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REFERENCE

FOREFOOT-REARFOOT KINEMATICS AS RISK FACTORS FOR TIBIAL STRESS INJURIES

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A thesis submitted in partial fulfilment of the requirements of Sheffield Hallam University for the degree of Doctor of Philosophy

July 2011

Abstract

Overuse injuries represent a significant problem for runners with tibial stress injuries common. Identifying risk factors is a crucial step in the development of effective preventative measures and treatments for specific injuries. The foot has been suggested to be an intrinsic risk factor for tibial stress injury, however, literature is contradictory. Few studies have investigated dynamic foot function as a potential risk factor. A recent area of biomechanics research has focussed on the development of multisegment foot models for use in research and clinical settings. However to date, few studies have used such models to try and answer questions relating to overuse injuries. The overall purpose of this thesis was to answer the question, are forefoot-rearfoot kinematics risk factors for tibial stress injuries in runners? Chapter II conducted a systematic review of the literature to determine any relationship between tibial stress injuries and foot type. Results proved conflicting with limited evidence implicating any one foot type as a potential risk factor. Importantly, dynamic measures of foot function were suggested to be more useful in predicting injury risk. Chapter III examined methodological considerations with the measurement of forefoot-rearfoot kinematics. A new multisegment foot model (forefoot, rearfoot and shank) for use with gait sandals was developed and presented. Data was also presented to support the use of gait sandals as an effective means of measuring the kinematic motions seen when wearing running shoes. The Chapter also addressed the reliability of selected kinematic measures and tibial shock. Selected kinematic variables focused on peak joint angles, excursions and velocities which define the motions of the foot during loading. Chapter IV presents an application of the developed model. The study investigated foot function in relation to foot type and sought to compare forefoot-rearfoot kinematics in high and low-arched populations. Differences between high and low-arched feet were observed for some dynamic forefoot-rearfoot variables. Specifically, greater forefoot abduction excursion, forefoot dorsiflexion excursion and forefoot abduction velocity were found in high-arched compared to low-arched individuals. Further, differences between groups were also noted in the coupling between the forefoot and rearfoot, but these differences did not appear to be transferred proximally to the shank. Chapter V studied dynamic foot function in relation to tibial stress injury risk. The study sought to compare forefoot-rearfoot kinematics in those with a history of tibial stress injury and a matched control group. Key differences were found in kinematic variables between injury and control groups. Specifically, peak rearfoot eversion, peak forefoot dorsiflexion velocity and peak forefoot abduction velocity were found to be greater in those with a history of tibial stress injury compared to a control group. Differences were also noted between the coupling of the forefoot, rearfoot and shank, these may characterise the unique loading pattern seen in those with tibial stress injuries. Although only initial findings, this thesis has enhanced understanding of important dynamic risk factors for tibial stress injuries associated with the foot. Furthermore, findings serve to highlight the importance of forefoot motions as risk factors. It is hoped the findings of this thesis will be a useful basis for future research and represent an important step in the development of effective preventative measures and treatments for these injuries.

Acknowledgments

I would like to express my thanks to the following people for their assistance throughout my PhD, I am indebted. To my supervisors, Professor Ian Maynard and Dr Jon Wheat who's guidance throughout the process has been invaluable. Special thanks go to Jon, who during my time at Sheffield Hallam has been instrumental in my personal development as both a researcher and academic. I would like to extend my gratitude to Dr Clare Milner, not least for her numerous light hearted words of encouragement during the peer review process. Her expert knowledge of this research area has been invaluable in improving the quality of this thesis.

I would also like to express my sincere thanks to 'those that lunch' for their continued banter throughout the completion of this thesis, in particular, to Professor Edward Winter for his numerous pearls of wisdom in TVC.

Last, but by no means least, I would like to thank my family and friends for their support over the course of the thesis. Especially my mum and dad, Mike and Linda, I shall forever be indebted for your support throughout my education. This one's for you.

Preface

The findings presented in this thesis have been peer reviewed as follows:

Barnes, A., Wheat, J., and Milner, C.E. (2007). The use of gait sandals in the measurement of rearfoot motion during walking and running. *In the Scientific Proceedings of the Fourth Salford Conference on Biomechanics of the Lower Limb in Health, Disease and Rehabilitation.* Salford, UK.

Barnes, A., Wheat, J., and Milner, C.E. (2008). Association between foot type and tibial stress injuries: a systematic review. *British Journal of Sports Medicine,* **42**, 93-98.

Barnes, A., Wheat, J., and Milner, C.E. (2008). The relationship between static arch height and arch stiffness. *In the Scientific Proceedings of the North American Congress of Biomechanics.* Ann Arbor, USA.

Barnes, A., Wheat, J., and Milner, C.E. (2008). Dynamic foot mobility in high and low-arched individuals. *In the Scientific Proceedings of the North American Congress of Biomechanics.* Ann Arbor, USA.

Barnes, A., Wheat, J., and Milner, C.E. (2010). Use of gait sandals for measuring rearfoot and shank motion during running. *Gait and Posture*, **32**, 133-135.

Barnes, A., Wheat, J., and Milner, C.E. (2011). Fore- and rearfoot kinematics in high- and low-arched individuals during running. *Foot and Ankle International,* ***32,*** 710-17.

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CHAPTER I

1.1 Introduction

Biomechanics may be defined as "the study of forces and effects of these forces on living things" (Grimshaw *et al.*, 2007, P11). Bartlett (2007, P1) defined sports biomechanics as "the study and analysis of human movement patterns in sport". One area of particular focus within sports biomechanics is running mechanics, an area of research which has generated much interest since the growth of running as a recreational activity in the 1970s. This explosion in running caused a comparable growth in the research and assessment of running biomechanics (Novacheck, 1998). Biomechanists have investigated questions related to both improving running performance as well as reducing injury risk. However, the main focus in this area has been identifying biomechanical risk factors that cause overuse injuries. Identifying risk factors is a crucial step in the development of effective preventative measures and treatments for specific injuries.

One common overuse injury in runners is stress fracture to bone (Arendt *et al.,* 2003). Running is a highly unconstrained activity, with the repetitive loading of the lower extremities resulting in ground impact forces. During running, these impact forces have been reported to reach between two and four times body weight (Nigg *et al.*, 1995) within the first 35ms of stance (Cavanagh and Lafortune, 1980). The repetitive loading during gait has been proposed to have a positive effect on the body, promoting an osteogenic response which helps to develop and maintain bone tissue, as well as helping to strengthen soft tissue structures (Nigg and Wakeling, 2001). Despite this, excessive impact forces have been implicated as a possible cause of several overuse injuries in runners (Hreljac *et al.,* 2000). Stress injuries occur when the bone is unable to successfully adapt quickly enough to the repetitive load imparted upon it. In runners, stress fractures are extremely prevalent, accounting for 15-20% of all musculoskeletal injuries in this population (Bennell *et al.*, 1999). Although they are found at a range of locations, the tibia is the most common site of injury, with tibial stress fractures accounting for between 35% and 49% of all stress fractures in runners (McBryde, 1985; Matheson *et al.,* 1987; Arendt *et al.*, 2003).

Intrinsic risk factors have been associated with overuse running injuries in up to 40% of cases (Lyshol and Wiklander, 1987). Foot biomechanics have become important in the study of running related injuries, with foot structure a factor which has received much attention in the literature (Willems *et al.*, 2005). Structural foot characteristics have been considered as potential risk factors associated with lower extremity injury (Sullivan *et al.*, 1984; Giladi *et al.*, 1985; Cowan *et al.*, 1993). However, much of this research is based on the use of static measures of the foot which fail to consider dynamic foot function. Traditional views of foot function have suggested that high-arched feet tend to be more rigid whilst low-arched feet are more flexible (Matheson *et al.*, 1987; Simkin *et al.*, 1989). Indeed, whether this is actually the case remains unclear. At present, little is known about how foot function relates to injury (Williams *et al.*, 2004). Given that tibial stress injuries are related to the repetitive stressing of the bone, how the foot behaves during the loading response is crucial to understanding foot function and its potential relationship to this type of injury.

The foot is an extremely complex structure, comprising numerous bones and articulations, which, until recently had been modelled during routine dynamic biomechanical analysis as a single rigid segment (Davis *et al.,* 1991). The study of the rearfoot has been the focus of analysis conducted on the foot (McClay and Manal, 1997). The rearfoot has also been investigated as an injury risk factor (Hreljac *et al.*, 2000). Recent studies have investigated the relationship between rearfoot mechanics and tibial stress injuries (Hetsroni *et al.*, 2008; Pohl *et al.*, 2008; Milner *et al.*, 2010). Greater peak rearfoot eversion has been identified as a risk factor for tibial stress fractures in female runners (Pohl *et al.*, 2008; Milner *et al.*, 2010). Greater rearfoot eversion may result in an increased torsional load being placed on the tibia during stance (Brukner and Bennell, 2005). However, a further study found no difference in peak rearfoot eversion or eversion excursion between a stress fracture and control group (Hetsroni *et al.*, 2008). These studies are the only ones to date which address dynamic risk factors for tibial stress injuries associated with the foot. Further research is needed to establish a clear link between rearfoot kinematics during gait and the risk of tibial stress injuries.

In addition to the rearfoot, motion of the forefoot has received no attention in the literature as a potential risk factor for tibial stress injuries. Recent *in vivo* studies of foot and ankle biomechanics have served to highlight the relative complexity of the foot's movements during gait (Ardent *et al.*, 2007; Lundgren *et al.*, 2008). These studies have sought to analyse individual articulations between the bones of the foot and point to significant motion at these joints. Given the close association between the bones of the rearfoot and forefoot (Ardent *et al.*, 2007;

Lundgren *et al.*, 2008), it is suggested that forefoot motion might also be significant in determining the foot's response to loading. Forefoot motion has been found to be coupled with motion of the rearfoot through the joints of the midfoot (Cornwall and McPoil, 2002; Pohl *et al.*, 2006). Given the reported association between rearfoot eversion and tibial stress fracture risk (Pohl *et al.*, 2008; Milner *et al.*, 2010), it appears pertinent to investigate forefoot kinematic risk factors. How the tibia is loaded during gait appears important with respect to bone fatigue. A synchronised (in phase) loading of axial and torsional loads has been found to cause a dramatic increase in cortical bone fatigue (George and Vashishth, 2005b). The phase angle of loading during gait is related to the kinematics of the loading response, with changes in kinematics altering the phase angle. Greater rearfoot eversion at the rearfoot is one kinematic measure that is known to increase torsional loads and alter the phase angle of tibial loading. Given that forefoot motion is coupled to the rearfoot, and rearfoot motion to the shank, forefoot-rearfoot kinematics during loading may represent a potential mechanism for tibial stress injury.

It is generally considered that the foot can be successfully modelled as at least two segments (Davis, 2004), although numerous researchers have attempted to model in greater complexity (Leardini *et al.,* 1999; MacWilliams *et al.,* 2003). Work in the past decade has seen the development of multisegment foot models for use in research and clinical settings (Carson *et al.,* 2001; Hunt *et al.,* 2001; Myers *et al.,* 2004; Woodburn *et al.,* 2004). These foot models have allowed more detailed study of the relative motions occurring within the foot, although it should be noted that numerous simplifications and assumptions still persist. Given that this type of analysis is still in its relative infancy, experimental research using multisegment models is still not commonplace. One important factor related to the use of multisegment models is that these have predominantly been limited to barefoot applications. However, barefoot analysis may not replicate loading seen in shod conditions. This presents a challenge when researching risk factors for overuse running injuries. Some recent studies have employed gait sandals as a means of trying to replicate the cushioning properties experienced during shod gait (Branthwaite *et al.*, 2004; Eslami *et al.*, 2007; Morio *et al.*, 2009). However, to date, the use of a multisegment foot model designed to be worn with sandals has not been used to investigate risk factors for overuse injuries associated with the foot.

1.2 Purpose of the thesis

The overall purpose of this thesis was to address the research question; are forefoot-rearfoot kinematics risk factors for tibial stress injuries in runners? It was hypothesised that those with a history of tibial stress injury would demonstrate greater peak forefoot and rearfoot joint angles and joint velocities compared to controls. In order to answer this question, a series of experiments were designed with associated sub research questions. To establish what is already known on the topic, a systematic review of existing literature was designed. This sought to answer the research question; is there an association between foot type and the risk of tibial stress injuries? To address methodological considerations relating to the measurement of forefoot-rearfoot kinematics, a series of experiments were undertaken. Firstly, to answer the question; are gait sandals a viable means of replicating the kinematic motions seen during shod gait? Secondly, to answer the research question; can a reliable two segment foot model be developed for use with gait sandals during running? Based on the successful completion of this work, a relevant application of the developed model was designed. This study sought to answer the research question; do forefoot-rearfoot kinematics differ in high and low-arched individuals? These studies and their associated research questions represent important steps in attempting to answer the overall thesis question.

1.3 Structure of the Report

To address the main research question, this thesis has five further chapters structured as follows:-

Chapter II This chapter is divided into two sections. The first provides a comprehensive review of the literature in the areas relevant to the programme of research. The second section provides a detailed systematic review of the association between foot type and tibial stress injuries. Data are provided to assess this relationship and directions of future research are suggested.

Chapter III examines methodological considerations with the measurement of forefoot and rearfoot kinematics. The Chapter is divided into three sections. The first presents a multisegment foot model for use in the present thesis. The second section presents data to support the use of gait sandals as an effective means of measuring kinematic motions seen during shod gait. The final section addresses the between-sessions reliability of selected dependent variables.

Chapter IV presents an application of the developed model. The study investigates foot function in high and low-arched individuals and seeks to compare forefoot-rearfoot kinematic motions during gait.

Chapter V studies dynamic foot function in those with a history of tibial stress injury and matched controls. The study seeks to investigate forefoot-rearfoot kinematic motion as a potential injury risk factor.

Chapter VI is a summary and discussion of the thesis findings. It identifies implications the thesis findings will have within the area of biomechanics running injury research. The Chapter also identifies limitations of the thesis and suggests future directions in which the present research is proposed to progress. Finally, an overall conclusion is provided.

2 CHAPTER II

2.1 Review of literature

2.1.1 Introduction

This literature review will introduce research looking at the problem of overuse injuries within running and more specifically tibial stress injuries. Consideration will be given to the aetiology and diagnosis of tibial stress injuries as well as risk factors for their development, both intrinsic and extrinsic. Particular attention will be given to the role of both kinematic and kinetic biomechanical factors and their relationship to potential injury mechanisms. The structure and function of the foot as well as issues relating to the measurement and classification of foot type will be addressed. Subsequently, what is already known regarding the association between foot type and injury will be presented. The review will then progress onto the modelling of the human foot for biomechanical analysis. This will detail what is currently known regarding dynamic function within the foot. Approaches to modelling the foot and the challenges such approaches present will also be considered. Finally the review will address the issue of dynamic foot function as a risk factor for tibial stress injuries.

2.1.2 Running Injuries

Given the associated health benefits, running is an increasingly popular activity. Compared to other active populations, the running population are more susceptible to overuse injuries, including stress injuries, owing to the repetitive load associated with the activity (Hamill and Bates, 1988; Nigg *et al.*, 1995). Van Gent et al. (2007) reported the lower extremities to be the most at risk of injury. In a systematic review of running injuries, these authors found the reported incidence of lower extremity injuries ranged from 19% to 79%. The most common location of injury was at the knee (7% to 50%), followed by the lower leg (9% to 32%) and the foot (6% to 39%) (Van Gent *et al.*, 2007).

Investigation of injury risk factors may assist in early identification of those who are more prone to injury. Many intrinsic and extrinsic variables have been related to the development of running injuries, including age, body weight alignment, previous running and/or exercise experience, shoes, running terrain, and psychological factors (Van Gent *et al.*, 2007). Extrinsic risk factors are those related to training factors and can often be manipulated easily. Intrinsic factors are often concerned with biomechanical factors such as lower limb alignment or dynamic function. Van Mechelen et al. (1992) stated that only 4 factors were consistently related to running related injuries in recreational runners, that is, previous injury, a lack of running experience, running to compete, and excessive weekly running distance. A history of previous injuries has been found to be a risk factor for the development of subsequent lower extremity injuries (Walter *et al.*, 1989). These authors suggested that re-injury would be 50% more likely in all runners who had previously sustained a musculoskeletal injury. After initial injury, a lack of adaptation of extrinsic risk factors can often result in re-injury. However, specific intrinsic risk factors may also predispose certain populations to a higher risk of injury. Despite this, there is little agreement in the literature regarding specific factors which relate to injury. An increase in training distance per week has been found to be a protective factor against the development of knee injuries (Wen *et al.*, 1998; Satterthwaite *et al.,* 1999). However, a similar relationship was not observed with injuries in other lower extremity locations. Much of this conflicting literature may be due to studies investigating risk factors in association with groups of injuries or injury locations. Specific types of injuries are likely to have specific mechanisms and therefore risk factors. Further studies are needed which address these specific relationships.

2.1.3 Tibial Stress Injuries

2.1.3.1 Terminology

When reviewing tibial stress injuries, a significant problem lies with the terminology used to describe injuries in this anatomical area. 'Shin splints' is a term synonymous with non specific exercise related lower leg pain (Brukner *et al.*, 2001). The term may be used to describe a multitude of different injuries in this area, including; tibial and fibular stress fracture, tibial stress reactions, tibial periostitis, anterior and deep posterior compartment syndrome, popliteal artery entrapment and tibial posterior and anterior muscle strain or tendinitis (Beck, 1998). Distinguishing these injuries and using specific terminology is essential when investigating injury mechanisms in this anatomical area. Stress reactions are thought to be a precursor to a stress fracture, with a stress reaction eventually developing into a stress fracture without intervention (Jones *et al.*, 1989; Fredericson *et al.*, 1995). These two injuries are related along a continuum of bone micro damage and associated reparative responses (Batt *et al.*, 1998) (Figure 1). In their study of military recruits, Milgrom et al. (1984) initially reported negative bone scans in three recruits with documented tibial pain. Bone scans taken a month later revealed positive stress fractures. A further study found that a large portion of the injuries classified as stress fractures, actually displayed no evidence of a break in the bone or a fracture line (Jones *et al.*, 1989). It was suggested that stress injuries which did not result in bone failure but exhibited various stages of remodelling injury, could be classified as stress reactions (Jones *et al.*, 1989). Indeed, stress reactions in the tibia of athletes have been suggested to account for up to 75% of exercise induced lower leg pain (Umans and Kaye, 1996). If a stress reaction is a precursor to a fracture, it seems reasonable to assume that the two injuries are caused by the same mechanisms. Thus, in the present thesis, the term 'tibial stress injury' will include both tibial stress fractures and tibial stress reaction.

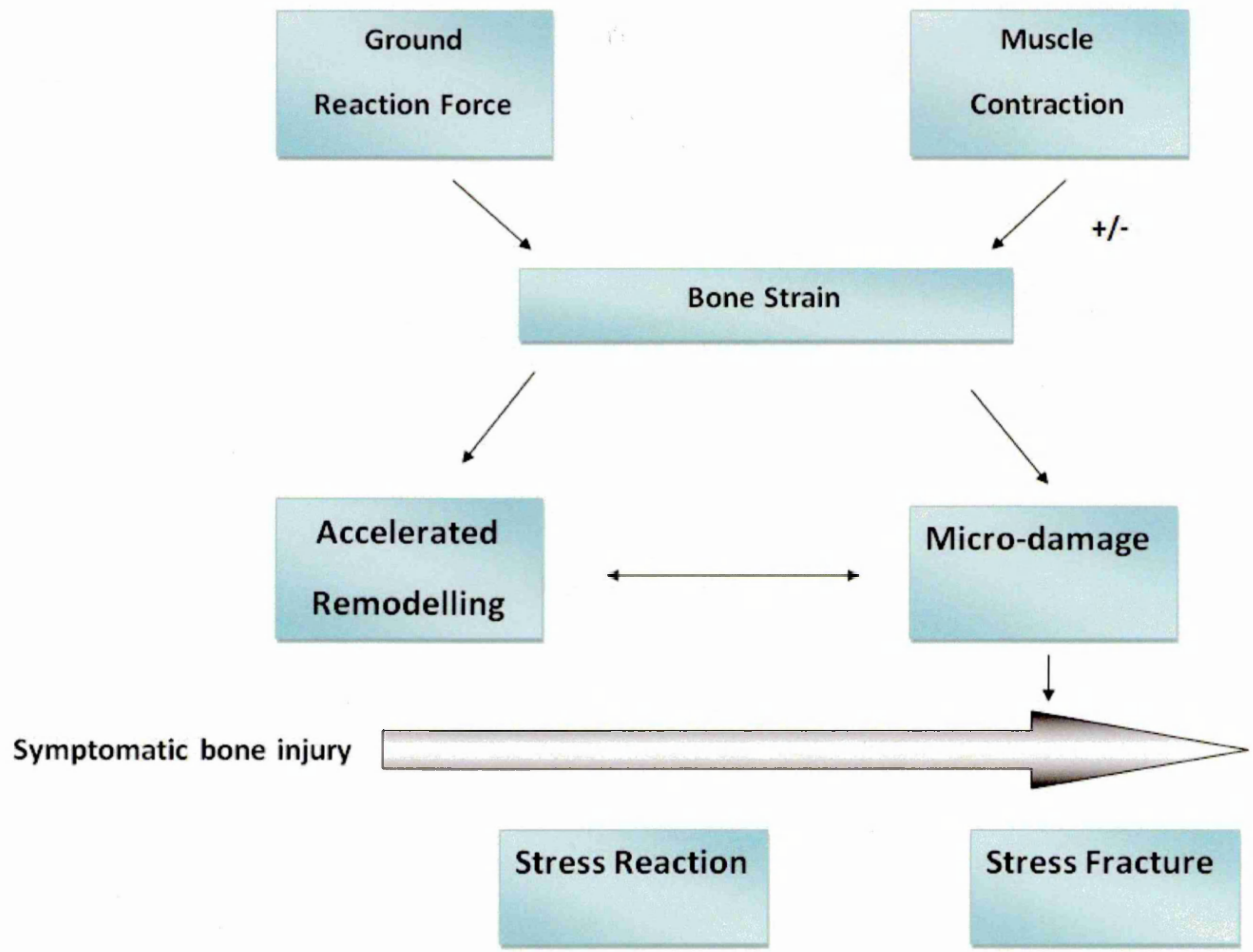


Figure 1. Pathogenesis of tibial stress injuries – taken from Bennell and Brukner (2005).

