4)	25.7(4.4)	25.4(4.2)
	P	23.6(7.5)	25.6(8.4)	26.3(7.1)	25.2(8.4)	26.1(6.3)	24.5(7.4)
T. Dist (cm)		69.5(12.2)	73.8(9.7)	71.0(10.1)	71.1(12.3)	72.5(10.5)	71.4(12.7)
V.Avg (cm/s)		23.2(4.1)	24.6(3.2)	23.7(3.4)	23.7(4.1)	24.2(3.5)	23.8(4.2)
95% EA (cm^2)		27.2(10.8)	39.3(12.5)*	26.0(5.7)	24.7(7.3)	26.2(4.7)	26.0(6.6)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). * Indicates significant difference in postural stability in comparison to pre-exercise condition ($P < .05$).

Results of Diagonal Hop Task at 200 ms; Neutral Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	1.5(1.3)	1.3(0.8)	1.1(0.7)	1.0(0.5)	1.3(0.6)	1.2(1.0)
	L	2.4(1.5)	2.3(1.5)	2.5(1.1)	2.6(1.0)	2.3(1.2)	2.3(1.2)
	A	6.8(1.6)	6.7(2.6)	6.8(2.4)	7.2(2.3)	7.3(2.7)	6.7(2.2)
	P	2.2(1.8)	2.6(3.2)	4.2(4.1)	3.6(3.9)	2.8(3.6)	2.5(2.9)
Mean (cm)	M	0.7(0.5)	0.7(0.4)	0.9(0.5)	0.8(0.6)	0.7(0.4)	0.6(0.5)
	L	0.7(0.4)	0.8(0.4)	1.0(0.3)	1.0(0.4)	0.9(0.4)	0.9(0.3)
	A	3.7(1.0)	3.6(0.9)	3.4(1.1)	3.6(1.4)	4.0(1.0)	3.8(0.8)
	P	1.1(0.7)	1.3(1.0)	2.0(1.6)	1.6(1.5)	1.7(1.1)	1.2(1.3)
Total (cm)	M	3.4(1.6)	3.8(1.8)	2.8(1.3)	3.4(1.9)	3.0(1.1)	3.2(1.5)
	L	4.8(2.1)	5.5(1.6)	5.1(1.6)	5.2(2.0)	4.9(1.4)	4.8(1.4)
	A	10.0(2.8)	11.2(3.0)	10.0(2.8)	11.1(2.6)	9.9(2.3)	9.4(2.9)
	P	5.1(3.7)	6.5(3.9)	7.2(5.4)	6.8(5.5)	5.6(4.5)	5.2(3.2)
T. Dist (cm)		18.8(6.9)	25.0(12.5)	20.2(5.8)	21.7(7.5)	19.0(5.4)	18.4(5.2)
V.Avg (cm/s)		88.2(26.1)	105.7(26.6)*	96.7(24.3)	98.4(25.6)	95.0(26.9)	86.8(16.9)
95% EA (cm^2)		29.4(13.6)	30.0(13.2)	31.1(15.1)	27.4(11.8)	31.9(14.4)	29.7(13.3)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). * Indicates significant difference in postural stability in comparison to pre-exercise condition ($P < .05$).