2.1.3.2 Aetiology

Stress fractures are focal structural weaknesses in bone, resulting from micro cracks which accumulate and eventually coalesce to form a stress fracture (Matheson *et al.*, 1987). They occur when the bone is unable to successfully adapt quickly enough to the repetitive loads imparted upon it. The nature of bone tissue means stress is required for normal development. Bone remodelling is a continuous process, involving osteoclastic resorption of bone followed by the formation of new bone (Burr *et al.,* 1990). The stimulus for this process is a combination of ground reaction forces and muscular contraction; both concentric and eccentric (Figure 1). Daily activities, particularly those involving weight bearing, generate the required stresses to stimulate the bone remodelling process (Bennell and Brukner, 2005). As a result of this loading, a number of different stresses are placed on the bone; these include compression, tension, shear, torsion and vibration (Beck, 1998). Increases in loading can lead to bone strengthening and has been shown to result in increased cortical thickness, density and widening of the diaphyseal bone diameter (Milgrom *et al.*, 1989).

While some loading of bones is essential for normal physiological development, too much loading can potentially cause injury. Fatigue is the progressive and localised structural damage that occurs when a material is subjected to cyclic loading. If the magnitude of this loading is sufficient, microcracks may form and eventually coalesce to form a fracture. During cyclic activities such as running, cortical bone is subjected to tensile, compressive and torsional loading (Vashishth *et al.,* 2001). However, little is known about physiological fatigue of cortical bone under a combination of these loads. Some limited evidence regarding fatigue damage may be drawn from in vitro studies subjecting bovine cortical bone to various loads (George and Vashishth, 2005a).These authors found a rapid initial bone stiffness loss when subjected to tension, where as in compression, stiffness loss was only observed after 90% of fatigue life (Figure 2). This data supports the notion that bone is weaker in tension and cracks more likely to initiate than when in compression. Furthermore, these authors found torsional loading represented the most aggressive damage mechanism with the greatest stiffness loss (Figure 2). Results suggested that fatigue of cortical bone is uniquely related to the individual components of physiological loading (tension, compression, torsion).

Using a rabbit tibiae model, Burr et al. (1990) found the location of stress fracture to coincide with the site of maximum bone loading. However, it is unlikely that increased load alone causes sufficient stress to lead to the development of a stress fracture (Beck, 1998). This increased loading at specific locations may occur in conjunction with increased bone porosity as a result of accelerated remodelling (Beck, 1998). Accelerated remodelling can occur if the level of bone resorption and bone replacement is not maintained, this can lead to increased bone porosity and weakening of the bone (Bennell and Brukner, 2005) (Figure 1). It has been hypothesised that an accumulation of mechanical forces results in accelerated remodelling (Hershman and Mailly, 1990; Nattive and Armsey, 1997). If the 'dose' of loading during activities such as running is excessive, this may provide the necessary stimulus for accelerated remodelling in bone.

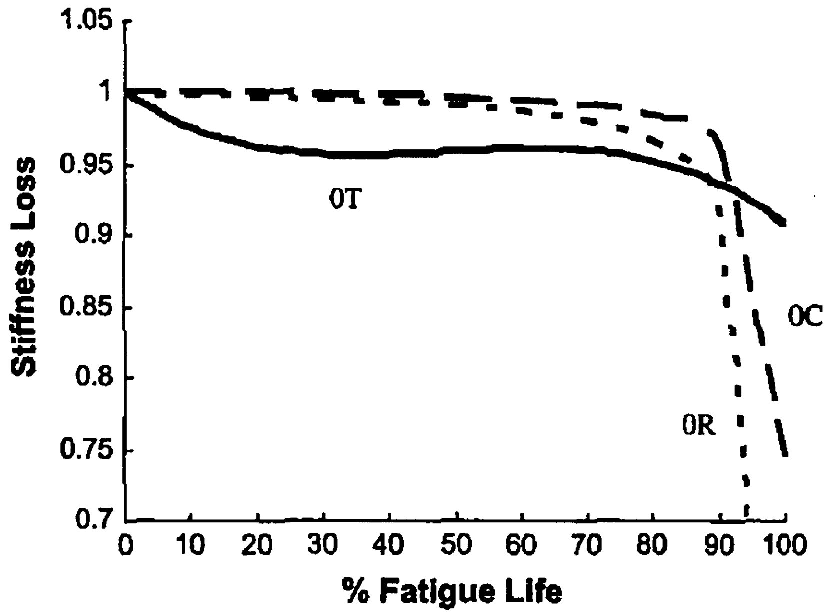


Figure 2. Average stiffness loss profiles for specimens subjected to Zero-Tension (OT), Zero-Compression (OC), and Zero-Torsion (OR). Taken from George and Vashishth (2005a).

Thus, if the dose of loading remains unchanged and microdamage accumulates and cannot be repaired through remodelling, injury will ensue. Therefore, stress injuries are likely caused by increased bone porosity coupled with a high dose of localised loading (Beck, 1998).

Ground reaction forces and contractile muscular forces have both been suggested to contribute to the fatigue of bone (Beck, 1998) (Figure 1). Goergen et al. (1981), noted stress fractures are a result of muscular action on bone. Lanyon et al. (1975) also reported that a combination of bodyweight and muscle tension resulted in tibial deformations during gait. Muscles have been found to have both a positive and negative effect on the loading of bone (Scott and Winter, 1990). Muscular contractile forces may increase the strain on bone, but at other sites may serve as a shock absorption mechanism. This mechanism involves the contraction of muscle in the opposite direction to the bone bending moment, serving to dissipate and neutralise stress (Grimston and Zernicke, 1993). Reduced muscle mass has been found to be a risk factor for the development of stress fractures in athletes (Bennell *et al.*, 1996). Given the important reported role of muscle in the aetiology of stress fracture, muscle fatigue is likely to affect this relationship. Both strain and strain rate of long bones have been found to increase as a result of fatiguing exercise (Fyhrie *et al.*, 1998). Further, the body's ability to attenuate ground impact forces during running have also been reported to be significantly reduced in a fatigued state (Mercer *et al.*, 2003). Therefore, a state of muscular fatigue may have a dual role in the mechanism of stress fracture, through increased ground impact forces coupled with a reduced ability to limit localised bone strain.

Ground reaction forces are a dominant factor in creating stress and strain which must be attenuated by the structures of the lower limb. The repetitive strain imparted on bone during running is dependent not only on the structure and properties of the bone, but also the magnitude of the applied load, the rate of loading and the number of loading cycles (Bennell *et al.*, 2004). Those with a history of tibial stress injuries may exhibit unique loading patterns compared to an un-injured population.

During gait, impact forces result in the tibia being exposed to a combination of bending, shearing and axial compressive forces (Ekenman *et al.*, 1998). Tibial strain has successfully been measured in gait and jumping through *in vivo* surgical attachment of bone strain gauges (Burr *et al.*, 1996; Milgrom *et al.*, 2000). Burr et al. (1996) found running produced tibial strains two to three times greater than walking. However, when studying jumping, increasing jump height was not found to increase tibial strain, with landing values comparable to those measured during running (Milgrom *et al.*, 2000). This suggests the body may employ mechanisms to limit high levels of tibial strain. Therefore, it may be that a high number of sub maximal loading cycles are more important than load magnitude. These studies provide useful insight into local tibial strain during load bearing activities, but are limited by the measurement of strain at a single location on the bone. Furthermore, the invasive nature of these procedures and ethical constraints of such methodology, limit their use in routine biomechanical analysis (Bennell *et al.*, 2004). Using a probalistic stress fracture model, Edwards and colleagues (2010) estimated tibial strain when running at three different speeds. A combination of tibial bending and axial compression was found which resulted in principle compressive strain on the posterior surface of the bone (Figure 3). This location of maximal compressive strain coincides with the most common site of tibial stress fractures in runners, the medial posterior border (Beck, 1998). However, it should be noted that these authors used a modelling approach. Absolute tibial strain values were not comparable to *in vivo* measurement during running, with strain found to be significantly higher than those reported by Burr et al. (1996). However, in agreement with these authors, Edwards et al. (2010) reported reduced tibial strain associated with slower running speeds. Data suggested that the risk of stress fracture development was more dependent on loading magnitude rather than the loading exposure.

In summary, the aetiology of stress fractures are still not fully understood. A degree of loading is necessary for improving bone strength and density. However, too much load can lead to microdamage in bone, eventually leading to a stress injury. Whether this is due to the dose of loading or loading magnitude remains unclear. Evidence does suggest the fatigue life of bone is related to the individual components of physiological load and how they are applied. Further, the body appears to use mechanisms to regulate this load at specific sites. Therefore, further investigation of these unique loading patterns is warranted in injured populations.

2.1.3.3 Incidence

Of all the musculoskeletal injuries, stress fractures are thought to be one of the most common and potentially serious overuse injuries (Jones *et al.*, 2002). Although stress fractures can occur in any bone of the body, they are most frequently observed in the lower extremities, particularly in runners. Fifteen to 20% of all musculoskeletal injuries have been attributed to stress fractures (Matheson *et al.,* 1987; Bennell *et al.*, 1999). The tibia is the most common site of stress fracture, with tibial stress fractures accounting for between 35% and 49% of all stress fractures in this population (McBryde, 1985; Matheson *et al.,* 1987).

Other high risk populations include military recruits, in whom a similar injury incidence of 41% has been reported (Beck *et al.*, 1996). Female runners are reported to be at higher risk than males. Arendt et al. (2003), reported a twofold increase in the incidence of stress fractures over men, while a further study found up to four times the incidence compared to males (Hauret *et al.*, 2001). The most common site of stress fracture is the junction of the middle and distal thirds of the tibial diaphysis (Fredericson *et al.*, 1995). Other less common sites of injury include proximal tibial stress fractures (Coady and Micheli, 1997) and those of the mid-anterior cortex (Rolf *et al.*, 1997), which, although rare prove problematic, often with a history of non-union. Fibula stress fractures are also less common but tend to occur in the distal portion of the bone (Fredericson, 2003). Since tibial stress injuries are the most common type of injury, further research of this injury is needed in these at risk populations.

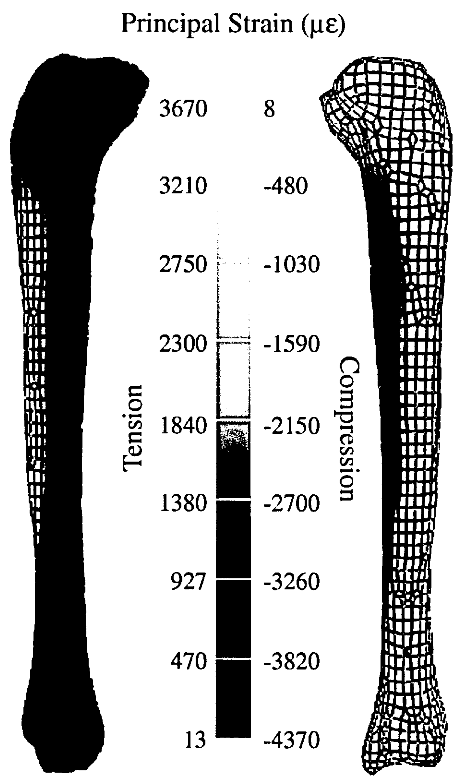


Figure 3. Sagittal views of a representative finite element model of the tibia displaying maximum principal strains during running. Note: peak compression on the posterior surface (right). Taken from Edwards et al. (2010).

2.1.3.4 Implications

The occurrence of a tibial stress injury can have significant implications, particularly for athletes and military recruits. The implications are potentially severe with significant loss of training in both groups and considerable individual frustration (Beck, 1998). The serious nature of a stress fracture requires a period of non-weight bearing and an extended recovery without running (James *et al.*, 1978). Suggested periods of immobilisation range from 3-12 weeks (Beck, 1998; Brukner *et al.*, 1998). Casting of the limb can have significant negative repercussions, including reductions in bone mineral density, muscle tissue wastage, as well as serious disruption to daily life. Full recovery from a tibial stress fracture has been reported to take up to 19 weeks in both athletes (Harmon, 2003) and military personnel (Ross and Allsopp, 2002). Tibial stress reactions can be managed by taking a more conservative approach. These often don't require immobilisation and heal without complication, usually permitting a return to sport within 4-8 weeks (Bennell and Brukner, 2005). Given the serious nature of these injuries, an understanding of risk factors is crucial to allow future preventative strategies to be developed.

2.1.3.5 Diagnosis

The diagnosis of stress fractures and, more so, stress reactions, can be problematic (Fredericson *et al.*, 1995). However, early diagnosis of tibial stress reactions is essential to allow appropriate precautions to be taken to prevent progression to a full fracture. These injuries may be diagnosed through a detailed medical history coupled with clinical examination (Bennell and Brukner, 2005). A thorough training history can often provide detailed insight into the nature of the injury; this should focus on the injury onset, mechanism of injury, location of pain, associated symptoms, alleviating or aggravating factors, specific timing of pain and training changes (Fredericson and Wun, 2003). Tibial stress injuries are characterised by a gradual onset of pain associated with repetitive activity. Initially pain may persist following training, with continued stress leading to pain during the activity (Fredericson *et al.*, 1995). Furthermore, pain when walking and pain over the tibial shaft with bone percussion are also often present. Diffuse pain along the posteromedial tibia (usually at least 5cm) is often present in a stress reaction, with a focus of tenderness more suggestive of a stress fracture (Batt *et al.*, 1998). On some occasions, noticeable swelling at the site of injury is also present (McBryde, 1985). Despite a common set of reported clinical symptoms, difficulty can arise from distinguishing stress injuries from other injuries in the same region which can display similar clinical symptoms, these include; undefined shin splints, medial tibial stress syndrome and compartment syndrome (Batt *et al.*, 1998).

The use of appropriate imaging technologies can often confirm clinical suspicion of a tibial stress injury. Radiographic X-rays have been used to diagnose stress fractures (Savoca, 1971). However, they only offer a sensitivity of diagnosis of approximately 26%, with 50% of stress fractures not detected on an X-ray (Tuan, 2004). Further, the diagnosis of stress reactions using x-rays is extremely difficult (Figure 4). The onset of symptoms tends to precede any positive findings on x-rays by at least three weeks (Tuan, 2004). A similar lack of sensitivity is offered by computerized tomography scanning (Figure 4), but these scans have been found to be an effective means of diagnosing longitudinal stress fractures (Roebuck *et al.*, 2001). Triple bone scans are a far more sensitive means of diagnosis, with a sensitivity approaching 100% (Shikare, 1997). This technology is also able to detect low bone activity and displays localised areas of increased uptake 'hot spots' (Bennell and Brukner, 2005), thereby making them powerful tools in the diagnosis of stress reactions.

More recently, magnetic resonance imaging (MRI) has emerged as a viable means of diagnosing tibial stress injuries (Batt *et al.*, 1998; Gaeta *et al.*, 2005). Gaeta et al. (2005) found MRI to be the single best method for the diagnosis of tibial stress injuries. MRI offers a similar high level of sensitivity to bone scans and has the advantage of showing any related soft tissue damage (Gaeta *et al.*, 2005). Common clinical symptoms have been found to correlate well with more severe tibial stress identified on both MRI and bone scans (Fredericson *et al.*, 1995). There is also some evidence to support the ability of MRI to detect early silent stress reactions (Bergman *et al.*, 2004). When studying 21 asymptomatic college runners, these authors reported bone abnormalities characteristic of a stress reaction in 43% of cases. Therefore, MRI appears to offer a sensitive means of early identification of injury risk. However, these abnormalities were not found to predict future incidence of tibial stress injury in any of the asymptomatic runners when followed for up to four years (Bergman *et al.*, 2004). MRI also has the advantage of no radiation exposure compared to other imaging technology, but is more costly and access to machines often limits its use (Bennell and Brukner, 2005).

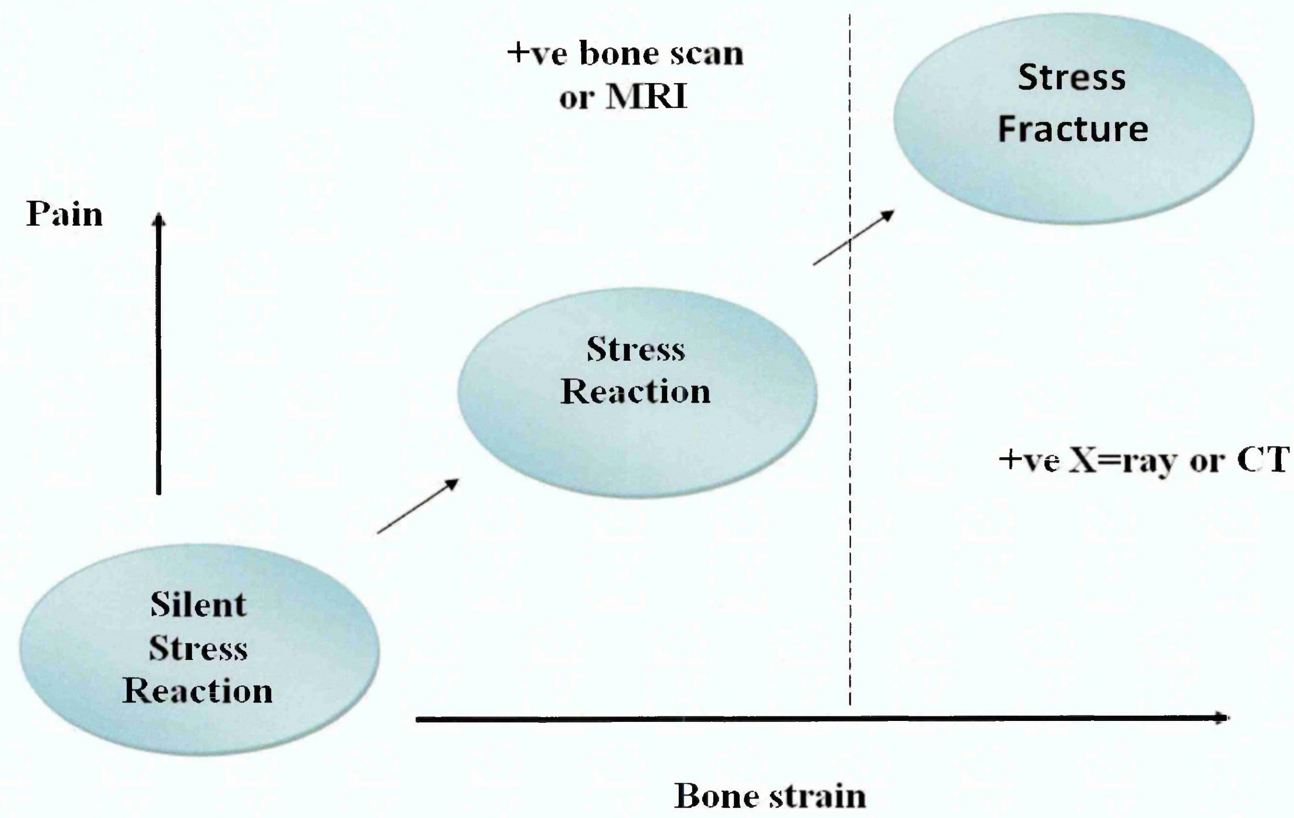


Figure 4. The bone strain continuum and its relationship to diagnostic findings –taken from Bennell and Brukner (2005).

In summary, a number of scanning technologies are currently available for fast accurate diagnosis of tibial stress injuries. However, high costs and access often limit their use for routine diagnosis. Careful clinical examination coupled with a detailed medical history have been found to offer a quick inexpensive means of injury diagnosis. Tibial stress injuries often present with a clear set of well defined clinical symptoms which allow for accurate reliable diagnosis through clinical examination.

2.1.4 Risk Factors

2.1.4.1 Extrinsic

Various risk factors, both intrinsic and extrinsic have been linked with an increased risk of tibial stress fracture (Beck, 1998). Extrinsic factors implicated include; footwear, running surface, weekly mileage, training adaptation and injury history (Kowal, 1980; Scully and Besterman, 1982; Greaney *et al.*, 1983; Montgomery *et al.*, 1989; Beck, 1996).

Given that tibial stress fractures appear to be related to a 'dose' of loading, factors which change this load are likely important. Increases in training loads have been found to increase the risk of tibial stress injuries (Fredericson *et al.*, 1995; Matheson *et al.*, 1987). This is often evident in military recruits who begin intensive training having led a relatively sedentary lifestyle. The significant increase in the magnitude, frequency and volume of the load, results in a failure of bone to adapt quickly enough to these demands (Fredericson *et al.*, 1995). In military recruits, the majority of stress fractures have been reported to occur over the first month of training (Montgomery *et al.*, 1989). This period coincides with the period of increased bone porosity as part of the remodelling process (Beck, 1998). Not surprisingly, the type of activity has been implicated strongly as a risk factor for tibial stress fracture, with high impact activities such as running, marching and jumping often responsible (Reeder *et al.*, 1996).

The use of appropriate footwear has been suggested to lower the risk of tibial stress injury (Beck, 1998). Associated issues with training shoes and injury risk may include; a lack of appropriate cushioning or support, an inappropriate match between footwear and foot type and the potential for shoes to become worn out (Bennell and Brukner, 2005). Fredericson et al. (1996) suggested that depending on other factors, footwear should be changed after 500-700km of running. Military training has traditionally been conducted in boots offering limited cushioning (Giladi *et al.*, 1985; Simkin *et al.*, 1989). In a systematic review of intervention studies, Gillespie and Grant (2000), found the use of insoles within military boots reduced the incidence of stress injuries by 50%.

Like footwear, terrain can also have a significant influence on the magnitude of tibial stress during weight bearing exercise. The notion that running on hard surfaces increases the risk of injury appears logical and is supported in the literature (James and Bates, 1978; Fredericson *et al.*, 1996). Furthermore, greater tibial strain has been reported when running up hill (Burr *et al.*, 1996). In the main, a level uniform surface of moderate firmness appears to provide the least stressful running surface (Beck, 1998).

Whether tibial stress injuries are related to either 'dose' or magnitude of loading, extrinsic factors appear to be strongly implicating in changing the external and internal loading within the body. Many of these extrinsic factors such as footwear and terrain can easily be modified. However, the ways in which external loads are measured and their relationship to internal loading is important in understanding injury mechanisms.

2.1.4.1.1 External loading

Vertical ground reaction forces of between two and four times body weight are associated with each foot strike during running (Nigg *et al.*, 1995). Ground reaction forces serve as a proxy measure of the magnitude and rate of loading on the lower extremity (Bassey, 1997). Increased ground reaction forces would likely result in greater bending moments experienced by the tibia (Milner *et al.*, 2006b). If the development of tibial stress injuries is associated with loading parameters, one would expect differences in ground reaction force characteristics between injured and un-injured groups (Grimston *et al.,* 1991). Grimston and colleagues (1991) found greater vertical and medial-lateral forces in those with a history of stress fracture compared to a control group. However, a study on a similar population (Grimston *et al.*, 1994) found greater vertical and anterior-posterior forces in a control group compared to those with a history of stress fracture. More recent studies have failed to find any difference in ground reaction force variables between those with and without a tibial stress fracture in both male (Crossley *et al.,* 1999) and female (Bennell *et al.,* 2004) athletes. Further, studies of this nature should be viewed with a degree of caution, as the relationship between internal loading and external markers of loading are often complex and non-intuitive (Miller and Hamill, 2009). One important issue arises from the study of Grimston and colleagues (1994) in aiding our understanding of the relationship between external loading, injury and fatigue. In their study of female runners with stress fractures, greater ground reaction forces were reported in the control group. However, after a 45 minute run, it was the stress fracture group who exhibited greater vertical and anterior-posterior forces compared to the controls. This suggests those with a history of stress fracture may exhibit a fatigue profile which potentially increases their risk of injury. Medial tensile bone strains have been reported to become increased in a fatigued state compared to rested (Milgrom *et al.*, 2007). Therefore, while numerous studies report no group difference in ground reaction force variables in a non-fatigued state (Crossley *et al.,* 1999; Bennell *et al.,* 2004), ground reaction forces when fatigued may give more insight into an individual's risk of stress injury.

More recent studies have provided some evidence of external loading differences in those with a history of tibial stress fracture (Milner, 2006b; Creaby and Dixon 2008). In line with previous studies, the magnitude of ground reaction force variables, in those with and without a history of tibial stress fractures, were found to be similar (Milner *et al.,* 2006; Creaby and Dixon, 2008). However, Milner et al. (2006) reported greater instantaneous (tibial stress fracture = 79.0BW/s vs. Control = 66.3BW/s) and average vertical loading rates (tibial stress fracture = 92.6BW/s vs. Control = 79.6BW/s) in the stress fracture group. Using a simulation modelling approach, Miller and Hamill (2009) found increases in tibial shear and compression loading estimates were accompanied by an increase in vertical ground reaction force loading rate. However, the authors did not report an increase in vertical force magnitude, suggesting loading rate to be a more important surrogate measure of tibial loading than simply the magnitude of force. Creaby and Dixon (2008) studied the angle of the ground reaction force vector in military recruits with a history of tibial stress fracture during gait. Differences were identified in the direction of the frontal plane ground reaction force vector at midstance, with a more medially directed vector seen in the tibial stress fracture group. No differences in any variables were seen during the initial loading phase of gait. It was suggested that this more medial ground reaction vector increased the moment arm to the tibia, resulting in increased bending moment acting on the bone (Creaby and Dixon, 2008). However, the vector was not calculated in relation to tibial position so support for this mechanism was not provided. Future work is needed to explore the relationship between ground reaction forces and tibial position through stance.

Free moment is a ground reaction force variable that can be measured using a force platform and provides a measure of the torque between the foot and ground (Holden and Cavanagh, 1991). The tight nature of the ankle mortise between the rearfoot and tibia means free moment can provide a surrogate measure of the torsional loads applied to the tibia (Milner *et al.*, 2006a). Given that during gait the tibia is exposed to numerous loads including torsion (Beck, 1998), it is likely that free moment may be an important variable in identifying tibial stress fracture risk. Indeed, recently, free moment has been found to be a risk factor for development of tibial stress fractures (Milner *et al.*, 2006a; Pohl *et al.*, 2008). Milner et al. (2006a) reported greater adduction and absolute free moment in those with a history of tibial stress fracture compared to controls.

Further, these authors reported that absolute free moment successfully predicted a history of tibial stress fracture in 66% of cases. Similar findings were also reported by Pohl et al. (2008). However, these studies were both limited to female running populations. A study of male military recruits failed to find a difference in free moment values between those with a history of tibial stress fracture and a matched control group (Creaby and Dixon, 2008).

It has been suggested that the early observed oscillations in ground reaction force data when running are the result of the shock wave propagating up the leg due to a sudden deceleration of the body (Bobbert *et al.*, 1991). It is thought that the high frequency components of this shock wave may be related to the injury of structures within the body (Johnson, 1986). While ground reaction force measurement is common, this measure only reflects the accelerations experienced by the whole body's centre of mass. A more sensitive measure can be obtained through the use of accelerometry to measure localised tibial accelerations on the limb at the site of injury. However, accelerometry has been characterised as one of the hardest techniques from which to gain accurate, reliable results in biomechanics (Valiant, 1990). Studies measuring tibial accelerations using surgically mounted bone transducers have been conducted during gait (Hennig and Lafortune 1989; Lafortune, 1991; Lafortune *et al.*, 1995). However, ethical constraints associated with these procedures have limited their use in routine analysis. Skin mounted transducers are more widely used in biomechanical analysis (Laughton *et al.*, 2003; Butler *et al.*, 2006). The accelerometer may be attached directly to the segment of interest at a position of minimal soft tissue movement, on the tibia this is often the distal anterior medial aspect of the bone (Laughton *et al.*, 2003). Despite this, the accuracy of skin mounted transducers has been questioned (Lafortune, 1991). Comparisons of bone and skin mounted transducers have found skin measurements to overestimate accelerations at the tibia (Hennig and Lafortune, 1989; Lafortune *et al.*, 1995). However, skin transducers are used in routine analysis (Laughton *et al.*, 2003; Butler *et al.*, 2006; Milner *et al.*, 2006b; Zifchcock *et al.*, 2006a; Milner *et al.*, 2007), so careful location selection and strict transducer application procedures are required.

Vertical ground reaction force loading rates have been found to be positively correlated to peak tibial shock during running (Hennig and Lafortune, 1989; Laughton *et al.*, 2003). Tibial shock serves as a proxy measure of bone loading and offers a more localised measure than ground reaction force variables (Milner *et al.,* 2006b). Peak tibial shock has been found to be a discriminating variable between those with a history of tibial stress fracture and controls (Milner *et al.*, 2006b; Milner *et al.*, 2007; Zifchcock *et al.*, 2006a). Milner et al. (2006b) found higher tibial shock in the stress fracture (7.7*g*) compared to the control group (5.8*g*). Tibial shock was found to be a more sensitive measure for predicting the risk of tibial stress fracture compared to ground reaction force parameters. Using logistic regression, tibial shock was found to be a strong predictor of injury risk, with its magnitude found to predict the incidence of tibial stress fracture in 70% of cases (Milner *et al.*, 2006b). Zifchcock and colleagues (2006a), compared peak tibial shock in symptomatic and asymptomatic limbs of previously injured female runners. Tibial shock was found to be 15.8% higher in the symptomatic compared to the asymptomatic limb. Higher tibial shock has also been noted for those with a history of lower extremity stress fracture when compared to controls (9.2*g* stress fracture group vs. 7.2*g* control group) (Ferber *et al.*, 2002). However, it should be noted that this study was not specific to tibial stress fractures and included any stress fracture of the lower extremity. All these studies provide strong evidence implicating high tibial shock as a risk factor for sustaining a stress fracture. However, the retrospective design of these studies makes it difficult to attribute tibial stress fracture to higher shock level, as it remains unclear whether these observations are the cause or effect of the injury. Early results of a prospective study have recorded greater tibial shock in the stress fracture group compared to controls (9.1*g* stress fracture group vs. 4.7*g* control group) (Davis *et al.*, 2004). These data suggest greater shock is a risk factor for the development of tibial stress fracture. However, it should be noted that the injury group contained only five subjects and these results should be treated with caution. Prospective studies are required to establish the cause and effect relationship between tibial stress injuries and shock related variables.

In summary, ground reaction forces appear to be the main external contributor to increased bone stain during gait. Despite this, evidence is conflicting regarding some ground reaction force variables as risk factors for tibial stress injuries. Variables which have been supported as risk factors include instantaneous and average loading rate, as well as free moment. Despite the ease of measurement, ground reaction forces do not measure localised accelerations within the body, accelerometry offers a more sensitive measure. Tibial shock serves as a proxy measure for bone loading and has been found to be greater in those with a history of tibial stress injuries compared to controls.

2.1.4.2 Intrinsic

Despite the highlighted importance of extrinsic factors, intrinsic factors have been associated with overuse running injuries in up to 40% of cases (Lysholm and Wiklander, 1987). As already noted females have significantly higher risk of tibial stress fracture than males. This risk is likely increased in those females suffering from female athlete triad, which involves a combination of osteoporosis, amenorrhea and eating disorders (Otis *et al.*, 1997; Fredericson, and Wun, 2003). Menstrual disturbance is common in sports women and is a factor that has been suggested to increase stress fracture risk (Bennell and Brukner, 2005). Low bone mineral density is linked to amenorrhoea and has been observed in trained females who present with stress fracture (Myburgh *et al.*, 1990). However, Grimston et al. (1991) found no difference in tibial bone density when comparing female runners with a history of stress fracture to controls. A further factor relates to dietary intake and eating habits. In particular, low energy availability can increase the risk of stress fracture (Bennell *et al.*, 1996). Low energy availability has been found to uncouple bone turnover and suppress the formation of new tissue (Ihle and Loucks, 2004).

Structural risk factors associated with the tibia are well established. Both a smaller mediolateral tibial width (Giladi *et al.*, 1987) and tibial moment of inertia (Milgrom et *al.*, 1989) have been identified as risk factors for tibial stress fractures in male military recruits. Furthermore, a smaller tibial cross-sectional area was found to be a risk factor in male runners (Crossley *et al.*, 1999). Interestingly, similar findings have not been observed in females (Bennell *et al.*, 2004). Bennell et al. (2004) measured several bone parameters on the tibia in female runners. They reported similar bone density, bone width and bone cross-sectional area between injury and control groups.

Intrinsic factors relating to lower extremity alignment have also been linked to tibial stress injury risk. However, there appears to be little agreement in the literature surrounding these factors (Beck, 1998). A leg length discrepancy was noted in 70% of athletes presenting with a stress fracture (Bennell *et al.*, 1996), but the injury was observed in both the shorter and longer limbs. Military recruits with a Q angle of greater than 15º were also found to be at an increased risk of developing a stress fracture (Cowan *et al.*, 1996). However, it should be noted that both these studies were not specific to stress fractures of the tibia and their results should be interpreted cautiously.

Given the strong association between loading and tibial stress injury risk, lower extremity kinematic variables are likely to be of importance in determining the nature of these loads experienced during gait. Altered lower extremity kinematics may be expected in those with a history of tibial stress injuries. Peak hip adduction during stance has been found to be a distinguishing variable in female runners with a history of tibial stress fracture when compared to controls (Pohl et *al.*, 2008; Milner *et al.*, 2010). A greater peak adduction of 3.5º (Milner *et al.*, 2010) and 4º (Pohl et *al.*, 2008) was reported in the stress fracture compared to the control group. One possible theory suggests this increased hip adduction may result in a lateral shift of the ground reaction force vector, placing greater axial load on the lateral aspect of the knee (Pohl *et al.*, 2008). This may result in increased compression of the lateral aspect of the bone and resulting tension on the medial side of the tibial diaphysis. Both of these studies analysed female runners and similar findings have not been reported in male military recruits (Creaby and Dixon, 2008). Creaby and Dixon (2008) found the ground reaction force vector to shift in a more medial direction in military recruits with a history of stress fracture, thereby contradicting the theory proposed by Pohl and colleagues (2008). However, it should be noted that in neither study was the direction of the ground reaction force vector directly related to frontal plane kinematics at the hip.

Sagittal plane joint stiffness in the lower extremity may be important in determining the body's ability to attenuate the shock associated with impact. Increased knee range of motion has been found to reduce peak loading when landing from jumps (McNitt-Gray *et al.*, 1994). This can be achieved by increasing the time over which the body is decelerated. A similar relationship between knee mechanics and loading may be expected during the stance phase of gait. Milner et al. (2007) examined knee mechanics in a group of female runners with a history of tibial stress fracture and controls. Both knee excursion and knee flexion at heel strike were found to be similar between groups. However, the stress fracture group were found to run with a stiffer knee, the result of a greater change in knee moment than seen in the control group. Furthermore, this stiffness was found to be positively correlated to tibial shock. These data suggest knee stiffness during gait to be a risk factor but a similar relationship has not been seen for ankle joint stiffness. In a different study, these authors reported significantly greater ankle stiffness in the control rather than the stress fracture group (Milner *et al.*, 2006b).

In addition to extrinsic factors, numerous intrinsic ones have been suggested to increase the risk of sustaining a tibial stress injury. These include physiological and structural factors, as well as alignment of the lower extremity. Since the loads experienced by the lower extremity during gait appear important, kinematic factors which alter these loads should be investigated. Some limited evidence has been presented to implicate hip adduction and rearfoot eversion as intrinsic risk factors for tibial stress injury.

2.1.5 The Foot

Foot morphology has been identified as an intrinsic risk factor for injury (Giladi *et al.*, 1985; Simkin *et al.*, 1989; Cowan *et al.*, 1993). The human foot is an extremely complex structure, with foot characteristics varying widely between individuals (Razeghi and Batt, 2002). The foot contains 26 bones (Figure 5) and numerous ligaments and muscles. The ligaments of the foot limit its motion and provide stability to its joints, but these are purely passive soft tissue structures. Intrinsic muscles provide active connections between bones within the foot, while extrinsic muscles link bones in the foot with bones in the upper and lower leg. These muscles along with ligaments and bone geometry determine the range of motion within the joints of the foot (Nester, 2009). Although the foot possesses many common anatomical characteristics, variation in the shape and biomechanics of the structures allow the classification of different foot types.

2.1.5.1 Foot Type Classification

To explore any association between foot type and injury, methods of foot classification must be considered. In their review of current literature, Razeghi and Batt (2002), suggest there is no clear consensus on an ideal method for foot type classification. Previous attempts classifying foot type fall loosely into four main areas; qualitative inspection, anthropometric measures, plantar pressure analysis and radiographic assessment. Qualitative classification is easy to employ and is often used by clinicians to make quick judgements about foot type. These judgements are often taken in either static or barefoot walking conditions. However, these assessments are subjective and agreement between clinicians have been found to be poor (Dahle *et al.,* 1991; Cowan *et al.*, 1994). Anthropometric measures on the foot offer a more objective approach to the classification of foot type (Razeghi and Batt, 2002). Measurements predominantly focus on the sagittal or frontal plane orientation of the foot segments. Typical sagittal plane measures of foot type include arch height (Hawes *et al.*, 1992), longitudinal arch angle (Cashmere *et al.*, 1999), and navicular drop (Saltzman *et al.*, 1995). Common frontal plane measures include the measurement of rearfoot angle (Kernozek *et al.*, 1993) and valgus foot index (Rose *et al.*, 1985).

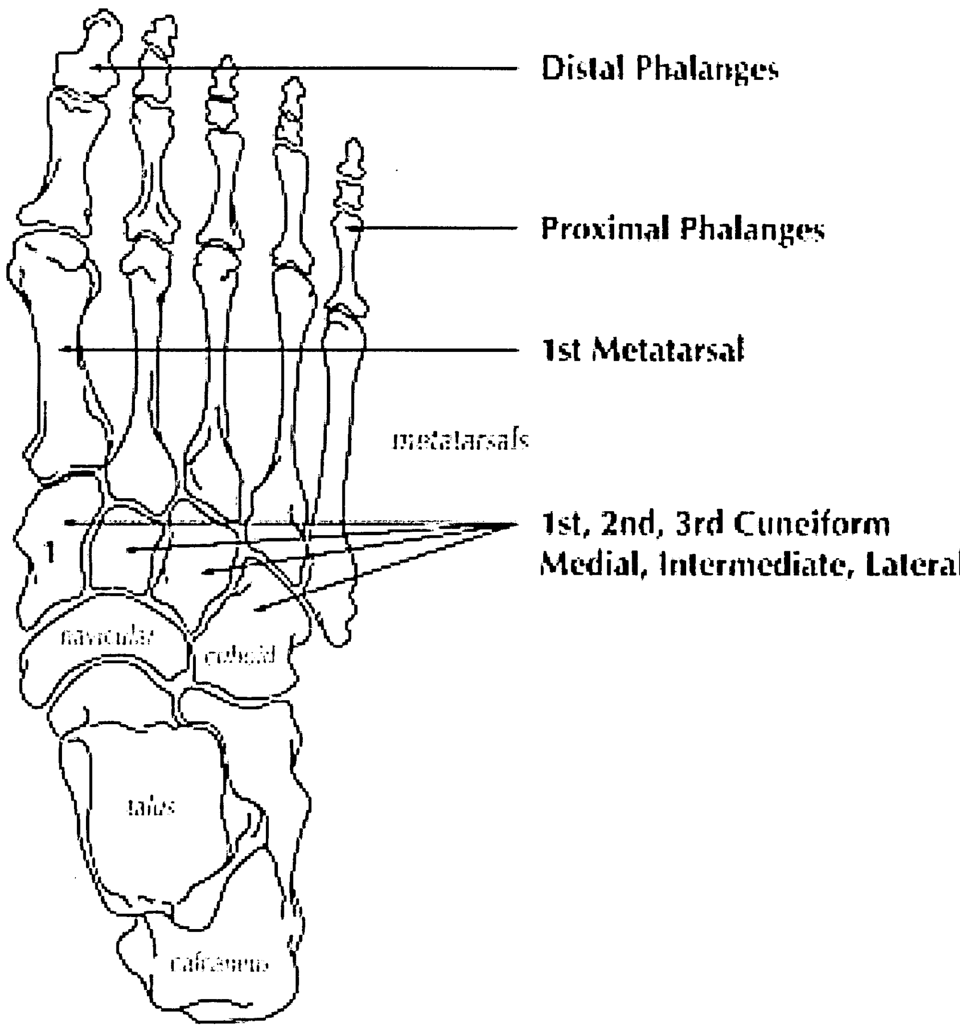


Figure 5. Bones of foot which make up the four functional units described by Wolf et al. (2008).