Results of Diagonal Hop Task at 200 ms; Pronated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	1.1(0.6)	1.1(0.7)	1.8(1.6)	1.6(0.9)	1.6(0.8)	2.2(1.5)
	L	2.8(1.2)	2.9(1.5)	1.9(1.0)	2.2(1.3)	2.5(1.6)	2.7(1.1)
	A	7.7(2.3)	6.4(2.8)	7.4(1.9)	8.4(2.6)	6.6(1.6)	7.0(1.2)
	P	2.3(2.3)	3.8(3.2)	2.6(2.0)	2.8(2.2)	3.1(3.4)	2.9(2.6)
Mean (cm)	M	0.6(0.4)	0.7(0.5)	0.7(0.4)	0.7(0.5)	0.7(0.4)	1.3(1.2)
	L	1.0(0.4)	1.0(0.3)	0.8(0.3)	0.7(0.4)	0.9(0.5)	1.1(0.4)
	A	3.8(1.0)	3.6(1.8)	3.9(1.1)	4.2(1.1)	3.2(0.9)	3.6(0.6)
	P	1.3(0.8)	1.3(0.9)	1.3(0.7)	1.4(0.8)	1.4(1.1)	1.4(1.0)
Total (cm)	M	4.2(1.6)	4.9(2.3)	4.3(1.8)	5.2(2.7)	4.8(1.5)	5.9(4.1)
	L	5.4(2.0)	6.0(2.3)	4.3(1.4)	6.6(3.2)	6.0(2.1)	7.0(3.4)
	A	11.3(4.1)	10.0(3.5)	10.5(2.3)	13.2(3.8)	11.4(2.9)	12.2(4.1)
	P	6.4(5.0)	7.8(4.8)	5.8(2.3)	8.2(4.1)	9.1(8.6)	10.0(8.5)
T. Dist (cm)		21.9(8.6)	23.0(7.6)	20.1(3.1)	27.2(8.8)	25.3(11.3)	28.3(14.9)
V.Avg (cm/s)		106.3(35.8)	106.8(28.0)	100.7(15.3)	122.2(30.9)	111.0(27.1)	110.0(27.8)
95% EA (cm^2)		35.3(11.3)	32.2(8.0)	28.8(13.9)	32.0(14.9)	35.3(19.6)	34.9(18.0)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Diagonal Hop Task at 200 ms; Supinated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	1.7(1.1)	1.6(0.9)	1.4(1.0)	1.3(0.6)	2.0(1.0)	1.8(0.8)
	L	2.2(1.7)	2.7(2.0)	2.8(2.0)	3.2(2.4)	2.8(1.7)	2.1(1.6)
	A	7.0(3.5)	6.9(3.0)	6.1(3.2)	5.9(3.4)	6.0(2.6)	6.1(2.5)
	P	3.2(4.4)	3.9(5.9)	3.9(5.2)	4.2(4.8)	4.3(4.4)	3.9(3.7)
Mean (cm)	M	0.9(0.5)	0.8(0.4)	0.7(0.3)	0.6(0.3)	1.0(0.5)	0.9(0.4)
	L	0.8(0.5)	0.9(0.4)	1.0(0.5)	1.0(0.6)	1.3(0.6)	0.9(0.7)
	A	3.5(1.5)	3.2(1.6)	2.8(1.3)	2.6(1.2)	3.1(0.9)	2.9(0.9)
	P	1.8(1.5)	1.4(1.7)	1.4(1.8)	1.7(1.3)	1.9(1.9)	1.8(1.5)
Total (cm)	M	4.9(1.4)	5.4(2.1)	4.6(1.8)	5.3(2.0)	4.7(1.7)	4.6(1.1)
	L	5.6(2.8)	7.6(4.0)*	6.4(3.6)	7.3(3.6)*	6.7(3.6)	6.4(2.8)
	A	11.4(3.8)	12.2(2.8)	9.1(2.7)	10.0(3.3)	9.0(2.8)	8.9(3.0)
	P	7.7(6.3)	10.6(10.4)	9.2(7.7)	9.3(6.8)	7.7(7.4)	8.1(5.7)
T. Dist (cm)		23.4(9.0)	28.4(14.2)	23.0(9.2)	25.0(8.4)	22.1(9.0)	21.9(5.9)
V.Avg (cm/s)		106.3(30.9)	142.1(71.1)	104.2(32.1)	124.9(42.1)	102.5(38.1)	109.6(29.3)
95% EA (cm^2)		36.7(28.2)	43.2(33.8)	39.3(31.0)	43.3(31.5)	31.3(15.2)	36.7(34.3)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). * Indicates significant difference in postural stability in comparison to pre-exercise condition (P < .05).

Results of Diagonal Hop Task at Three Seconds; Neutral Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	2.2(0.9)	2.2(0.6)	2.0(0.5)	2.1(0.4)	2.1(0.3)	1.9(0.6)
	L	2.9(1.2)	2.6(1.3)	2.6(1.0)	2.8(0.9)	2.7(0.9)	2.6(1.0)
	A	7.1(1.6)	7.1(2.2)	7.1(2.1)	7.3(2.1)	7.4(2.7)	6.7(2.2)
	P	3.3(1.5)	3.7(2.8)	5.4(3.8)	4.5(3.5)	4.1(3.0)	3.4(2.6)
Mean (cm)	M	0.8(0.2)	0.9(0.2)	0.9(0.3)	0.9(0.2)	0.9(0.2)	0.7(0.1)
	L	0.9(0.2)	0.7(0.1)	0.8(0.3)	0.8(0.2)	0.8(0.2)	0.7(0.1)
	A	1.8(0.7)	1.7(0.7)	1.5(0.6)	1.5(0.4)	1.5(0.5)	1.4(0.4)
	P	1.1(0.4)	1.2(0.4)	1.3(0.4)	1.2(0.5)	1.1(0.2)	1.1(0.3)
Total (cm)	M	18.2(1.5)	18.2(2.4)	17.1(1.7)	18.2(3.6)	17.9(2.3)	17.9(2.6)
	L	18.9(2.8)	19.5(2.8)	18.9(1.9)	19.9(3.4)	19.4(1.9)	19.1(1.7)
	A	28.5(3.7)	28.5(3.0)	27.7(4.2)	28.8(4.6)	28.2(3.5)	26.7(3.6)
	P	22.0(4.7)	22.1(4.0)	24.4(6.0)	22.8(5.3)	22.6(3.2)	20.8(3.2)
T. Dist (cm)		69.8(9.0)	70.3(7.2)	69.8(7.0)	71.5(10.6)	70.0(4.5)	67.3(5.1)
V.Avg (cm/s)		23.3(3.0)	23.4(2.4)	23.3(2.3)	23.8(3.5)	23.3(1.5)	22.4(1.7)
95% EA (cm^2)		32.6(13.1)	32.8(14.8)	33.7(20.0)	32.0(11.3)	33.0(10.6)	25.9(7.9)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Diagonal Hop Task at Three Seconds; Pronated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	1.7(0.4)	2.0(0.5)	2.3(1.4)	2.3(0.7)	2.2(0.7)	2.9(1.1)
	L	2.9(1.1)	3.0(1.4)	2.3(0.7)	2.6(1.0)	2.9(1.3)	2.9(1.2)
	A	7.8(2.0)	6.7(2.3)	7.5(1.7)	8.5(2.5)	6.7(1.2)	7.2(1.1)
	P	3.4(1.8)	4.4(2.8)	3.6(1.7)	3.5(2.1)	4.4(3.0)	4.0(2.3)
Mean (cm)	M	0.7(0.1)	0.8(0.1)	0.7(0.2)	0.8(0.2)	0.8(0.2)	1.3(0.6)*
	L	0.8(0.1)	0.7(0.2)	0.8(0.2)	0.8(0.2)	0.9(0.2)	1.1(0.7)
	A	1.6(0.5)	1.8(0.7)	1.6(0.7)	1.8(0.8)	1.5(0.5)	1.9(0.6)
	P	1.3(0.2)	1.2(0.3)	1.3(0.3)	1.2(0.4)	1.3(0.5)	1.6(0.9)*
Total (cm)	M	18.6(3.5)	19.2(3.4)	20.1(3.9)	21.2(5.2)	19.7(3.4)	22.0(6.5)
	L	19.9(3.8)	20.4(4.2)	19.8(3.5)	22.1(5.3)	20.6(3.0)	21.9(5.3)
	A	28.8(4.6)	28.3(2.9)	30.4(5.6)	33.1(3.9)	30.5(4.6)	33.3(8.8)
	P	21.7(6.8)	22.7(6.5)	24.2(3.8)	25.4(4.3)	26.4(9.8)	27.5(12.4)
T. Dist (cm)		70.5(12.7)	72.1(9.9)	74.7(9.5)	81.0(11.0)	77.2(14.2)	83.4(24.5)
V.Avg (cm/s)		23.5(4.2)	24.0(3.3)	24.9(3.2)	27.0(3.7)	25.7(4.7)	27.8(8.2)
95% EA (cm^2)		28.2(8.4)	28.8(10.1)	32.0(12.8)	35.1(15.4)	34.3(15.7)	38.9(11.2)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2). * Indicates significant difference in postural stability in comparison to pre-exercise condition ($P < .05$).