Qualitative classification is easy to employ and is often used by clinicians to make quick judgements about foot type. These judgements are often taken in either static or barefoot walking conditions. However, these assessments are subjective and agreement between clinicians have been found to be poor (Dahle *et al.,* 1991; Cowan *et al.*, 1994). Anthropometric measures on the foot offer a more objective approach to the classification of foot type (Razeghi and Batt, 2002). Measurements predominantly focus on the sagittal or frontal plane orientation of the foot segments. Typical sagittal plane measures of foot type include arch height (Hawes *et al.*, 1992), longitudinal arch angle (Cashmere *et al.*, 1999), and navicular drop (Saltzman *et al.*, 1995). Common frontal plane measures include the measurement of rearfoot angle (Kernozek *et al.*, 1993) and valgus foot index (Rose *et al.*, 1985).

Further measures such as navicular drift have been used in an attempt to quantify transverse plane motion of this bone as a means of classifying foot type (Menz 1998). While such methods may offer a more objective measure of foot type, they are limited to one plane of motion and often fail to account for the triplanar characteristics which are seen in different foot types. Some authors have combined planar measures in attempt to provide an improved classification of foot type (Sneyers *et al.*, 1995; Song *et al.*, 1996). Song et al. used calcaneal angle, subtalar position and forefoot-rearfoot alignment to classify feet as planus, rectus or cavus. However, Syneyers et al. (1995) used a combination of arch height, leg to rearfoot alignment and forefoot to rearfoot alignment to classify feet into the same three categories. While such methods do attempt to incorporate triplanar foot characteristics, a major limitation with their use remains the fact they are based on static measurements and fail to account for dynamic foot function.

A further approach to classifying the foot is through the use of plantar pressure analysis. This relies on structural changes within the foot resulting in changes in the shape and orientation of the plantar imprint (Razeghi and Batt, 2002). Numerous authors have devised indices based on plantar pressure measures to classify the foot (Cavanagh *et al.*, 1987; Hawes *et al.*, 1992; Chu *et al.*, 1995). Plantar pressure methods offer the potential advantage of making dynamic assessments of foot type. However, any link between plantar foot pressures and foot type remains unclear. Foot prints have been suggested to only indicate the presence or absence of ground contact (Hamill *et al.*, 1989) and this cannot be linked sensitively enough to either structure or function. A further limitation of these methods is their inability to detect extremes of foot type, particularly very low arches (Razeghi and Batt, 2002). Radiographic approaches allow for accurate measurement of skeletal landmarks under static weight bearing conditions (Simkin *et al.*, 1989; Saltzman *et al.*, 1995). Sagittal plane radiographic measures include calcaneal inclination angle (Simkin *et al.*, 1989), arch height to length ratio (Simkin *et al.*, 1989) and calcaneal-first metatarsal angle (Smith *et al.*, 1986). Transverse plane foot orientation has been quantified from radiographs using rearfoot-forefoot angle (Freychat *et al.*, 1996). Video fluoroscopy offers a dynamic means of radiographic assessment (Wearing *et a*l., 1998). While both methods offer high reliability (Razeghi and Batt, 2002), they are expensive, expose subjects to radiation and are limited to two dimensions.

While various approaches to static foot type classification exist, much controversy arises when deciding whether to measure in a loaded or unloaded state (Williams and McClay, 2000). Nigg et al. (1998) suggested a new method for quantifying arch deformation by measuring the arch structure in an unloaded and loaded state with results then being normalised to body weight. Further, Williams and McClay (2000) compared arch measurements at 10% and 90% of weight bearing. It was found that dorsum height at 50% of foot length divided by truncated foot length was the best measure of arch height across weight bearing conditions. Measures such as this can provide a measure of arch deformation and therefore flexibility, but measurements remain in two purely static postures. Quantifying arch deformation in dynamic situations such as gait is essential to better understand foot structure and its dynamic behaviours.

When analysing the foot, numerous assumptions, relating to dynamic foot function are based on static measurements of the foot (Razeghi and Batt, 2002). However, the ability of static structure to predict dynamic function has been questioned (McPoil and Cornwall, 1994; Cavanagh *et al.*, 1997). Cavanagh et al. (1997) suggested that static structure cannot be used to infer relative motion within the foot during dynamic situations. These authors found that only 35% of the variance in dynamic plantar pressure could be explained by radiographic measures of the foot. Further, Nachbauer and Nigg (1992) found no relationship between standing arch height and arch flattening during running. Despite this, static measures represent an often simple way of classifying the foot and have been used when investigating the relationship between foot type and injury.

2.1.5.2 The foot as a risk factor for injury

Foot type is a suggested intrinsic risk factor for lower extremity injury (Bennell and Brukner, 2005). To understand how foot type may serve as a potential risk factor, it is necessary to examine literature analysing high and low-arched populations. Both high and low-arched individuals have been shown to display differing patterns of injury, in both type and location (Giladi *et al.,* 1985; Simkin *et al.,* 1989; Williams *et al.,* 2001). Giladi and colleagues (1985) found low-arched subjects to be less likely to develop lower extremity stress fractures. Further, Simkin et al. (1989) found high-arched individuals to exhibit a greater incidence of long bone stress injuries, including tibial and femoral fractures. There is also evidence to suggest high-arched individuals are at an increased risk of bony shock-related injuries, whereas low-arched runners are more prone to soft tissue injuries (Williams *et al.*, 2001).

It has been suggested previously that high-arched feet tend to be more rigid than low-arched feet, resulting in reduced shock absorption capacity (Matheson *et al.*, 1987; Simkin *et al.*, 1989). Despite this, Nachbauer and Nigg, (1992) found similar ground reaction parameters in high and low-arched runners. More recent studies have reported significantly greater loading rates in high-arched runners (Williams *et al.,* 2001b; Williams *et al.,* 2004). Further, higher tibial shock values have been reported in high-arched compared to low-arched individuals (Butler et al., 2006). These data suggest high-arched feet may have a reduced shock absorption capacity. These findings may be related to the degree of subtalar pronation permitted during gait. The lower orientation of the subtalar joint in low-arched individuals permits a greater degree of subtalar pronation than those with high arches (Williams *et al.*, 2006). During stance, significantly less rearfoot eversion has been reported in high-arched compared to low-arched runners (2.9º Butler *et al.*, 2006 and 2º Williams *et al.*, 2004). Importantly, limited rearfoot eversion excursion has been demonstrated to increase the magnitude of impact loading experienced during gait (Perry and Lafortune, 1995). Current evidence suggests that dynamic kinematic differences appear to be present between foot types. Further, it seems that different foot types exhibit difference injury profiles both in terms of injury type and location. However, at present the specific relationship between foot type and tibial stress injuries remains unexplored.

2.1.6 Modelling of the foot

Traditional biomechanical analysis of the foot used two dimensional techniques, foot motion was often described in the sagittal plane in terms of dorsiflexion and plantarflexion around the ankle joint. The foot was assumed to act as a single rigid segment defined by a line through its long axis (Davis *et al.,* 1991). However, such a simplistic model is inadequate for analysing the complex three dimensional rotations occurring within the foot. The foot has been suggested to act as a number of functional units (Wolf *et al.,* 2008). A functional unit is a term which may be used to describe movements at joints of the foot acting together during stance (Wolf *et al.,* 2008). An understanding of these functional units and how they interact is essential for our understanding of dynamic foot function and potential injury mechanisms associated with them. Of particular interest has been the study of the rearfoot. Success has been achieved in the three dimensional tracking of the rearfoot in both barefoot and various footwear conditions during gait (Stacoff *et al.*, 2000; Laughton *et al.*, 2003; Pohl *et al.*, 2006; Eslami *et al.*, 2007;). This has permitted research attempting to understand rearfoot running mechanics and identify potential injury mechanisms. Although rearfoot motion has remained the focus of much research, other functional units of the foot play an important role in gait but at present are far less understood (Nester, 2009).

An ideal experimental scenario would allow direct measurement of the kinematics of all the bones within the foot (Nester, 2009). To date, our understanding of foot mechanics is largely based on the data of either cadaver models or invasive *in vivo* human research (Nester, 2009). However, these approaches are not without their limitations. Cadaver models allow for access to all tissues within the foot, including bones difficult to reach through more superficial means of analysis. However, the main disadvantage of this approach concerns replicating the motions, loads and tissue behaviour seen during *in vivo* study (Nester *et al.*, 2007). This is particularly challenging in the soft tissue structures of muscles and tendons which require active contraction to simulate normal function. Invasive *in vivo* studies typically involve the short term surgical insertion of intracortical bone pins in selected bones of the foot (Lundgren *et al.*, 2008; Ardent *et al.*, 2007; Wolf *et al.*, 2008). However, only a limited number of the bones of the foot can be safely accessed. Further issues include ethical concerns regarding the invasive nature of the analysis and the possibility of altered gait kinematics as a result of the insertions (Nester, 2009). Studies are typically conducted on very low subject numbers, making it difficult to generalise to the wider population. Despite these shortcomings, these methods provide the most accurate means of describing dynamic foot function available currently.

2.1.6.1 Foot modelling challenges

Significant motions in numerous joints of the foot have been reported during gait (Ardent et al., 2007; Lundgren et al., 2008; Wolf *et al.*, 2008). Using *in vivo* methods, four functional units of the foot have been identified during gait: calcaneus, navicular-cuboid, medial cuneiform-first metatarsal, fifth metatarsal (Wolf *et al.*, 2008) (Figure 5). Given this evidence, it appears logical to model the foot as multiple segments during routine biomechanical analysis. However, there are several inherent problems which make this task challenging. The joint motions within the foot are very small compared to other joints of the body, thus presenting problems in tracking segment motion using external markers. Furthermore, soft tissue movement during dynamic activities makes tracking small segments using external markers a problem. While it is generally accepted that the foot can be modelled successfully as several smaller segments, some simplification of foot structure must occur (Davis, 2004). The past decade has seen numerous multisegment foot models proposed in the literature. Several researchers have described three segment models, comprising the rearfoot, forefoot and hallux segments (Carson *et al.,* 2001; Hunt *et al.,* 2001; Myers *et al.,* 2004; Woodburn *et al.,* 2004). Others have attempted to model more segments, proposing five (Leardini *et al.,* 1999), and even nine segment models (MacWilliams *et al.,* 2003).

A major problem encountered when modelling the foot is skin marker movement, and whether external markers can accurately track the motions of underlying bones (Maslen and Ackland 1994; Cappozzo *et al.,* 1996; Tranberg and Karlsson, 1998). All recently developed models, with the exception of one study (Leardini *et al.,* 1999) have been based on skin mounted markers. Leardini et al. (1999) used metallic clamps and adhesive tape to attach lightweight marker clusters to the skin. While this method prevents relative marker movement within each segment, this method is still skin mounted and absolute marker movement remains an issue.

Errors due to skin marker movement on the foot have been reported to range between 1.5mm and 4mm when assessed using radiographic methodology (Maslen and Ackland 1994; Tranberg and Karlsson 1998). The smallest deviations were observed in forefoot markers placed on the heads of the first and fifth metatarsals. Greater marker movement error was seen in rearfoot markers placed close to the subtalar joint. However, these studies only measured two dimensional errors and did not include dynamic measures during gait. Nester et al. (2007) compared foot kinematics during walking when measured using bone mounted markers, external markers and plate mounted markers. These authors reported minimal differences between external markers and plate mounted marker conditions, with larger differences reported when both were compared to bone markers. The greatest errors in skin measurement were reported for the navicular relative to the calcaneus and the navicular relative to the first metatarsal (Nester *et al.*, 2007). Motions at these joints are closely associated with lowering of the medial longitudinal arch (Sammarco, 2004). For the calcaneonavicular joint, mean differences between skin and bone markers were found to be 2.8º, 3.8º and 5.1º for the sagittal, frontal and transverse plane motions respectively. Further, similar differences were seen for the navicular/first metatarsal joint (3.5º sagittal, 3.9º frontal, 3.8º transverse planes) (Nester et al., 2007). It should be noted that skin markers showed no systematic pattern of over or under estimation in the motion of any joint.

Multisegment foot models have primarily been developed for clinical gait analysis purposes, and as a result have been limited to barefoot walking applications. The availability of such multisegment models has seen them used to identify subjects with gross pathology (Woodburn *et al.*, 2004; Rattanaprasert *et al.*, 1999) and adapted for paediatric populations (Stebbins *et al.,* 2006). Recent studies have extended their use to the analysis of barefoot running (Pohl *et al.,* 2006, 2007; Pohl and Buckley, 2008). During a highly dynamic activity such as running, greater skin marker movement error may be expected compared to walking. However, *in vivo* analysis of the rearfoot found only marginal differences in marker movement error between running and walking (Reinschmidt *et al.*, 1997). Sagittal plane errors were only 1.6°, and frontal plane errors only 1.2° greater in running compared to walking. At present, no published data compares error differences between walking and running in other areas of the foot. Similar differences in error may not be seen for the more distal segments of the foot. Further studies are needed to quantify skin marker movement error in the mid and forefoot during running.

Barefoot analysis offers easy access to all areas of the foot for marker application. However, in the context of running injury mechanics, barefoot analysis lacks ecological validity. The study of foot motion when shod presents a unique set of challenges associated with tracking segment motions within the shoe. Markers can be placed on the heel counter of shoe, with their motion assumed to reflect that of the rearfoot within the shoe. However, differences in peak eversion of $2\text{-}4^{\circ}$ (Stacoff *et al.,* 1992 Van Gheluwe *et al.,* 1995) between skin and shoe markers have been reported. Further, when compared with bone mounted markers, Stacoff et al. (2001) found shoe markers to overestimate rearfoot eversion by almost 100%. A more recent approach for routine analysis has been to apply markers directly to the skin through specially cut windows in the footwear (Laughton *et al.*, 2003). Problems with this method include the necessary adaptation to footwear and the potential to compromise the integrity of the footwear by making such adaptations.

Problems persist with the application of forefoot markers directly to the skin during shod gait. These problems have led to demand for alternative methods of accurately tracking the motions of all segments of the foot when shod. Gait sandals have been used as an alternative to shoes in an attempt to accurately capture these motions (Branthwaite *et al.*, 2004, Eslami *et al.*, 2007; Morio *et al.*, 2009). Sandals allow for the easy application and tracking of skin markers, offering a potentially effective means of measuring forefoot and rearfoot motion during routine analysis. While they offer little in the way of motion control, they offer a sole construction likened to running shoes, with similar outsoles and midsoles. Despite this, one study reported no difference in rearfoot eversion and tibial rotation between barefoot and sandal conditions (Eslami *et al.*, 2007). More recent data suggest sandals restrict the natural forefoot frontal and transverse motions associated with barefoot running (Morio *et al.*, 2009). To date, no studies have compared kinematics in sandals and shod conditions. Despite this, sandals may offer a potentially effective means of measuring shod forefoot and rearfoot motion during routine analysis.

In summary, it is now widely acknowledged that the foot can successfully be modelled as more than one segment. However, problems with this persist. The small degree of joint motion coupled with soft tissue movement presents challenges with tracking small segments using external markers. A further issue relates to tracking motions of the foot when shod, gait sandals have recently been used as a way of allowing the direct application of external markers. The recent advances in this area of gait biomechanics has allowed foot mechanics to be investigated in relation to injury.

2.1.6.2 Forefoot-rearfoot mobility

Significant motion at the ankle joint during gait has been acknowledged to extend beyond sagittal plane movement (Nester, 2009). Frontal and transverse motions are also sizeable and have been reported to be 8.1º and 7.9º respectively during walking (Lundgren *et al.*, 2008), increasing to 12.2º and 8.7º during slow running (Ardent *et al.*, 2007). Significant freedom of movement about the joints of the midfoot is also permitted, particularly those of the talonavicular and calcaneocuboid joints (Nester, 2009). At the talonavicular joint, Lundgren et al. (2008) reported 8.4º, 14.9º and 16.3º respectively of sagittal, frontal and transverse plane motion. Similar ranges of motion were seen during running, with the exception of the transverse plane for which sizeably less motion was reported (8.7º) (Ardent *et al.*, 2007). The calcaneocuboid joint has also been found to demonstrate significant freedom during gait. Sagittal, frontal and transverse motions were found to be 9.7º, 11.3º and 8.1º respectively during walking (Lundgren *et al.*, 2008) and 7.8º, 6.3º and 6.9º respectively during running (Ardent *et al.*, 2007). This considerable freedom about the talonavicular and calcaneocuboid joints allows for relative motion between the rearfoot and forefoot (Bojsen-Møller, 1979). Further mobility in the midfoot has also been reported between the navicular and cuboid bones (Lundgren *et al.*, 2008).

With regards to the forefoot, the first three metatarsals appear to function as a stable unit, with the first ray demonstrating a degree of motion with respect to the medial cuneiform (5.3º, 5.4º and 6.1º respectively in the sagittal, frontal and transverse planes) (Lundgren *et al.*, 2008). However, these values are less than those reported between the fifth metatarsal and the cuboid during both walking (13.3º, 10.4º and 9.8º in the sagittal, frontal and transverse planes respectively) (Lundgren *et al.*, 2008) and running (11.4º, 5.1º, 9.6º in the sagittal, frontal and transverse planes respectively) (Ardent *et al.*, 2007). These data demonstrate considerable mobility on the lateral aspect of the foot during gait. Therefore, while the medial longitudinal arch and its collapse during loading is frequently considered, due consideration should also be given to flattening of the lateral arch (Nester, 2009).

Interestingly, some important observations arise from the close scrutiny of these two *in vivo* studies analysing foot biomechanics during gait (Lundgren *et al.*, 2008; Ardent *et al.*, 2007). Firstly, it was noted that high subject variability was present for foot kinematics during gait. It is known that foot structure varies widely between individuals (Razeghi and Batt, 2002), these data suggest this is also true for foot function during gait. Secondly, in the main, range of motion in the joints of the foot during slow running (Ardent *et al.*, 2007) was found to be smaller than when walking (Lundgren *et al.*, 2008). This suggests a potentially stiffer foot when running, a factor which may be controlled through greater use of intrinsic muscles (Nester, 2009). This evidence points to significant and complex movements within the foot during loading.

In summary, the complex structure of the foot and the interaction between bone and soft tissue appear to permit significant motions in its joints when loaded. A large degree of this motion appears to occur at the joints of the midfoot in all three planes. Successfully measuring these motions represents an important step in understanding how functional foot mechanics relate to injury.

2.1.6.3 Forefoot-rearfoot motion as a risk factor for tibial stress injuries

To date, studies investigating the foot as a risk factor for tibial stress injuries have been limited to rearfoot parameters. Peak rearfoot eversion has been found to distinguish female tibial stress fracture sufferers from controls (Pohl *et al.*, 2008; Milner *et al.*, 2010). These publications, from the same larger study, both report a 2.7º greater peak rearfoot eversion in the tibial stress fracture group compared to controls. These findings provide some evidence to suggest peak rearfoot eversion is a risk factor for the development of tibial stress fractures.

Several theories have been proposed to explain the tibial stress injury risk associated with greater rearfoot eversion. The orientation of the subtalar joint in the frontal plane and its function as a mitered hinge means that rearfoot eversion is transferred into internal rotation of the tibia (Inman et al., 1981). In support of this theory, a strong coupling relationship has been observed between rearfoot eversion and shank internal rotation (Pohl *et al.*, 2006, 2007; Eslami *et al.*, 2007). Therefore, excessive rearfoot motion may result in increased rotation of the shank and potentially altered loading on the tibia. Greater rearfoot eversion may increase torsional load being applied to the tibia during stance (Brukner and Bennell, 2005) Increased free moment has been reported when running in shoes modified to increase rearfoot eversion (Holden and Cavanagh, 1991). Furthermore, this theory is supported by the finding of increased free moment in those with a history of tibial stress fracture (Milner *et al.*, 2006a; Pohl *et al.*, 2008).

A further injury theory that relates greater peak eversion may be increased fatigue of the muscles responsible for rearfoot motion (Brukner and Bennell, 2005). Tibialis anterior has been reported to play a large role on controlling rearfoot motion during gait (Cornwall and McPoil, 1994). This muscle acts eccentrically during loading to control rearfoot eversion. Greater rearfoot eversion would cause the muscle to have to work harder to control excessive amounts of joint motion. This localised muscular fatigue may result in a reduced ability of the rearfoot to attenuate the shock associated with impact. In support of this theory, less tibial shock has been found to be attenuated when running in a fatigued state (Mercer *et al.*, 2003).