Results of Diagonal Hop Task at Three Seconds; Supinated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	2.3(0.7)	2.2(0.6)	2.1(0.7)	1.9(0.3)	2.4(0.8)	2.3(0.5)
	L	2.7(1.3)	3.0(1.9)	3.2(1.8)	3.3(2.3)	3.0(1.6)	2.6(1.2)
	A	7.3(3.1)	7.2(2.6)	6.7(2.6)	6.3(2.9)	6.3(2.3)	6.8(2.0)
	P	4.3(4.0)	5.4(5.1)	5.3(4.4)	5.4(4.2)	5.2(3.8)	4.6(3.4)
Mean (cm)	M	0.9(0.3)	0.9(0.3)	0.8(0.2)	0.7(0.1)	0.8(0.2)	0.8(0.1)
	L	0.9(0.3)	0.8(0.2)	0.9(0.3)	0.6(0.1)	0.8(0.2)	0.7(0.1)
	A	1.6(0.3)	1.6(0.5)	1.6(0.6)	1.4(0.6)	1.7(0.3)	1.8(0.6)
	P	1.2(0.4)	1.3(0.5)	1.2(0.3)	1.2(0.3)	1.2(0.2)	1.2(0.3)
Total (cm)	M	19.0(1.7)	20.2(2.5)	19.9(5.6)	20.0(5.1)	20.3(4.0)	21.1(3.9)
	L	19.6(3.0)	21.6(3.9)	21.4(6.1)	22.2(6.2)	21.4(4.5)	21.5(4.9)
	A	28.9(3.7)	29.4(3.1)	27.5(4.6)	27.4(4.4)	26.0(4.5)	26.4(4.2)
	P	23.6(7.0)	25.0(10.7)	25.6(6.6)	25.1(7.5)	23.0(6.9)	22.4(5.5)
T. Dist (cm)		71.8(8.9)	76.2(13.6)	74.3(12.7)	74.6(14.5)	71.4(11.5)	72.0(9.3)
V.Avg (cm/s)		23.9(3.0)	25.4(4.5)	24.8(4.2)	24.9(4.8)	23.8(3.8)	24.0(3.1)
95% EA (cm^2)		34.4(14.4)	33.3(16.0)	34.9(16.7)	24.9(8.0)	32.8(7.1)	30.7(10.1)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Lateral Hop Task at 200 ms; Neutral Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	1.4(1.1)	1.1(0.9)	1.1(0.6)	1.4(1.4)	1.4(1.1)	1.6(1.2)
	L	3.3(2.4)	3.2(1.4)	3.6(2.2)	3.2(1.8)	2.7(0.8)	2.6(1.3)
	A	9.0(2.3)	8.9(2.5)	9.7(1.8)	9.7(1.8)	9.3(2.7)	9.7(2.7)
	P	2.1(1.9)	2.0(1.8)	2.6(1.9)	2.0(1.4)	1.8(1.0)	2.1(1.4)
Mean (cm)	M	0.8(0.4)	0.8(0.7)	0.8(0.4)	0.8(0.7)	0.9(0.8)	0.8(0.6)
	L	1.0(0.6)	1.0(0.4)	1.3(0.6)	1.2(0.6)	1.0(0.2)	1.0(0.5)
	A	4.6(0.9)	4.6(0.8)	4.9(0.8)	4.6(0.8)	4.5(1.0)	4.6(0.7)
	P	1.0(0.9)	1.0(0.7)	1.1(1.0)	1.0(0.8)	0.8(0.2)	0.9(0.6)
Total (cm)	M	3.6(1.7)	3.5(1.2)	3.8(2.4)	4.2(2.4)	3.5(1.7)	4.2(1.9)
	L	6.2(3.2)	6.2(2.2)	6.3(3.1)	6.5(2.8)	5.6(1.4)	6.1(1.7)
	A	12.1(3.8)	12.9(3.1)	13.8(4.0)	13.5(4.1)	13.0(3.8)	13.2(5.0)
	P	4.9(3.1)	5.9(3.3)	5.8(3.7)	6.1(3.6)	5.7(3.3)	6.2(3.5)
T. Dist (cm)		21.5(7.7)	22.9(7.2)	24.0(9.4)	24.5(8.5)	22.4(7.2)	23.9(9.0)
V.Avg (cm/s)		107.3(38.4)	114.3(36.1)	119.9(47.2)	122.5(42.3)	111.8(36.0)	119.6(44.8)
95% EA (cm^2)		48.4(29.3)	45.9(16.2)	55.4(35.7)	49.7(26.3)	47.1(19.9)	46.5(22.1)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Lateral Hop Task at 200 ms; Pronated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	1.2(0.9)	1.3(0.7)	1.4(0.7)	1.5(1.1)	2.1(1.4)	1.9(1.2)
	L	2.4(1.0)	2.7(1.5)	2.7(1.4)	2.5(1.0)	2.5(2.0)	2.7(2.0)
	A	9.7(1.0)	9.0(2.4)	9.6(2.4)	10.6(1.8)	8.9(2.8)	9.9(2.1)
	P	1.8(0.8)	1.6(1.1)	1.8(1.1)	1.7(0.8)	2.3(2.3)	2.1(1.7)
Mean (cm)	M	0.7(0.5)	0.8(0.5)	0.9(0.5)	0.8(0.5)	1.2(1.0)	1.1(0.8)
	L	1.0(0.4)	0.9(0.4)	1.0(0.3)	1.1(0.4)	0.9(0.5)	0.9(0.5)
	A	5.0(1.1)	4.6(1.3)	4.6(1.0)	5.1(0.9)	4.3(1.2)	4.9(0.9)
	P	0.8(0.2)	0.9(0.9)	0.8(0.5)	0.8(0.5)	1.0(0.9)	1.0(0.8)
Total (cm)	M	3.6(1.6)	5.5(2.7)	4.1(1.6)	5.2(3.3)	4.9(2.4)	4.9(2.9)
	L	5.5(1.8)	6.6(2.8)	5.8(0.8)	7.1(4.1)	6.1(2.6)	6.3(3.2)
	A	11.4(1.8)	12.8(2.7)	13.0(3.0)	14.3(2.8)	12.8(4.8)	13.3(5.0)
	P	4.2(2.4)	6.2(3.0)	4.9(2.9)	7.5(4.7)	6.8(6.9)	7.6(6.8)
T. Dist (cm)		19.9(5.5)	24.7(7.9)	22.2(4.6)	26.9(10.3)	24.6(12.3)	26.0(13.3)
V.Avg (cm/s)		99.4(27.3)	123.4(39.5)	110.9(23.0)	134.7(51.4)	122.8(61.7)	130.1(66.7)
95% EA (cm^2)		39.5(13.0)	40.2(14.3)	44.9(24.6)	53.7(22.6)	45.6(21.1)	43.9(22.1)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Lateral Hop Task at 200 ms; Supinated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	1.7(0.9)	1.9(1.2)	1.0(0.6)	1.5(0.8)	1.7(1.0)	2.0(1.3)
	L	1.9(0.9)	2.5(2.7)	4.2(3.0)	3.6(2.4)	3.8(2.7)	3.2(2.2)
	A	9.1(2.4)	9.0(3.1)	8.7(3.8)	8.6(3.1)	9.6(3.2)	8.5(3.6)
	P	1.2(0.9)	1.5(0.9)	3.5(3.8)	2.1(1.3)	2.8(2.3)	3.0(2.9)
Mean (cm)	M	1.0(0.5)	1.1(0.7)	0.7(0.5)	0.9(0.5)	0.7(0.5)	1.0(0.6)
	L	0.7(0.4)	0.8(0.4)	1.3(0.6)	1.1(0.4)	1.5(0.8)	1.1(0.4)
	A	4.5(1.4)	4.3(1.6)	4.1(1.5)	3.9(1.3)	4.9(0.8)	3.9(1.3)
	P	0.6(0.4)	0.8(0.3)	1.6(0.7)	0.9(0.5)	1.7(1.3)	1.7(1.0)
Total (cm)	M	4.2(1.6)	4.9(0.9)	3.5(2.0)	4.8(1.7)	4.1(2.9)	4.9(2.5)