A further potential mechanism relates to the phase angle of tibial loading and its relationship to bone fatigue. During loading, the tibia is exposed to numerous loads including axial and torsional loads (Ekenman *et al.*, 1998). The phase angle between axial and torsional loading has been reported to have a significant impact on the fatigue life of bone (George and Vashishth, 2005b). These authors found that in-phase loading of axial and torsional loads caused a dramatic increase in bone fatigue. Significantly more bone fatigue was observed when axial and torsional loading were synchronised (in phase). The phase angle seen during gait is related to the kinematics of the loading response, with changes in kinematics altering the phase angle. Greater eversion at the rearfoot is one kinematic measure that is known to increase torsional loads and alter the phase angle, which may represent a potential injury mechanism associated with rearfoot motion. Given this evidence, excessive rearfoot eversion may be an important risk factor for tibial stress fracture development through one or more of these described mechanisms. While traditional research has focussed on the rearfoot and its coupling to shank motion (Deleo *et al.*, 2004), recent research has included more distal segments of the foot to help understand the coupling between the foot and lower extremity (Pohl *et al.*, 2006, 2007; Eslami *et al.*, 2007; Pohl and Buckley, 2008).

As previously discussed, significant motions have been reported in the joints of the midfoot. Modelling the foot as just two segments (forefoot and rearfoot) represents a much simplified foot structure. However, the measurement of forefoot-rearfoot kinematics attempts to quantify the motions associated with the joints of the midfoot; in particular, those of the talonavicular and calcaneocuboid joints. Forefoot motion has been found to be coupled with motion of the rearfoot through the joints of the midfoot (Cornwall and McPoil, 2002; Pohl *et al.*, 2006). Motion of the foot segments during stance may be important in determining unique loading patterns associated with specific injuries.

The foot has been suggested to act as a twisted plate model, which produces counter rotations of the forefoot with respect to the rearfoot during loading (Sarrafian, 1987). Therefore, rearfoot eversion seen during loading would be accompanied by inversion of the forefoot. Evidence to support this notion has been reported during walking, with forefoot inversion found to be coupled with rearfoot eversion (Cornwall and McPoil, 2002). These authors reported counter rotations between the motions occurring between the calcaneus and navicular and those between the navicular and the first metatarsal. In contrast, the forefoot has been reported to evert with respect to the rearfoot during the loading phase of running (Pohl *et al.,* 2006). However, it should be noted that this study employed a two segment model (forefoot-rearfoot) and failed to distinguish motion at specific joints. Rearfoot eversion is also accompanied by dorsiflexion and abduction of the forefoot. A correlation between rearfoot and forefoot motion has showed that rearfoot in/eversion is highly correlated to both forefoot plantar/dorsiflexion and add/abduction (Pohl *et al.,* 2006). However to date, forefoot-rearfoot kinematics during gait have not previously been investigated in those with a history of tibial stress injury.

Previous research has suggested that motions of the forefoot and rearfoot are linked through the joints of the midfoot. Given the strong coupling of the rearfoot and shank, it is suggested that motions of the forefoot may have some influence on the transfer of movement between the rearfoot and shank. Given the reported association between rearfoot eversion and tibial stress fracture risk (Pohl *et al.*, 2008; Milner *et al.*, 2010), it appears pertinent to include related motions of the forefoot and their potential influence of more proximal segments.

In summary, greater peak rearfoot eversion has been identified as an important risk factor for the development of tibial stress injuries. However, the mechanisms through which this risk factor and injury are related, remain speculative. Given the strong coupling of the rearfoot with the shank and the rearfoot with the forefoot, the relationship between these segments in relation to injury is worthy of future study.

2.2 Association between foot type and tibial stress injuries: a systematic review

Work from this section has previously been published as follows:

Barnes, A., Wheat, J., and Milner, C.E. (2008). Association between foot type and tibial stress injuries: a systematic review. *British Journal of Sports Medicine,* **42**, 93-98.

2.2.1 Introduction

As already outlined (Section 2.1.5.1), methods of foot type classification vary greatly with limitations associated with all approaches. Furthermore, the current literature regarding foot type and injury is somewhat contradictory, with different foot characteristics having been considered as potential risk factors associated with lower extremity injury (Cowan *et al.*, 1993; Giladi *et al.*, 1985; Sullivan *et al.*, 1984). The present review was concerned only with foot type as a risk factor for developing a tibial stress injury. Different types and locations of injury likely have different injury mechanisms and, therefore, risk factors associated with them. In an attempt to reduce the incidence of tibial stress injuries, it is important to identify definitively those risk factors specific to this injury mechanism. This is a critical step in the development of preventative measures to help reduce the incidence of tibial stress injuries amongst high risk populations.

Therefore, the aims of the present review were threefold. The primary aim was to determine whether foot type and foot structural characteristics are risk factors for developing tibial stress injuries, by conducting a systematic review of the available literature. Secondly, it sought to provide an assessment of the quality of the current research in this area. Thirdly, the study undertook to highlight specific areas in which further research is needed.

2.2.2 Methods

2.2.2.1 Searching

A search of the following electronic databases was used to identify relevant papers for inclusion in the review: Amed 1985-2011, Cinahl 1982-2011, Index to UK theses, Medline (SilverPlatter) 1950-2011, Pubmed 1966-2011, Scopus 1966-2011, Sports discus 1975-2011, Web of science 1970-2011. The reference lists of review articles were also searched by hand for relevant articles. The search only included articles available in the English language. The following search terms were used: stress fracture, stress injuries, overuse injuries, running injuries, impact injuries, arch height, medial longitudinal arch, high arch, low arch, foot arch, *pes cavus*, *pes planus*, anatomical factors, etiological factors, foot type, foot structure and lower extremity alignment. An example of the search strategy used in Medline is outlined in Table 1; similar strategies were used when searching other databases.

Table 1. Example of search strategy used in Medline (SilverPlatter).

Search Strategy	
1.	Arch adj height
2.	Medial adj longitudinal adj arch
3.	High adj arch*
4	Low adj arch*
5.	Foot adj arch*
6.	Pes adj cavus
7.	Pes adj planus
8.	Anatomical adj factor*
9.	Etiological adj factor*
10.	Foot adj type*
11.	Foot adj structure
12.	Lower adj extremity adj alignment
13.	1-12
14.	Stress adj fracture*
15	Stress adj injur*
16	Tibial adj stress adj fracture*
17.	Overuse adj injur*
18.	Running adj injur*
19.	Impact adj injur*
20.	14-19
21.	13 and 20

adj: Limits searches to adjacent terms; *: Explodes terms

Table 2. Criteria on which studies were assessed for Inclusion. The full scoring system can be seen in Appendix A.

Inclusion Criteria	Scoring
(1). Were the inclusion/exclusion criteria clearly defined?	1-3
(2). Were there sufficient subject numbers included?	1-3
(3). What was the quality of the research design used?	1-3
(4). Comparability of injury and control group?	1-3
(5). What methods/measures were used to classify foot type?	1-3
(6). Were appropriate statistical methods used?	1-3

2.2.2.2 Study Inclusion

Based on the title and abstract, the first reviewer (Primary researcher) identified potentially relevant articles and the full papers were retrieved for further review. The first reviewer excluded irrelevant studies after an initial screening of the full text. In the case of lack of clarity, studies were advanced to the next stage of screening for further examination

The remaining studies were assessed independently by two reviewers (Primary researcher and supervisor) and scored based on six separate inclusion criteria (Table 2). The quality assessment scoring system can be seen in Appendix A. The appraisal tool used was developed for this study, given that validated assessment tools for studies which are not randomised controlled trials do not exist. The criteria were based on those within existing appraisal tools (Crombie, 1996; Ellwood, 2000), as well as key criteria identified as being specific to this review. The appraisal key was based on previous keys developed by The Cochrane Collaboration Injuries Group. The maximum inclusion assessment score available was 18, with three representing the maximum, and one the minimum score for each question. A scoring system of 1-3 was used for all six questions so as not to weight the scoring towards any one inclusion criteria. If disagreements concerning the scoring of studies occurred, discussion was used as a resolution tool. The score of each study was converted to a percentage and the quality system of McKay employed, whereby a score of 0-49% was classed as poor, 50-89% moderate and >90% good (Mackay *et al.*, 2001). Studies of 50% or above were deemed of high enough quality for inclusion in this review. Studies which did not define tibial stress fractures or stress reactions as specific injuries were not included. Further, care was taken not to include studies which grouped tibial stress injuries with other injuries in the same region. These included shin splints that were not clearly defined, medial tibial stress syndrome and compartment syndrome.

2.2.2.3 Data Extraction and Appraisal

Data were extracted using a custom designed data extraction form (Appendix B). The forms were piloted on a sub sample of the studies and adapted accordingly before standardised data extraction was completed. The form included details of study design, inclusion criteria, participants, and aspects of methodology as well as the study results. Given that none of the included studies were randomised controlled trials and differed in population and statistical procedures, it was considered inappropriate to carry out a statistical meta-analysis. Further, given the considerable methodological variations between studies, it was felt a meta-analysis would be unable to correct for these confounding factors. Instead, a descriptive account of studies was formulated to characterise the research and identify potential strengths and weaknesses in the literature.

2.2.3 Results

Searches in all databases identified 479 unique studies. Based on title and abstract, 57 of these were identified as potentially relevant and their full texts retrieved. After an initial review by the first reviewer, 32 of these were determined to fall outside the parameters of this review and were excluded. The remaining 25 studies were assessed against the inclusion criteria by both reviewers independently. Of the 25 articles assessed, nine studies achieved inclusion scores of greater than 50% and were therefore included in this review, details of which are presented in Table 3. Of these nine studies, all were found to be of moderate quality (50-90%), with scores ranging from 56% (Ekenman *et al.*, 1996; Taunton *et al.*, 2002) to 78% (Kaufman *et al.*, 1999).

Of the nine studies, six involved sporting populations (Matheson *et al.*, 1987; Ekenman *et al.*, 1996; Taunton *et al.*, 2002; Busseuil *et al.*, 1998; Korpelanien *et al.*, 2001; Williams *et al.*, 2001a), and three involved military recruits (Montgomery *et al.*, 1989; Simkin *et al.*, 1989; Kaufman *et al.*, 1999). The three military studies were limited to male participants, whilst the athlete studies had cohorts comprising both male and female participants. The three military investigations were all prospective studies, with follow up periods ranging from 3.5 (Simkin *et al.*, 1989) to 24 months (Kaufman *et al.*, 1999). The six athlete studies were retrospective designs (Matheson *et al.*, 1987; Ekenman *et al.*, 1996; Taunton *et al.*, 2002; Busseuil *et al.*, 1998; Korpelanien *et al.*, 2001; Williams *et al.*, 2001a). The number of tibial stress injuries reported ranged from 6 (Williams *et al.*, 2001a) to 157 (Matheson *et al.*, 1987). It should be noted that study participants in all studies were injury-free at the time of participation.

2.2.3.1 Injury Diagnosis

Clinical examination by a medical professional, coupled with injury questionnaires were used to diagnose stress fractures in two of the included studies (Montgomery *et al.*, 1989; Williams *et al.*, 2001a). However, Montgomery and colleagues (1989) validated these methods with confirmation of a positive fracture on a sample of injured athletes using criterion methods. All other studies used the presentation of clinical symptoms confirmed by imaging technologies to diagnose injury (Matheson *et al.*, 1987; Simkin *et al.*, 1989; Ekenman *et al.*, 1996; Kaufman *et al.*, 1999; Korpelanien *et al.*, 2001). One study used triple bone scans (Matheson *et al.*, 1987), whilst another used either nuclear bone scan or radiography to confirm the presence of tibial stress injuries (Kaufman *et al.*, 1999). Three further studies used a combination of radiography and bone scintigrams as diagnostic tools (Simkin *et al.*, 1989; Ekenman *et al.*, 1996; Korpelanien *et al.*, 2001). The two remaining studies (Taunton *et al.*, 2002; Busseuil *et al.*, 1999), did not state explicitly the methods used to diagnose injury, however, the use of "appropriate imaging methods" was reported in one of these articles (Taunton *et al.*, 2002).

2.2.3.2 Foot Type Classification

There are numerous methods for classifying foot type and this can prove problematic for comparison purposes with wide variation in both the methods used to classify foot type and the way in which the methods were reported. The classification of foot type across studies ranged from subjective determination to more detailed anatomical measurements. Three of the included trials assessed foot type subjectively through visual inspection of the participants (Matheson *et al.*, 1987; Montgomery *et al.*, 1989; Taunton *et al.*, 2002). In two of these, feet were classified as *pes cavus*, normal or *pes planus* (Matheson *et al.*, 1987; Montgomery *et al.*, 1989), whilst in the third arch height was grouped as low, normal or high (Taunton *et al.*, 2002). Another study (Korpelanien *et al.*, 2001) obtained foot prints using a podoscopic mirrored table and classified feet as *pes* *cavus*, normal or *pes planus* based on these observations. Static arch height based on anthropometric measures was used in three of the reviewed articles (Simkin *et al.*, 1989; Kaufman *et al.*, 1999; Williams *et al.*, 2001a). Simkin et al. (1989) measured calcaneal angle based on lateral radiographs of the foot. In other studies, external measures of the feet, specifically navicular height (Kaufman *et al.*, 1999) and dorsum height (Williams *et al.*, 2001a) divided by foot length were both used as quantitative measures of arch index. Three further studies used foot pressure analysis to provide a measure of foot type (Ekenman *et al.*, 1996; Kaufman *et al.*, 1999; Busseuil *et al.*, 1999). Of these, one study (Ekenman *et al.*, 1996) used pressure distribution under the tarsal region of the foot to classify feet, whilst Kaufman et al. (1999) calculated an "arch ratio" defined by midfoot contact area to total foot contact area. A third study (Busseuil *et al.*, 1999) calculated rearfoot to forefoot angle, using this measure to indicate either a pronated or an open foot type. The three studies using pressure analysis were the only ones to include dynamic analysis of the foot, with two studies analysing pressures during walking (Ekenman *et al.*, 1996; Kaufman *et al.*, 1999), and one during running (Busseuil *et al.*, 1999). Furthermore, only one study incorporated dynamic measures whilst walking shod (Kaufman *et al.*, 1999). Assessment of shod gait may be more ecologically valid than the barefoot assessments, although likely highly dependent on footwear type.

Variation between studies was also seen in the methods used to classify foot type for comparison purposes. In the studies in which subjective grouping of foot type was conducted (Matheson *et al.*, 1987; Montgomery *et al.*, 1989;

Taunton *et al.*, 2002; Korpelanien *et al.*, 2001), those classed as having normal feet were used as a reference group for comparison.

After measuring arch parameters, two of the studies subdivided the population arbitrarily in two (high/low arches) (Simkin *et al.,* 1989) and three (high/normal/low) (Kaufman *et al.*, 1999) equal groups. These subgroups provided the basis for comparison, with Kaufman et al. (1999) using the normal arched group as a reference for all comparisons. A further study (Williams *et al.*, 2001a) made direct comparisons of injury incidence between those with very high and low arches. In this study arch height was determined relative to a normative database. This ensured that the arch height was determined relative to the population, not arbitrarily assigned relative to the sample recruited into the study. Regression comparison of measured foot parameters between injury and control groups formed the basis for analysis in the two remaining investigations (Ekenman *et al.*, 1996; Busseuil *et al.*, 1999).

Table 3. Outline of Included Studies

Study	Quality score (%)	Participants	Foot Type definition	Injury Definition	Outcomes
Busseuil *et al.* 1998	67	66 Ath (50M; 16 F) age 34±10 yrs. 15 TSI.M/con (n=216)	ST, DN: Pressure analysis, rearfoot:forefoot angle (α)	NS	Lower static rearfoot:f angles (a more pronated f those with TSI.
Ekenman *et al.* 1996	56	29 Ath with TSF (11M; 18F) age yrs. M/con (n=30)	DN: Pressure analysis, tarsal region during gait.	Radiography or bone scintigrams	Similar occurrence of HA i TSF and M/con (approx 30
Kaufman *et al.* 1999	78	449M Mil. age 22.5±2.5 yrs. F/u 24 mo. 29 TSF	ST: Navicular height:foot length DN: pressure analysis (midfoot:total foot contact area)	Radiography or nuclear bone scan	Increased risk ratios ass with both PC and PP feet.
Korpelainen *et al.* 2001	61	31 Ath (19M; 12F) age 20yrs. 45 TSF. M/con (n=15)	ST: Subjective classification based on footprint analysis (PC/normal/PP)	Radiography or bone scintigrams	TSF 40% HA, control 13%
Matheson *et al.* 1987	61	320 Ath (145M; 175F) age NS.157 TSF	ST: Subjective classification (PC/normal/PP)	Triple bone scan	TSF approx 53% PP, app PC.
Montgomery *et al.* 1989	61	505M Mil, age 22±3 yrs. f/u 6mo. 27 TSF	ST: Subjective classification (PC/normal/PP)	Clinical Examination	20% TSF were PP. 0% T PC.
Simkin *et al.* 1989	67	295M Mil. (age NS) f/u 3.5mo. 286 TSF.	ST: Calcaneal angle measured from lateral radiographs	Radiography or bone scintigrams	TSF incidence of 9.8% in group compared to 17.3% i
Taunton *et al.* 2002	56	67 Ath with TSF (27M; 40F) age±32.3 yrs. M/con not available	ST: Subjective classification (HA/normal/LA)	Appropriate imaging modalities (NS)	PP 11% and PC 7% in tho TSF.
Williams *et al.* 2001a	61	20HA (age 28±8.1 yrs), 20LA (27.7±7.5 yrs) Ath, 18M; 22F. 6 TSF	ST: Arch ratio (dorsum height/truncated foot length)	Clinical Examination, injury history questionnaire	50% more TSF observed high arched group.

Abbreviations: Ath: Athletes; Mil: Military Recruits; M: Males; F: Females; mo: Months; f/u: Follow up period; M/con: Matched control group; TSF: Tibial Stress Fractures; TSI: Tibial Injuries; ST: Static measurements; DN; Dynamic Measurements; HA: High Arched; LA:Low Arched; PC: Pes Cavus; PP: Pes Planus; NS: Not specified.

2.2.3.3 Foot type as a risk factor for the development of tibial stress injuries

Of the nine studies included, two failed to find any association between foot type and tibial stress injuries (Ekenman *et al.*, 1996; Taunton *et al.*, 2002). One of these studies (Taunton *et al.*, 2002) found *pes planus* and *pes cavus* feet to be present in 11% and 7% respectively of subjects with tibial stress fractures. However, foot type distribution could not be compared to a similar uninjured population. A further study (Ekenman *et al.*, 1996) reported a similar incidence of high arched feet in both injury and control groups (approximately 30%).

Four of the studies presented data suggestive of an increased risk of tibial stress injury associated with a more planus or low arched foot (Matheson *et al.*, 1987; Montgomery *et al.*, 1989; Kaufman *et al.*, 1999; Busseuil *et al.*, 1999). Matheson et al. (1987) found that in those with previous tibial stress fracture, considerably more of the population had pronated (approximately 53%) compared to cavus (approximately 2%) feet. Another study reported 20% of those with tibial stress fracture as having planus feet, whilst none were classified as having cavus feet. (Montgomery *et al.*, 1989) Although neither study (Matheson et al., 1987; Montgomery *et al.*, 1989) conducted a statistical analysis, subjective comparisons suggest an association between low arches and tibial stress fracture. Busseuil et al. (1998) found significantly lower static rearfoot to forefoot angles when comparing those with tibial stress injuries to healthy controls. Similar differences were seen for dynamic measures and, although not significant, these results are suggestive of a more pronated foot in those with tibial stress injuries. One study (Kaufman *et al.*, 1999) found evidence which is suggestive of an increased risk of tibial stress fracture associated with both planus and cavus feet. Increased injury risk was reported for both extremes of foot type compared to the normal group. These findings however, were only significant for a *pes planus* foot type in the dynamic shod condition (risk ratio of 2.45).

Three further included studies also present data which suggest a high arched or cavus foot may increase the risk of tibial stress fracture (Simkin *et al.,* 1989; Korpelanien *et al.*, 2001; Williams *et al.*, 2001a). When classifying arch height based on calcaneal angle, Simkin et al. (1989) reported a tibial stress fracture incidence of 9.8% in the low arched group compared to 17.3% in those with high arches. These differences were not found to be significant, but they do suggest an association between high arches and tibial stress fracture. In their study, Williams et al. (2001a) adopted a different study design based on recruiting runners with very high or very low arches. Despite low subject numbers in the study, high arched runners reported twice as many tibial stress fractures as those with low arches. In addition, Korpelainen et al. (2001) found high arches to be more prevalent in those with stress fractures (40%) than in the control group (13%).

2.2.4 Discussion

The aim of the present review was to determine if foot type is a risk factor in developing a tibial stress injury. Based on the nine studies reviewed, there is insufficient evidence to indicate a definitive link between foot type and tibial stress injuries. Limited evidence can be found in support of an increased risk associated with either high or low-arched foot types. The present findings indicate that the measures of foot structure used currently provide at best a limited indication of tibial stress injury risk. This finding lends evidence to the multiplicity of risk factors, particularly external factors, which are likely to relate to the development of tibial stress injuries.

The relationship between arch height and arch flexibility is one which acts on a continuum. However, it has been suggested that high-arched feet tend to be more rigid compared to low-arched feet which are considered more flexible (Subotnick, 1985). A flexible low arch may be better able to absorb the shock associated with impact than a stiffer high arch. Despite this, recent evidence found only a weak relationship between the parameters of arch height and stiffness (Zifchock *et al.*, 2006b). Whilst a higher arch did tend to relate more to a stiffer one, only 9% of the variance in arch height could be attributed to stiffness measures.