	L	5.5(1.7)	7.2(3.5)	7.3(3.6)	7.4(3.5)	7.2(3.2)	7.8(3.0)
	A	12.4(3.2)	12.4(4.1)	11.6(5.8)	12.3(3.7)	10.8(3.7)	10.7(4.5)
	P	5.5(2.4)	7.0(3.6)	6.6(4.4)	6.5(3.3)	5.4(3.7)	6.1(4.8)
T. Dist (cm)		22.0(5.8)	24.9(8.2)	23.0(8.4)	24.5(7.4)	21.5(7.9)	22.9(7.9)
V.Avg (cm/s)		110.1(29.0)	124.7(40.9)	114.9(42.1)	122.5(36.9)	107.4(39.6)	114.7(39.6)
95% EA (cm^2)		33.9(10.3)	34.9(21.8)	51.6(29.3)	41.5(22.2)	41.3(17.8)	47.0(24.3)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Lateral Hop Task at Three Seconds; Neutral Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	2.3(1.0)	2.2(0.7)	1.9(0.4)	2.1(1.2)	2.0(0.8)	2.2(1.0)
	L	3.7(2.0)	3.4(1.2)	3.7(2.0)	3.4(1.5)	2.8(0.7)	2.8(1.2)
	A	9.0(2.3)	8.9(2.4)	9.7(1.8)	9.7(1.8)	9.3(2.7)	9.7(2.7)
	P	3.6(1.7)	3.5(1.5)	3.7(1.6)	3.3(1.3)	2.9(0.8)	3.3(1.3)
Mean (cm)	M	0.9(0.5)	1.0(0.3)	0.9(0.2)	0.9(0.2)	0.7(0.1)	0.8(0.2)
	L	0.9(0.4)	0.9(0.3)	0.9(0.2)	0.8(0.2)	0.8(0.2)	0.8(0.2)
	A	1.8(0.6)	1.6(0.5)	1.5(0.4)	1.6(0.3)	1.4(0.4)	1.6(0.4)
	P	1.3(0.4)	1.4(0.5)	1.3(0.2)	1.2(0.4)	1.0(0.2)	1.2(0.4)
Total (cm)	M	18.8(2.4)	18.4(2.2)	18.2(2.3)	19.0(3.5)	18.6(2.6)	19.3(3.1)
	L	21.2(3.7)	20.3(2.2)	20.6(2.7)	20.8(3.0)	20.2(1.5)	21.0(2.1)
	A	30.5(6.9)	30.2(4.2)	30.9(3.3)	30.7(3.7)	28.9(3.3)	30.7(4.8)
	P	21.4(4.9)	21.9(4.0)	21.6(4.2)	21.4(3.6)	20.6(2.8)	22.0(3.7)
T. Dist (cm)		70.4(6.1)	71.9(8.3)	72.5(8.7)	73.1(8.5)	70.1(5.0)	73.8(8.8)
V.Avg (cm/s)		23.5(2.0)	24.0(2.8)	24.2(2.9)	24.4(2.8)	23.4(1.7)	24.6(2.9)
95% EA (cm^2)		34.2(8.2)	41.6(25.8)	35.1(7.0)	36.8(15.5)	28.3(8.9)	33.6(9.6)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Lateral Hop Task at Three Seconds; Pronated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	2.0(0.5)	1.9(0.5)	2.0(0.5)	2.4(1.0)	2.6(1.1)	2.7(1.0)
	L	2.7(0.8)	3.0(1.3)	2.8(1.3)	2.9(0.8)	3.1(1.6)	3.1(1.8)
	A	9.7(1.0)	9.0(2.4)	9.6(2.4)	10.6(1.8)	9.1(2.6)	9.9(2.1)
	P	3.1(0.6)	2.9(0.6)	3.4(1.0)	3.7(1.4)	3.9(1.8)	3.3(1.5)
Mean (cm)	M	0.8(0.1)	0.8(0.2)	0.8(0.3)	0.8(0.2)	1.0(0.3)	1.0(0.4)
	L	0.8(0.2)	0.8(0.2)	0.8(0.2)	0.9(0.3)	0.9(0.2)	0.9(0.3)