Assumptions relating to injury risk are often based on static measurements of the foot. Several of the included studies based their findings on qualitative static foot type (Matheson *et al.*, 1987; Montgomery *et al.*, 1989; Taunton *et al.*, 2002; Korpelanien *et al.*, 2001), whilst two further studies used quantitative measures taken solely in static postures (Simkin *et al.,* 1989; Williams *et al.*, 2001a). In the present review only three studies incorporated dynamic foot type measurements (Ekenman *et al.*, 1996; Busseuil *et al.*, 1999; Kaufman *et al.*, 1999), and only two of these (Busseuil *et al.*, 1999; Kaufman *et al.*, 1999) measured both static and dynamic characteristics. Evidence provided by Kaufman et al. (1999) suggests a greater associated injury risk with dynamic as opposed to static measures of foot type. It has been suggested that static measures are of little use for inferring relative motions within the foot during dynamic situations (Hamill *et al.*, 1989; Hennig and Milani, 1993; Cavanagh *et al.*, 1997). This may account for the lack of consensus regarding foot type and tibial stress injury risk in the literature. Whilst static measures tell us much about the anatomy of different foot types, they offer little information regarding how these foot types function during dynamic activities. Although there are clear advantages to being able to quantify injury risk via simple static measures, interaction between the foot and the environment may be overlooked. A more complex measure, for example one that incorporates both arch height and arch stiffness, may be more strongly related to tibial stress injury risk.

One crucial factor which might account for the conflicting findings of this review is the various methods used to classify foot type. In the present review, four studies used experienced testers to classify foot type in a subjective manner (Matheson *et al.*, 1987; Montgomery *et al.*, 1989; Korpelanien *et al.*, 2001; Taunton *et al.*, 2002). Such static qualitative procedures offer a simple and efficient grouping method, and one that is particularly useful for clinicians (Razeghi and Batt, 2002). However, subjective methods such as these have been shown to introduce a degree of error associated with misclassification

(Dahle *et al.*, 1991; Cowan, 1994). When grouping feet, clinicians must base their judgement on previous experience of a wider population, which may result in significant variation between testers. Dahle and colleagues (1991) reported only a 73% agreement between clinicians when classifying feet into three groups, whilst Cowan et al. (1994) observed high inter-tester variability using a five point grouping scale. Qualitative assignment has the further problem of being potentially skewed towards the grouping of more planus than cavus feet, as they are considered more prevalent within the population. Evidence of this can be found in one included study, which reported more than double the number of feet classified as *pes planus* (11.8%) compared to *pes cavus* (5.4%), in the total population observed (Montgomery *et al.*, 1989). Results suggesting a relationship between foot type and tibial stress injuries based on subjective classification should therefore be treated with caution.

Quantitative foot assessment methods have been shown to offer improved measurement reliability (Hawes *et al.*, 1992). However, the classification of foot type based on these measures is crucial to the validity of study outcomes. Ekenman et al. (1996) used plantar pressure patterns from literature sources to assign foot type into three groups. The finding of approximately 30% cavus and no planus feet in both groups, suggests skewness in either the sample population or the measurement method. Two studies classified foot type by dividing the measured population arbitrarily into groups (Simkin *et al.,* 1989; Kaufman *et al.*, 1999). Such approaches however, may not represent populations beyond that of the study, as classification is not in relation to a wider sample. Williams et al. (2001a) deliberately sampled foot type extremes for comparison based on a larger normative database. This is a strength of the study and a factor which may account for the large observed difference in tibial stress fracture incidence between the groups.

The definition of tibial stress injuries and the methods used to diagnose them are also important. Diagnostic methods were not limited to imaging technologies in the present review but care was taken to exclude other leg injuries such as non-specific shin splints, medial tibial stress syndrome and compartment syndrome. Despite this, evidence of stress reactions cannot be seen on conventional radiographs, with tibial stress fracture only evident at an advanced stage. Therefore, in those studies relying solely on radiological confirmation, it is possible that early signs of a tibial stress fracture could have been overlooked. Further, in military recruits, Montgomery et al. (1989) suggested injury incidence may go underreported due to the high level of motivation to continue training.

Prospective studies are often considered the best study design for determining the aetiology of injuries, as they allow the mechanics of the lower extremities to be studied prior to injury occurring. When assessing research design, the present review awarded prospective studies higher inclusion scores than retrospective designs. In this review, only three of the studies were prospective type designs (Montgomery *et al.*, 1989; Simkin *et al.,* 1989; Kaufman *et al.*, 1999). However, it should be noted that some initial evidence suggests retrospective and prospective studies produce similar results when relating anatomical factors to tibial stress fracture (Hamill and Davis, 2006). Further, more confidence can be placed in retrospective studies if we can assume that the anatomical structure and functional mobility of the foot is not affected by tibial stress injuries. Whilst this may be true for the majority of retrospective designs, in one such study subjects with a history of stress fracture were recruited from up to 23 years previously (Korpelainen *et al.*, 2001). It has been suggested that arches tend to fall with age, and a greater incidence of low-arched feet in older adults has been observed (Funk *et al.*, 1986). After such a long period post fracture, it is likely that foot characteristics may have changed such that the foot structure being measured is not the same as the foot structure when the injury was sustained.

2.2.4.1 Implications for future research

This review has highlighted the need for research regarding intrinsic foot parameters as risk factors for tibial stress injuries. Studies using multi-segment foot models to investigate functional mobility and flexibility within the foot in dynamic situations such as running are essential. Robust quantitative but simple measures of the foot need to be employed as opposed to the more traditional subjective classification methods. In the present review only one investigation studied foot type when shod (Kaufman *et al.*, 1999). Future attention should be given to how the foot functions during dynamic activities and how mobility characteristics interact with external conditions such as footwear. This may enable the development of interventions designed to reduce the risk of tibial stress injuries within high risk populations.

2.2.5 Conclusion

This review adopted a systematic approach in which strict selection criteria were used to assess the literature surrounding foot type and tibial stress injuries. The outcomes of the nine investigations included were difficult to compare due to varying methods. Results proved conflicting with limited evidence implicating any one foot type as a potential risk factor. However, limited evidence was found to suggest that both high and low-arched feet pose an increased risk of tibial stress injuries compared to normal feet. Dynamic measures of foot function may prove to be more useful in predicting the risk of tibial stress injury.

2.3 Summary of Literature Review

The literature review has sought to provide an overview of the current research and issues associated with the programme of study. It has also sought to investigate the specific association between foot type and tibial stress injuries through a systematic review. Consideration has been given to the mechanisms of tibial stress injuries as well as risk factors for their development. Attention has also been given to the role of loading variables and kinematic indicators. The issues surrounding foot type classification and its potential link to overuse injuries are discussed. A more detailed exploration of this relationship is provided in the form of a systematic review. The close scrutiny of the included studies failed to provide sufficient evidence to indicate a definitive link between foot type and tibial stress injuries. Importantly, this review highlighted a need for dynamic measures of foot type to investigate the relationship between foot function and tibial stress injury risk.

Methodological considerations associated with the use of multisegment foot models have been highlighted with a view to developing a model for use in the programme of research. Particular focus has been on the foot and the large degree of mobility highlighted during loading. Forefoot-rearfoot motion has been discussed and the idea of forefoot-rearfoot motions as risk factors for tibial stress injuries introduced. This thesis sought to study forefoot-rearfoot kinematics in specific relation to tibial stress injuries.

CHAPTER III

3 Methodology

3.1 Introduction

The review of previous literature in Chapter II has highlighted the need for research into forefoot and rearfoot kinematics during running. However, to investigate forefoot- rearfoot kinematics in selected populations, methodological considerations need to be addressed; in particular, the model used for analysis. This chapter examines methodological considerations with the measurement of forefoot and rearfoot kinematics. The Chapter is divided into three sections. The first presents a multisegment foot model for future use in answering relevant research questions. The second section presents data to support the use of gait sandals as an effective means of measuring kinematic motions seen during shod gait. The final section addresses the reliability of specific dependent variables, including selected forefoot and rearfoot kinematic parameters as well as tibial shock.

3.2 Multisegment foot model

3.2.1 Introduction

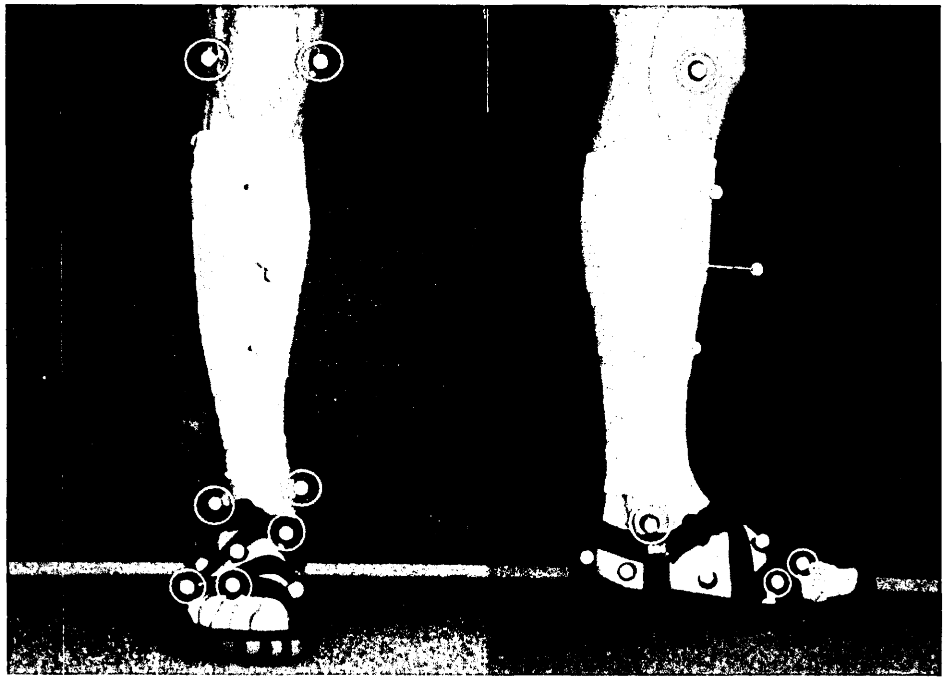
The past decade has seen numerous multisegment foot models proposed in the literature (section 2.1.6). Several researchers have described a multisegment foot which models the rearfoot and forefoot segments separately (Carson *et al.,* 2001; Hunt *et al.,* 2001; Myers *et al.,* 2004; Woodburn *et al.,* 2004). These studies have largely been limited to barefoot walking, although more recent work has seen their application to barefoot running (Pohl *et al.*, 2006, 2007) and running in sandals (Eslami *et al.*, 2007). This thesis sought to develop a three segment (shank, rearfoot, forefoot) multisegment foot model for use with gait sandals during running.

3.2.2 The model

The model developed for this thesis was based on Carson et al. (2001), Digby et al. (2005) and Nester and Findlow (2006), employed rigid body assumptions, and consisted of three segments (shank, rearfoot and forefoot). Sixteen retroreflective markers (Table 4) were placed on the right foot and shank of each participant (Figure 6). The model was developed to be used when wearing gait sandals (Bite Orca, Orthosport, Anatom, Edinburgh, UK). Individual segment coordinate systems were constructed from a static standing trial used to establish the relationship between static and dynamic markers. In accordance with Cappozzo et al. (1995), the long axes of segments were defined first.

Table 4. Names and positions of markers used in the foot model. Markers in bold are used in the static trial only and are removed for dynamic trials. The placement of markers on the lower extremity can be seen in Figure 6.

Marker Name	Position	Segment
MKN	Medial mid femoral condyle	Shank
LKN	Lateral mid femoral condyle	Shank
MMAL	Medial malleolus	Shank
LMAL	Lateral malleolus	Shank
SH1	Marker cluster on anterior crest of tibia	Shank
SH2	Marker cluster on anterior crest of tibia (wand)	Shank
SH3	Marker cluster on anterior crest of tibia	Shank
PCAL	Posterior aspect of the calcaneus	Rearfoot
MCAL	Medial aspect of the calcaneus	Rearfoot
LCAL	Lateral aspect of the calcaneus	Rearfoot
P1MT	Proximal head of the first metatarsal	Forefoot
P5MT	Proximal head of the fifth metatarsal	Forefoot
D1MT	Distal head of the first metatarsal	Forefoot
D2MT	Midpoint of heads of second and third metatarsals	Forefoot
D5MT	Distal head of the fifth metatarsal	Forefoot
DOR	Dorsum of the foot	Forefoot



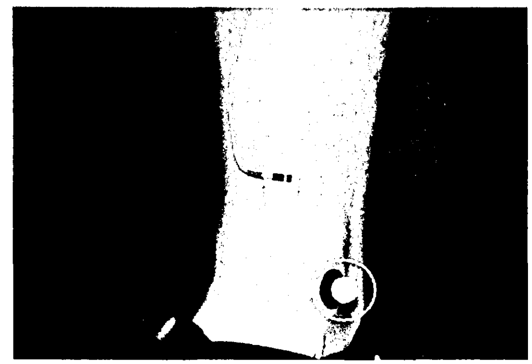


Figure 6. Left - Front and lateral view of all static marker placements. Static markers are circled: these were removed during dynamic trials. See section 3.2.2 for description of marker placement. Right- Accelerometer placement on the anterior medial aspect of the tibia.

3.2.2.1 Shank

The precise location of all markers and the segments they define can be seen in Table 4. The shank was defined as the tibia and fibula, assumed to move as a single rigid segment. MKN and LKN markers were placed on the medial and lateral mid femoral condyles. MMAL and LMAL markers were placed on the most medial and lateral aspects of the malleoli. A three marker cluster with one projecting wand marker was firmly attached to the anterior crest of the tibia, and over wrapped to help minimize skin movement (Digby *et al.*, 2005). After the static trial, MKN, LKN, MMAL and LMAL markers were removed for dynamic trials.

3.2.2.2 Rearfoot

The rearfoot segment comprises the calcaneus and talus bones of the foot, it is assumed to move as a single rigid segment. The PCAL marker was placed on the most posterior aspect of the distal calcaneus, with LCAL and MCAL markers placed on the lateral and medial aspects of the distal calcaneus. Markers were not placed at specific anatomical points but were placed equidistant to the planter surface, with LCAL and MCAL markers the same distance from the PCAL marker (Nester and Findlow, 2006). All three markers were used as tracking markers for the segment and were all present during dynamic trials.

3.2.2.3 Forefoot

The forefoot consisted of five metatarsals assumed to act as a single rigid segment, as defined by Carson et al. (2001). The P5MT marker was placed laterally over the proximal head of the fifth metatarsal, whilst the P1MT was placed on the proximal head of the first metatarsal. D1MT and D5MT were placed medially and laterally on the distal heads of the first and fifth metatarsals. The D2MT marker was placed at the midpoint between the distal heads of the second and third metatarsals. A tracking marker placed just distal to the dorsum of the foot (DOR) in between straps allowed for direct application of the marker to the skin. Markers P1MT, D5MT and D2MT were removed for dynamic trials, leaving P5MT, DOR and D1MT as tracking markers.

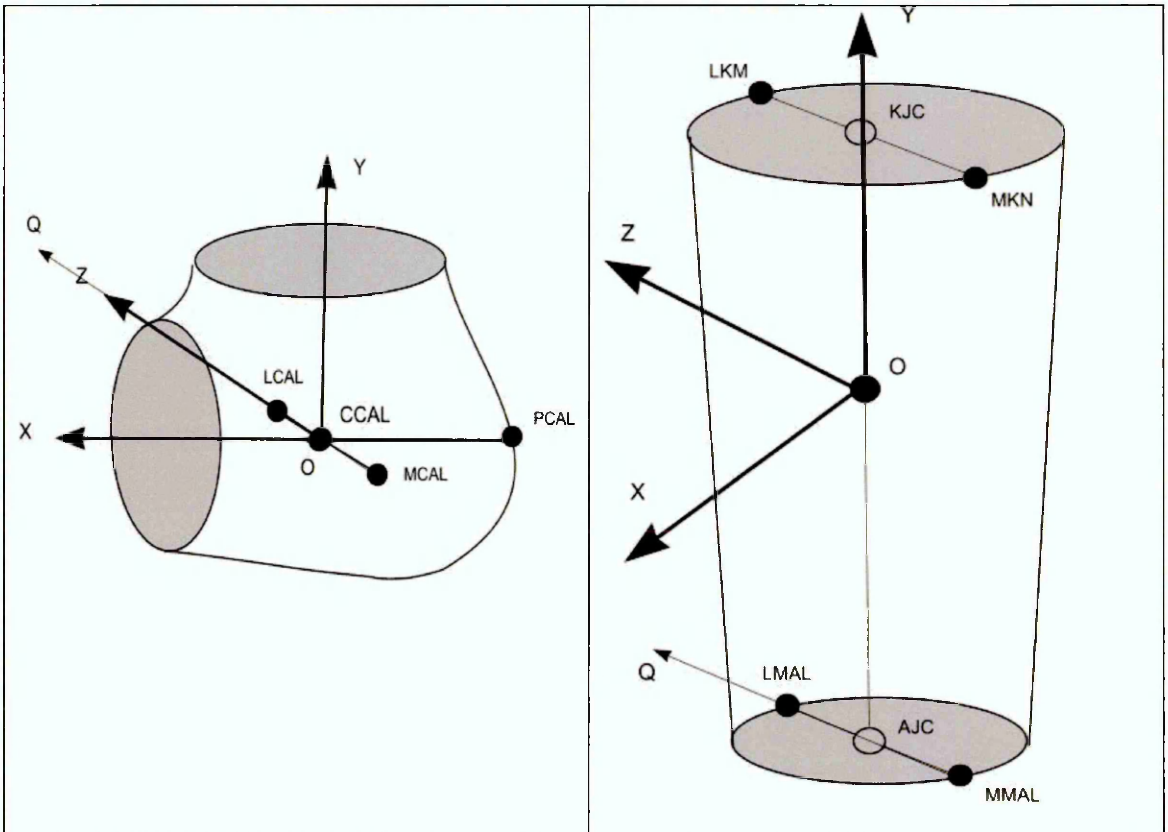
3.2.2.4 Joint Coordinate Systems

The following section outlines details of the segment coordinate systems and subsequently the joint coordinate systems used at the shank, rearfoot and forefoot.

Table 5. Definitions of terms included when defining segment and joint coordinate systems.

Definition of terms	
X, Y, Z	The three orthogonal axes of the segment coordinate system. Orientated approximately anterior-posteriorly, interior-superiorly and medio-laterally respectively (all with the body in the anatomically neutral position).
e_1 e_2 e_3	First, second and third axes of the non-orthogonal joint coordinate system.
MKN	Medial mid femoral condyle
LKN	Lateral mid femoral condyle
KJC	The knee joint centre located midway between MKN and LKN
MMAL	Medial malleolus
LMAL	Lateral malleolus
AJC	The ankle joint centre located midway between MMAL and LMAL
PCAL	Posterior aspect of the calcaneus
MCAL	Medial aspect of the calcaneus
LCAL	Lateral aspect of the calcaneus
CCAL	Calcaneus centre located midway between MCAL and LCAL
P1MT	Proximal head of the first metatarsal
P5MT	Proximal head of the fifth metatarsal
MPMT	Proximal forefoot centre located midway between P1MT and P5MT
D1MT	Distal head of the first metatarsal
D2MT	Midpoint of heads of second and third metatarsals
D5MT	Distal head of the fifth metatarsal
O	Origin of segment coordinate system

The definitions for the rearfoot and shank coordinate systems used in the ankle joint coordinate system are summarised in Figure 7.



O: Origin of segment coordinate system. **Q:** An intermediate vector passing from MCAL to LCAL **X**: X-axis: The line coincident with that passing through PCAL and CCAL, directed anteriorly. **Y:** Y-axis: The cross product of **Q** x **X** **Z:** Z-axis: The cross product of **X** x **Y**	**O:** Origin of segment coordinate system. **Q:** An intermediate vector passing from MM to LM **X**: X-axis: The cross product of **Y** x **Q** **Y:** Y-axis: The line coincident with that passing through AJC and KJC, directed superior. **Z:** Z-axis: The cross product of **X** x **Y**

Figure 7. The definitions of the rearfoot (left) and shank (right) segment coordinate systems used.

Subsequently, the ankle joint coordinate system was defined as follows (Figure 8).

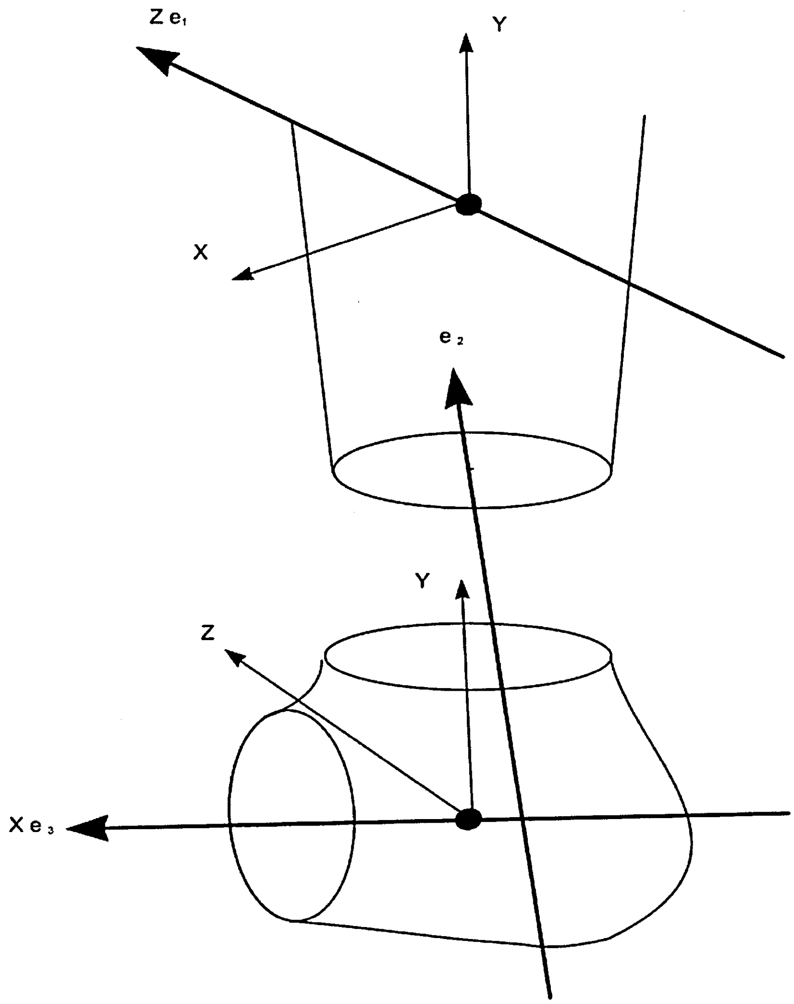


Figure 8. The ankle joint coordinate system.

Where:

e_1 : Z-axis of the shank coordinate system - flexion/extension axis

e_2 : The floating axis, defined as the cross product of the e_1 and e_3 axes - abduction/adduction axis.

e_3 : X-axis of the rearfoot coordinate system - inversion/eversion axis.

The definitions for the forefoot and rearfoot coordinate systems used are summarised in Figure 9.

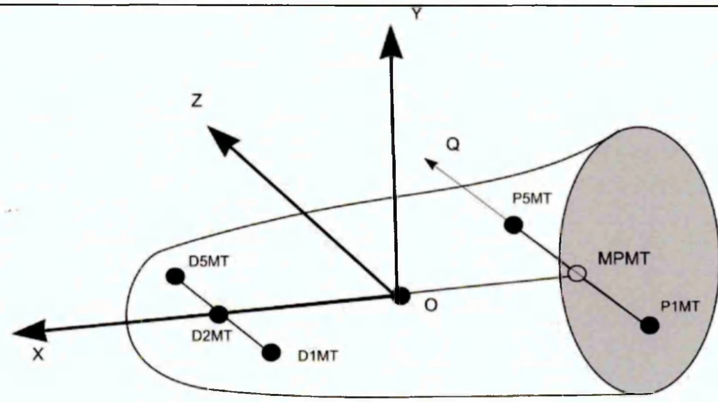
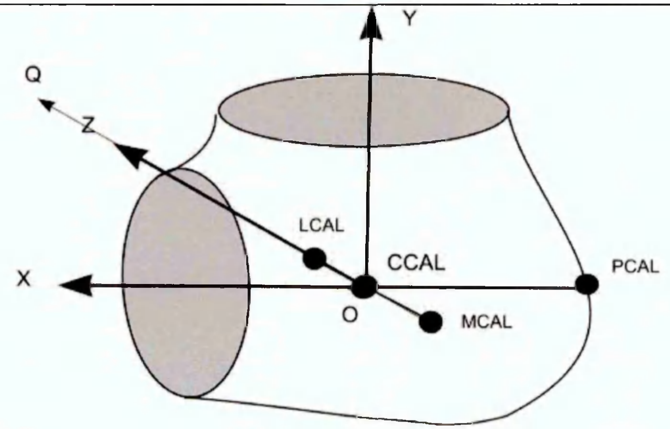
	
O: Origin of segment coordinate system. **Q:** An intermediate vector passing from P1MT to P5MT **X**: X-axis: The line coincident with that passing through MPMT and D2MT, directed anteriorly. **Y:** Y-axis: The cross product of **Q** x **X** **Z:** Z-axis: The cross product of **X** x **Y**	**O:** Origin of segment coordinate system. **Q:** An intermediate vector passing from MCAL to LCAL **X**: X-axis: The line coincident with that passing through PCAL and CCAL, directed anteriorly. **Y:** Y-axis: The cross product of **Q** x **X** **Z:** Z-axis: The cross product of **X** x **Y**

Figure 9. The definitions of the forefoot (left) and rearfoot (right) segment coordinate systems used to calculate the joint coordinate system of the forefoot relative to the rearfoot.

Subsequently, the forefoot to rearfoot joint coordinate system was defined as follows (Figure 10).

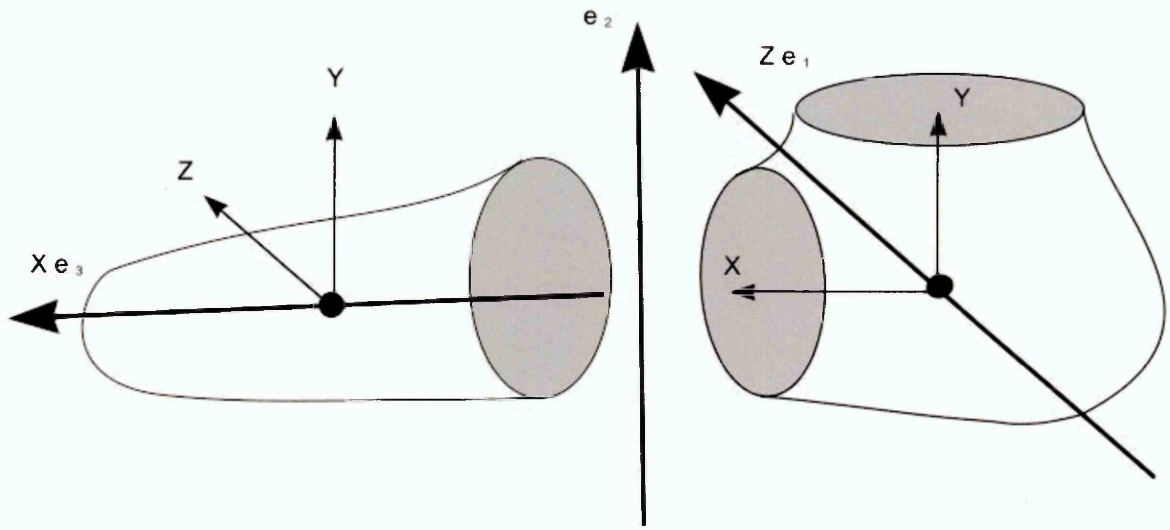


Figure 10. The joint coordinate system for the forefoot relative to the rearfoot.

Where:

e_1 : Z-axis of the rearfoot coordinate system - flexion/extension axis

e_2 : The floating axis, defined as the cross product of the e_1 and e_3 axes - abduction/adduction axis.

e_3 : X-axis of the forefoot coordinate system - inversion/eversion axis.

3.3 The use of gait sandals for measuring rearfoot and shank motion during running

Work from this section has previously been published as follows:

Barnes, A., Wheat, J., and Milner, C.E. (2010). Use of gait sandals for measuring rearfoot and shank motion during running. *Gait and Posture*, **32**, 133-135.

3.3.1 Introduction

Rearfoot and shank kinematics and the coupling of these movements, has been a focus in attempting to understand running mechanics and injury mechanisms (DeLeo *et al.*, 2004). Tracking foot motion within a running shoe presents a unique set of challenges. Markers can be placed on the heel counter of the shoe as a proxy measure of actual rearfoot motion. However, significant differences in rearfoot eversion have been reported between both shoe and skin (Stacoff *et al.*, 1992) and shoe and bone markers (Stacoff *et al.*, 2001). A more recent approach has been to apply markers directly to the skin through windows cut in the footwear (Laughton *et al.*, 2003). Difficulties with this method include the potential to reduce heel counter rigidity by making the necessary adaptations. A further challenge is ensuring that wand markers move freely within the window.

The increased use of multisegment foot models in biomechanical analysis has led to demand for tracking both the rearfoot and forefoot. Gait sandals have been used as an alternative to shoes in an attempt to capture these motions accurately (Eslami *et al.*, 2007). Section 3.2 presents a new multisegment foot model developed to be used with gait sandals. Gait sandals allow the easy application and tracking of skin markers on the whole foot. Furthermore, they offer a sole construction likened to running shoes, with similar outsoles and midsoles. Although their use is increasingly common, little research exists on the effect of gait sandals on lower extremity kinematics. One recent study reported no difference in rearfoot and shank kinematics between barefoot and sandal conditions (Eslami *et al.*, 2007). However, no study has compared kinematics in gait sandals and running shoes.

The purpose of this study was to compare rearfoot and shank kinematics between barefoot, shod and gait sandal conditions during running. It was hypothesised that gait sandals would more closely replicate the kinematics seen when wearing running shoes than when barefoot.

3.3.2 Methods

After institutional ethics approval, 14 male rearfoot strikers (age 21.2 ± 2.1 years; height 1.82 ± 0.06 m; mass 82.0 ± 11.7 kg) gave informed consent to take part in the study (Appendix C). Kinematic data were collected at 500Hz using an eight-camera digital motion capture system (Motion Analysis Corporation, Santa Rosa, USA), whilst force data (Kistler, 9281CA) were collected simultaneously at 1000Hz. Participants completed ten running trials (3.5m/s ± 5%) in three different footwear conditions; barefoot, gait sandals and running shoes.

Both the gait sandals (Bite Orca, Orthosport, Anatom, Edinburgh, UK) and running shoes (Kalenji, Decathlon, UK) used had a neutral foot bed. Ten retroreflective markers were placed on the right foot and shank of each participant, with the shank defined as previously outlined (section 3.2.2) (0Figure 11). In both barefoot and sandal conditions, three markers were placed directly on the right medial, lateral and posterior aspects of the calcaneus at an equal distance to the plantar surface (Nester and Findlow, 2006) (Figure 12). In the running shoe condition, three circular slots were cut in the heel counter to allow the projection of rearfoot markers through the shoe in the same configuration as barefoot and sandal conditions. Thermoplastic plates were firmly fixed to the calcaneus and the markers projected on wands (3cm) through the slots in the shoes (Figure 12). Pilot work using high speed video ensured the slots were of sufficient size so as not to limit motion of the wand makers.

Rearfoot motion was calculated relative to the shank, and shank motion relative to the rearfoot, using a joint coordinate system (Cole *et al.*, 1993). Data were cropped to stance and normalised to a static calibration trial, with peak eversion defined as the maximum value during stance; eversion excursion as the difference between eversion at heel strike and peak eversion (EV); and shank excursion (TIR) as the difference between internal rotation at heelstrike and the peak internal rotation during stance. EV/TIR ratio was calculated by dividing eversion excursion by shank excursion, to provide a measure of the relative degree of rearfoot to shank motion.

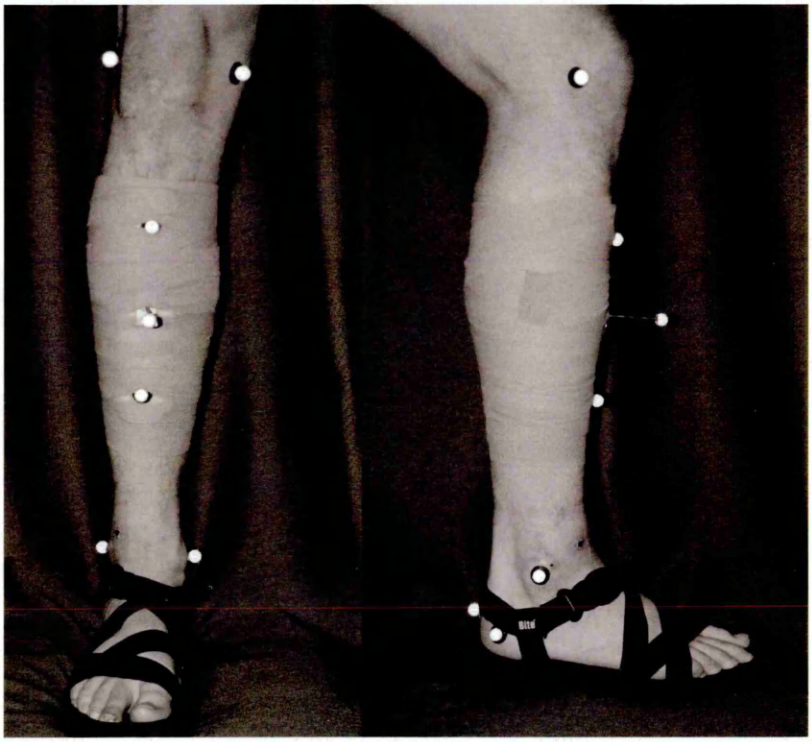


Figure 11. Complete marker set used in all three conditions.

Figure 12. Rearfoot marker placements in the three footwear conditions (barefoot, gait sandals and shod).

Each variable was determined for each of the ten trials per participant and then averaged within the participant. Variables were then averaged across participants for each condition. A one-way repeated measures analysis of variance (α=0.05) with *post-hoc* Tukey least-significant difference test was used to compare differences across conditions for each variable. Effect sizes (ES) were also calculated and interpreted using the following classifications: small 0.20, medium 0.50, and large 0.80 (Cohen, 1988).

3.3.3 Results

Mean kinematic variables are presented in Table 6. Footwear had a significant effect on peak eversion ($F_{1.4,17.9}$=7.6, *P*=0.002). Both barefoot (ES=0.74) and sandals (ES=0.58) resulted in greater peak eversion than running shoes. No differences were observed across conditions for eversion excursion ($F_{1.4,18.5}$=2.1, *P*=0.13). A significant effect for shank excursion was observed ($F_{2,26}$=4.6, *P*=0.023), with barefoot being significantly greater than both sandals (ES=0.62) and running shoes (ES=0.68). No differences were seen between sandals and running shoes (ES=0.07). Footwear also had a significant effect on EV/TIR ratio ($F_{2,26}$=5.6, *P*=0.006), with a difference only found between barefoot and running shoes (ES=0.7).

Table 6. Mean values (SD) for the kinematic variables of interest, across conditions.

	Running (3.5m/s)		
	Barefoot	Gait Sandals	Shod
Peak rearfoot eversion (°)	6.3 (3.4)[a]	5.7 (3.0)[c]	4.2 (2.1)[ac]
Rearfoot eversion excursion (°)	11.3 (2.8)[b]	12.6 (3.1)[b]	12.0 (4.3)
Shank excursion (°)	8.6 (3.0)[ab]	7.1 (1.6)[b]	7.0 (1.4)[a]
EV/TR ratio	1.3 (0.7)[a]	1.6 (0.5)	1.7 (0.4)[a]

a) Significant difference between barefoot and shod. b) Significant difference between barefoot and gait sandal. c) Significant difference between gait sandal and shod.

3.3.4 Discussion

It was hypothesised that sandals would provide an alternative to running shoes for gait analysis. With the exception of peak eversion, sandals closely replicated the rearfoot and shank excursions found with running shoes. Eversion excursion was similar across conditions. These results are partially supported by previous research which reported no difference in eversion excursion or shank excursion between barefoot and sandals (Eslami *et al.*, 2007). Values for eversion excursion and shank excursion in the present study were greater than those reported previously (eversion excursion = 9.0 ± 4.1º, shank excursion = 4.0 ± 2.0º) (Eslami *et al.*, 2007), possibly due to the use of different models.

Both barefoot and sandals resulted in a greater peak eversion than running shoes. This may be due to the lack of structural rearfoot motion control offered in these conditions. However, the similar eversion excursion across conditions suggests a less inverted rearfoot (flatter foot position) at initial contact in both sandal and barefoot conditions. Previous research supports a less inverted rearfoot position when barefoot compared to running in shoes (De Wit *et al.*, 2000). Importantly, no differences were seen for EV/TIR ratio between sandals and running shoe conditions, suggesting that sandals do not alter the rearfoot and shank coupling mechanics compared to running shoes. This study only considered the motions of the rearfoot and shank. However, similar findings may be exhibited in the motions of the forefoot. Future research should seek to establish if a similar trend is observed in the forefoot.

3.3.5 Conclusion

In conclusion, rearfoot and shank excursions were similar between sandals and running shoes, with sandals resulting in greater peak rearfoot eversion compared to running shoes. These findings suggest that sandals can be used in place of running shoes during gait analysis. Given the easy application of markers to all areas of the foot, plus similar outsoles and midsoles to running shoes, sandals appear to be a useful tool for gait analyses, particularly those utilising multisegment foot models. These data support the use of the model presented in section 3.2 for future studies examining gait mechanics when wearing running shoes.

3.4 Between-sessions reliability of selected kinematic variables and peak tibial shock during running

3.4.1 Introduction

The use of motion analysis as a tool for identifying clinical differences between groups is underpinned by the need for reliable kinematic data. Ensuring data are reliable is essential if measures are to be used in research or clinical settings to make informed decisions. Better reliability of selected kinematic variables has been found within a test session (between trials) compared to consecutive sessions (between sessions), in both walking (Carson *et al.*, 2001) and running (Ferber *et al.*, 2002). A significant problem that can reduce between session reliability is the repeatability of marker placements on the skin. The misalignment of markers has been found to introduce an overall shift of absolute joint angles between sessions (Carson *et al.*, 2001). This shift would likely have the most significant effect on absolute kinematic measures such as peak joint angles. Better reliability has been found for joint excursions and velocities, as the range of motion through which a joint passes remains relatively constant (Carson *et al.*, 2001; Ferber *et al.*, 2002). Furthermore, improved reliability has been reported for sagittal plane kinematics compared to frontal or transverse plane movements (Manal *et al.*, 2000; Steinwender *et al.*, 2000; Carson *et al.*, 2001). In the present study, selected kinematic variables focused on sagittal, frontal and transverse plane excursions, peak angles and velocities which define the motions of the foot during loading.

Tibial shock has been found to be a sensitive variable in predicting those with a history of tibial stress injury (Milner *et al.*, 2006b). Skin mounted accelerometers are widely used in biomechanical analysis, although the accuracy of their use has been questioned (Lafortune, 1991). Comparisons of bone and skin mounted transducers have found skin measurements to overestimate accelerations at the tibia (Hennig and Lafortune, 1989; Lafortune *et al.*, 1995) Careful location selection and strict application procedures have been suggested to minimise this problem (Laughton *et al.*, 2003). Further, the use of skin pretension on the tibia has been found to lower accelerations, suggesting an improvement in accuracy on traditional skin mounting techniques (Schnabel and Hennig, 1995; Pearsall *et al.*, 2002). The use of skin pretension on the anterior aspect of the tibia prior to the attachment of the transducer has been used to try and improve the reliability of measurements (Schnabel and Hennig, 1995; Pearsall *et al.*, 2002). Skin pretension was found to reduce tibial shock values by up to 5*g* (Schnabel and Hennig, 1995) and 6.6*g* (Pearsall *et al.*, 2002), compared to conventional skin mounting techniques. Developing a transducer attachment procedure that is both repeatable and produces reliable tibial shock values is essential if this measure is to be used in future research.

Using the multisegment model presented in section 3.2.2, the aim of this study was to assesses the between day reliability of kinematic variables presented in this thesis, during running. Selected kinematic variables focused on sagittal, frontal and transverse plane excursions, peak angles and velocities which define the motions of the foot during loading. A further aim was to establish the reliability of peak tibial shock when using a skin mounted transducer.

3.4.2 Methods

3.4.2.1 Kinematic Measurements

After institutional ethics approval (Appendix C), twelve male rearfoot strikers (age 19.9±1.1, height 1.80±0.07m, mass 83.5±11.6kg.) gave written informed consent to take part in the investigation. Participants were tested on two separate occasions, approximately seven days apart. Sixteen retroreflective markers were placed on the right foot and shank of each participant (Figure 6). All participants wore the same model of running sandal with a neutral midsole (Bite Orca, Orthosport, Anatom, Edinburgh, UK). Participants performed ten over ground running trials after practice ensured they could contact the centre of the force plate (Kistler, 9281CA, Winterthur, Switzerland) without apparent gait alteration - determined through visual inspection by the investigator. A consistent running speed of 3.5m/s (within 5%) was maintained and monitored using timing gates (Brower Timing Systems, Draper, UT, USA). All three-dimensional kinematic data were collected using an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, USA), sampling at 500Hz.