	A	1.9(0.5)	1.7(0.4)	1.7(0.3)	1.6(0.5)	1.6(0.3)	1.8(0.4)
	P	1.3(0.2)	1.1(0.3)	1.3(0.4)	1.3(0.4)	1.3(0.3)	1.2(0.2)
Total (cm)	M	18.4(3.6)	20.3(4.5)	18.9(2.6)	20.5(5.9)	20.5(4.8)	20.3(4.8)
	L	19.6(3.6)	21.2(5.1)	20.7(2.0)	22.0(5.5)	21.1(4.0)	21.0(4.0)
	A	29.1(3.0)	30.7(3.3)	30.8(3.8)	33.0(4.4)	31.5(6.6)	32.8(7.4)
	P	19.0(4.4)	21.2(4.0)	21.0(3.0)	23.7(5.0)	23.3(8.1)	23.8(8.7)
T. Dist (cm)		68.3(10.7)	73.8(10.8)	72.5(7.1)	78.4(14.6)	76.2(16.5)	97.3(60.1)
V.Avg (cm/s)		22.8(3.6)	24.6(3.6)	24.2(2.4)	26.1(4.9)	25.4(5.5)	32.4(20.0)
95% EA (cm^2)		38.7(12.1)	35.0(11.9)	35.6(13.7)	42.6(19.2)	43.6(18.0)	57.1(42.9)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).

Results of Lateral Hop Task at Three Seconds; Supinated Group

		No-Tape		Ankle Tape		Arch Tape	
		Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak (cm)	M	2.3(0.7)	2.5(0.7)	1.9(0.4)	2.1(0.5)	2.4(0.9)	2.6(1.0)
	L	2.4(0.6)	3.0(2.4)	4.2(2.9)	3.8(2.1)	3.9(2.7)	3.6(2.0)
	A	9.2(2.3)	9.0(3.1)	8.8(3.5)	8.6(3.0)	9.6(3.2)	8.6(3.2)
	P	2.9(0.9)	3.1(1.0)	5.0(3.4)	4.1(0.9)	4.4(1.9)	3.9(2.5)
Mean (cm)	M	0.8(0.3)	0.8(0.2)	0.9(0.2)	0.8(0.2)	0.9(0.2)	1.0(0.2)
	L	0.8(0.2)	0.8(0.1)	0.9(0.1)	0.9(0.2)	0.9(0.3)	0.8(0.1)
	A	1.6(0.4)	1.7(0.7)	1.7(0.4)	1.4(0.4)	1.9(0.4)	1.6(0.4)
	P	1.2(0.3)	1.3(0.6)	1.3(0.3)	1.4(0.3)	1.8(0.5)	1.1(0.3)
Total (cm)	M	19.4(3.8)	20.5(3.5)	19.2(6.1)	20.3(6.1)	20.9(5.8)	21.4(5.4)
	L	20.1(3.0)	21.9(5.1)	22.8(6.7)	22.7(6.8)	23.9(5.9)	23.3(5.0)
	A	29.9(3.5)	30.4(4.8)	30.2(6.9)	30.6(4.3)	30.0(5.7)	28.4(6.0)
	P	20.7(4.2)	21.7(3.7)	23.1(3.8)	21.9(3.7)	22.1(4.7)	22.0(5.6)
T. Dist (cm)		71.7(9.9)	74.7(10.3)	75.6(14.6)	75.6(13.7)	76.0(13.7)	74.5(12.4)
V.Avg (cm/s)		23.9(3.3)	24.9(3.4)	25.2(4.9)	25.2(4.6)	25.3(4.6)	24.8(4.1)
95% EA (cm^2)		35.3(13.8)	35.3(19.6)	40.0(14.1)	35.7(15.5)	49.5(20.5)	36.0(11.1)

Values are mean (SD). A: anterior; M: medial; P: posterior; L: lateral; T. Dist: total distance (cm); V.Avg: average velocity (cm/s); 95% EA: 95% Ellipse Area (cm^2).