3.4.2.2 Tibial Shock Measurement

Eight of the included participants (age; 21.3 ± 2.2 years, height; 1.82 ± 0.03 m, mass; 77.4 ± 10.2 kg) formed the population for between day reliability analysis of peak tibial shock. Participants were tested on two separate occasions, approximately seven days apart. A force plate (Kistler, 9281CA, Winterthur, Switzerland) and uni-axial accelerometer (PCB Piezotronics, Stevenage, UK) were sampled at 1000Hz and synchronised with the motion capture system. The accelerometer was attached to the distal portion of the antero-medial aspect of the tibia in a manner previously described by Laughton et al. (2003). The sensor was positioned 5 cm above the medial malleolus which allowed for the tibial marker cluster to be placed directly above (Figure 6). To help minimize soft tissue motion, skin tension was applied immediately superior to the attachment site, using two Velcro straps adhered to the skin surface (Pearsall *et al.*, 2002). The skin was pinched and held in place with the Velcro straps before the transducer was attached. The accelerometer was attached to a small piece of thermoplastic (total weight, 1.65grams) and adhered to the skin. The sensitive axis of the transducer was aligned with the long axis of the tibia. Both the transducer and Velcro attachment were over wrapped tightly with bandage around the circumference of the shank. The same investigator applied the transducer on each occasion, ensuring consistency across participants and sessions.

3.4.2.3 Data Analysis

A rearfoot strike pattern was confirmed in all subjects through visual inspection of individual trial data. Raw coordinate data were filtered using a zero-lag fourth order low pass Butterworth filter (8Hz). This cut-off frequency was determined through visual inspection using a range of frequencies (6-12 Hz) with a sample of the data to decide on the most appropriate cut-off. Visual 3D software (C-Motion Inc, Germantown USA) was used to calculate joint rotations. Rearfoot motion was resolved relative to the shank and forefoot motion relative to the rearfoot, using a joint coordinate system adapted from Cole et al. (1993). The sequence of rotations for each joint coordinate system is outlined in section 3.2.2.4. Custom written MATLAB software (Mathsworks, Natick, USA) was used to further analyse kinematic data. Joint angles were cropped to stance using the ground reaction force data (15N threshold) and interpolated to 101 data points using a cubic spline. Angles were not normalised to a standing trial as absolute joint positions were of interest to indicate whether a joint was moving towards the edge of its range of motion. Absolute positions were deemed more important in identifying potential injury risks, information that would be missed if motions were normalized to a static standing trial. Peak rearfoot eversion and peak forefoot abduction were defined as the maximum negative value during stance, while peak forefoot inversion and peak forefoot dorsiflexion were defined as the maximum positive value during stance. Subsequently, rearfoot eversion excursion, forefoot inversion excursion, forefoot dorsiflexion excursion and forefoot abduction excursion were all defined from foot strike to the peak value during stance. Before being cropped and interpolated, joint velocities were calculated using a three point differential method. Joint velocities were defined as the maximum value between foot strike and peak joint angle.

Tibial acceleration data were filtered using a zero-lag fourth order low pass Butterworth filter (50Hz). This frequency had commonly been used in the literature (Shorten and Winslow, 1992) and has been shown to remove signal associated with the resonant frequency of the accelerometer while preserving the signal of interest. Data were then corrected for the effects of angular motion and gravity (Lafortune and Hennig, 1991). The component of the signal due to centripetal acceleration was removed through the calculation of shank angular velocity in the sagittal plane multiplied by the square of the distance the sensor was from the ankle. Further, the signal associated with the effect of gravity was removed by calculating the angle of the shank throughout stance and the changing effects of gravity as stance progresses. The mean tibial acceleration over the stance phase was removed from the signal (Shorten and Winslow, 1992), before peak positive acceleration (tibial shock) was calculated. Between sessions reliability was assessed with intra-class correlation coefficients (ICC), using a two-way random model with single measure reliability and absolute agreement. For analysis purposes, ICC values above 0.75 were indicative of excellent reliability, with values of 0.4-0.75 deemed moderate to good (Fleiss, 1986).

3.4.3 Results

Group mean kinematic variables and measures of reliability for key kinematic variables and tibial shock are presented in Table 7. Further, mean curves for rearfoot and forefoot rotations of interest are presented in Figure 13. The rearfoot was found to evert through the first half of stance, followed by inversion in the second half. Excellent between days reliability was seen for both rearfoot eversion excursion (ICC = 0.78) and peak rearfoot eversion (ICC = 0.75). The forefoot was found to invert with respect to the rearfoot during loading. Poor reliability was observed for all variables associated with forefoot inversion (ICC = 0.33-0.42).

Table 7. Group means (SD) for key variables on both testing occasions, as well as Intra Class Correlations for between sessions reliability.

Variable	Day 1	Day 2	ICC (95% CI)
Tibial Shock (*g*)	5.9 (1.2)	6.1 (1.3)	0.87 (0.50-0.97)
Rearfoot eversion excursion (°)	14.4(2.4)	13.3 (3.0)	0.78 (0.30-0.95)
Peak rearfoot eversion (°)	-4.6 (2.9)	-5.7 (3.0)	0.75 (0.33-0.92)
Rearfoot eversion velocity (°/s)	251.5 (65.6)	235.6 (67.1)	0.61 (0.11-0.87)
Forefoot inversion excursion (°)	3.6 (1.8)	3.0 (1.3)	0.42 (-0.12-0.78)
Peak forefoot inversion (°)	5.5 (4.5)	7.2 (4.1)	0.35 (-0.13 –0.74)
Forefoot inversion velocity (°/s)	83.8 (23.7)	77.6 (25.6)	0.33 (-0.29-0.75)
Forefoot abduction excursion (°)	4.6 (1.4)	4.3 (0.9)	0.84 (0.30-0.97)
Peak forefoot abduction (°)	-11.6 (3.8)	-11.0 (5.5)	0.76 (0.37-0.93)
Forefoot abduction velocity (°/s)	84.2 (24.6)	85.6 (18.4)	0.64 (0.17-0.92)
Forefoot dorsiflexion excursion (°)	5.6 (2.4)	6.0 (2.2)	0.86 (0.62-0.98)
Peak forefoot dorsiflexion (°)	8.7 (4.8)	7.8 (4.2)	0.71 (0.28-0.91)
Forefoot dorsiflexion velocity (°/s)	109.0 (36.5)	108.8 (16.1)	0.43 (-0.21-0.80)

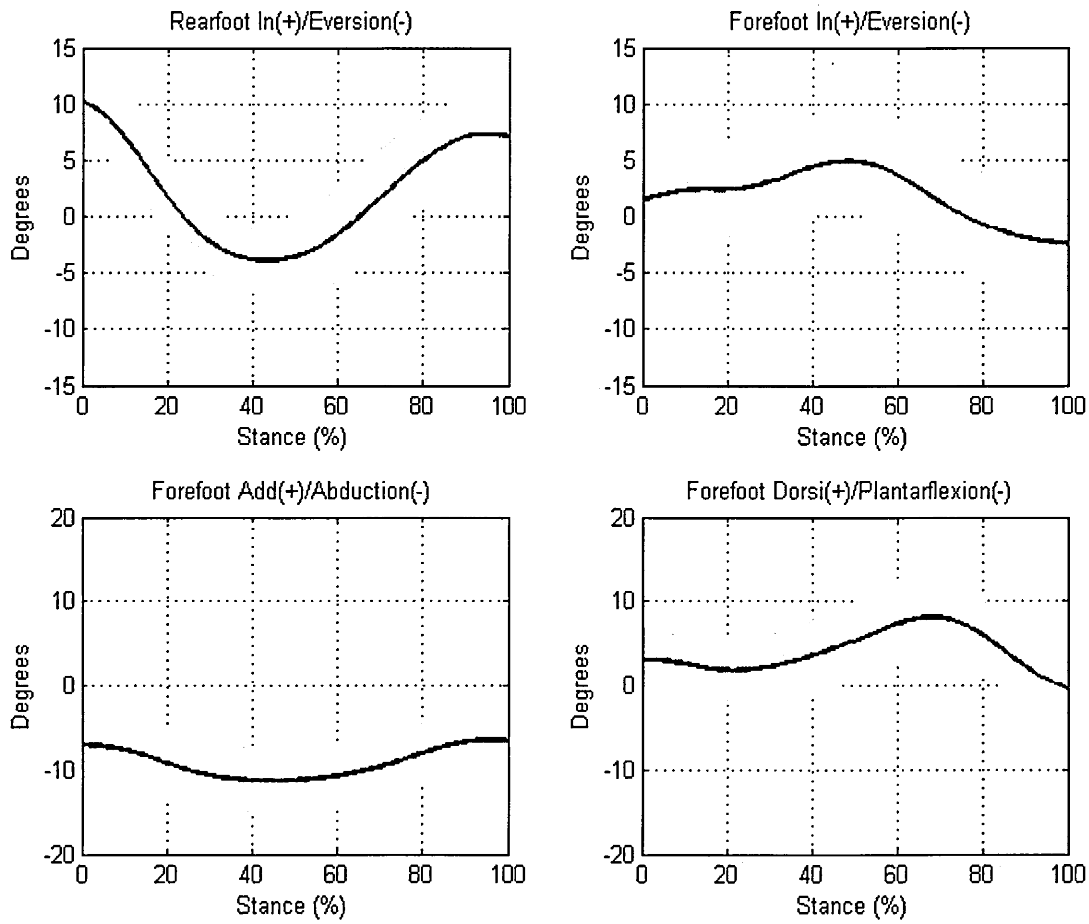


Figure 13. Day 1 angular displacement curves for rearfoot and forefoot rotations during running. The group mean (±SD) for all subjects over 100% of stance.

Abduction of the forefoot relative to the rearfoot was observed during loading, followed by adduction in the latter part of stance. Both forefoot abduction excursion (ICC = 0.84) and peak forefoot abduction (ICC = 0.76) showed excellent between days reliability. Dorsiflexion of the forefoot was seen during the loading phase of gait.

Sagittal plane forefoot motions were found to be highly reliable with excellent between days agreement found for both forefoot dorsiflexion excursion (ICC = 0.86) and peak dorsiflexion (ICC = 0.71).In general, peak joint velocities showed poorer reliability than both joint excursions and absolute peak joint angles. However, both peak rearfoot velocity (ICC=0.61) and peak forefoot abduction velocity (ICC=0.64) still displayed good levels of reliability according to the interpretations of Fleiss (1986). Finally, excellent reliability was seen for peak tibial shock between days (ICC=0.87).

3.4.4 Discussion

The purpose of this study was to assess the between days reliability of selected kinematic variables and peak tibial shock. The results demonstrated excellent reliability for tibial shock and variable levels of reliability for selected kinematic variables. ICC values ranged from 0.33-0.87, with classifications ranging from poor to excellent reliability according to the classifications of Fleiss (1986).

Lower reliability has been reported for frontal and transverse plane movements compared to the sagittal plane (Kadaba *et al.*, 1989; Manal *et al.*, 2000; Steinwender *et al.*, 2000; Carson *et al.*, 2001; Ferber *et al.*, 2002). Previous studies have reported the greatest variability in the transverse plane (Kadaba *et al.*, 1989; Ferber *et al.*, 2002). Furthermore, Manal et al. (2000) found the greatest error between skin and bone markers in the transverse plane. Contrary to these findings, the present study found improved reliability for forefoot transverse plane motions compared to motion of the forefoot in the frontal plane. In the present study, similar levels of reliability were observed for sagittal and transverse plane kinematics.

Both transverse plane motion of the forefoot and frontal plane motion of the rearfoot were found to be reliable measures. However, high variability was seen for all variables associated with forefoot inversion. There are numerous factors which have been suggested to cause variability within the measured kinematics, including measurement error, skin marker movement and physiological variability during gait (Ferber *et al.*, 2002). A further issue when assessing between sessions reliability is marker re-application and the repeatability of marker placements on anatomical locations. Marker positions are used to establish segment coordinate systems about which angles are derived. Small changes in marker position associated with re-application can cause cross talk between planes of motion and cause a phase shift in kinematic data (Kadaba *et al.*, 1989; Carson *et al.*, 2001). When assessing between session reliability, one may expect a systematic shift in inter-segment angles, associated with the re-application of markers on the foot (Kadaba *et al.*, 1989). In an attempt to minimise any systematic error of marker placement, the same experienced investigator applied all markers to try and ensure consistency both between sessions and across subjects. However, in line with previous findings (Carson *et al.*, 2001; Ferber *et al.*, 2002), it was predicted that this shift would have the most significant effect on peak joint angles and these variables would display greater variability than joint excursions and velocities. In the present study, only marginally poorer reliability was seen for peak joint angles compared to joint excursions. Surprisingly, greater variability was seen for joint velocities compared to absolute values (peaks joint angles).

Analysis of the data for the selected kinematic variables between sessions suggests there to be very little systematic error between days, with the majority of error being random (Table 7). While some ICC values for kinematic variables showed only good reliability, in some cases little difference was seen when comparing specific values between days (Table 7). For example, the between session ICC for forefoot abduction velocity was only 0.64 while the mean values for this variable were very similar (Day 1 = 84.2°/s vs. Day 2 = 85.6°/s). These data suggest there to be individual subject differences between testing sessions but these differences appear to be equally and randomly distributed across participants, evidenced by similar mean values. These findings suggest caution when comparing individuals between sessions, but greater confidence between sessions when comparing group data.

Results of the present study were also compared to kinematic patterns previously reported in the literature. Rearfoot eversion excursion values for both days (14.4º and 13.3º) were greater than the 9 ± 4.1° reported in a previous study when running in sandals (Eslami *et al.*, 2007). Values were closer to those previously reported when running shod (12.7±3.5º) (McClay and Manal, 1997). Forefoot abduction excursion for both sessions (4.6° and 4.3°) were found to be smaller than values previously reported using a barefoot model

(Pohl *et al.*, 2007). Values were less than the 5.8±1.6º mean abduction excursion reported for medium pace running but were comparable to the 4.5 ±1.3º observed during walking (Pohl *et al.*, 2007). Forefoot dorsiflexion excursion for both sessions (6.6° and 6.0°) were less than the 8.5° previously reported during jogging (Pohl *et al.*, 2007). However, a reduced range of motion between the forefoot and rearfoot seems logical given that participants in the present study wore gait sandals and the study of Pohl *et al.*, (2007) was conducted barefoot. The rigidity of the shoe sole could act to limit motion of the forefoot relative to the rearfoot.

A small forefoot inversion excursion was observed during the loading phase of stance (3.6º and 3.0º). The forefoot was found to invert with respect to the rearfoot throughout the first half of stance (Figure 13). This suggests that counter rotation of the segments is occurring as the foot is loaded. The kinematic pattern observed in the present study was different to that seen during barefoot running analysis using a similar forefoot-rearfoot model (Pohl *et al.*, 2007). These authors reported initial forefoot eversion during early stance followed by a period of little angular displacement. These findings suggest that the forefoot everts with the rearfoot as the foot is loaded. Again, differences may be evident due to the present study being conducted in gait sandals. Rigidity of the sole may prevent the forefoot from everting with the rearfoot and torsion of the sole may result in an apparent counter rotation of the forefoot. Despite this, evidence of forefoot-rearfoot counter rotations agree with the theory that the foot acts as a twisted plate model (Sarrafian, 1987). This suggestion is supported by the findings of Cornwall and McPoil (2002), who reported counter rotations at the midtarsal joints during the stance phase of barefoot walking. These authors found inversion between the first metatarsal and the navicular to act in the opposite direction to motion between the navicular and calcaneus. Although the model used in the present study only comprises two segments (forefoot and rearfoot), similar kinematic patterns of movement appear to present. Despite this, the questionable reliability of forefoot frontal plane measures and the fact that forefoot-rearfoot frontal plane coupling has been found to be highly subject dependent (Pohl *et al.*, 2007), suggests general trends should be interpreted with caution.

The between sessions reliability of peak tibial shock using the outlined sensor attachment method was found to be excellent. However, peak tibial shock values in the present study (Day 1 = 5.9*g* and Day 2 = 6.1*g*) were found to be considerably higher than those previously reported when running at 3.5m/s (Lafortune, 1991). Lafortune (1991) reported a peak tibial shock of 2.98*g* during running, when using a proximally mounted bone transducer. This finding is consistent with previous literature in which skin compared to bone mounted transducers were found to overestimate accelerations (Lafortune *et al.*, 1995). However, Lafortune (1991) used a proximally mounted transducer and failed to correct for the relative contributions of angular motion and gravity. This correction of acceleration values has been reported to add approximately 1.5*g* for distal (Lake and Greenhalgh, 2005), and up to 4*g* for proximal transducer mountings (Lafortune and Hennig, 1991). Therefore, peak accelerations reported by Lafortune (1991) would be approximately 6.98*g*, much closer to the 5.9*g* and 6.1*g* reported in the present study. Mean peak tibial shock values in the present study were much closer to those reported by Schnabel and Hennig

(1995), who recorded a mean value of 6.8*g* when using skin pretension on participants running at a similar speed.

3.4.5 Conclusion

The aim of this study was to assesses the between day reliability of selected kinematic variables and tibial shock. Excellent between sessions reliability was seen for peak tibial shock when measured using the outlined transducer attachment procedure. Consequently, the attachment method in the present study can be used with confidence in future analysis to establish clinically meaningful differences between groups. After development of a multisegment foot model for use with gait sandals, between sessions reliability analysis on kinematic variables revealed mixed results (ICC values ranged from 0.33-0.86). Good to excellent reliability was found for rearfoot eversion, forefoot dorsiflexion and forefoot abduction related variables. However, caution should be exercised when interpreting data for variables associated with frontal plane motions for the forefoot, specifically, forefoot inversion excursion, peak forefoot inversion and forefoot inversion velocity. In the main, the relative motions of the segments within the present model were found to be consistent with previous findings of foot and ankle kinematics.

3.5 Summary

This Chapter has explored methodological considerations associated with the measurement of forefoot-rearfoot kinematics and tibial shock. A multisegment foot model comprising forefoot, rearfoot and shank segments designed to be used with gait sandals has been developed and outlined (section 3.2). Furthermore, data has been presented to support the use of gait sandals in place of running shoes during gait analysis. Motions of the rearfoot and shank were found to be similar when wearing running shoes and gait sandals. These findings suggest sandals to be a useful tool for gait analyses, and support their use in the present thesis for exploring forefoot-rearfoot kinematics during loading.

The chapter was also concerned with assessing the reliability of selected kinematic variables of interest. In the main, reliability was found to be good to excellent. However, caution should be exercised when interpreting variables associated with forefoot inversion. These data support the use of the model presented in section 3.2 for future studies examining forefoot-rearfoot mechanics during running. Finally, a reliable attachment method for measuring tibial shock was developed. This method may be used with confidence in future analysis to establish clinically meaningful differences between groups.

CHAPTER IV

4 Forefoot-rearfoot kinematics in high and low-arched individuals during running

Work from this chapter has previously been published as follows:

Barnes, A., Wheat, J., and Milner, C.E. (2010). Fore- and rearfoot kinematics in high- and low-arched individuals during running. *Foot and Ankle International,* ***32*** (7).

4.1 Introduction

Foot structure is a suggested intrinsic risk factor for lower extremity injury (Bennell and Brukner, 2005). However, the relationship between foot structure and injury is not clear. Cowan et al. (1993), reported an increased risk of injury associated with greater arch height. Furthermore, individuals with both low-arched and high-arched feet have been suggested to be at greater risk of lower extremity injury (James *et al.*, 1978; Kaufman *et al.*, 1999). High-arched individuals have been found to be at an increased risk of bony shock related injuries such as stress fractures (Sullivan *et al.*, 1984; Giladi *et al.*, 1985; Matheson *et al.*, 1987; Simkin *et al.*, 1989; Williams *et al.*, 2001a). However, the evidence presented in Chapter II of this thesis failed to find a specific relationship between foot type and tibial stress injuries. Despite this, higher tibial shock is a risk factor for stress fractures (Milner *et al.*, 2006b), with higher values reported in high-arched compared to low-arched individuals (Butler *et al.*, 2006). By contrast, a higher incidence of soft tissue injuries has been reported in low-arched individuals, particularly at the knee (Williams *et al.*, 2001a). This evidence suggests that low-arched individuals are potentially more effective at attenuating the repetitive impacts associated with shock related injuries. However, the biomechanical mechanisms behind these differences remain unclear.

It has been suggested previously that high-arched feet tend to be more rigid than low-arched feet, resulting in reduced shock absorption capacity (Matheson *et al.*, 1987; Simkin *et al.*, 1989). The degree of motion permitted in the joints of the foot is determined by both the skeletal structure and soft tissue function. Forefoot-rearfoot kinematics might explain differences between foot types during running. Kinematic differences at the rearfoot have previously been reported between high-arched and low-arched runners (Butler *et al.*, 2006, Williams *et al.*, 2001b). These authors reported 2.9º and 2º, less rearfoot eversion excursion in high-arched runners than in those with low-arched feet, respectively. Importantly, limited rearfoot eversion excursion has been demonstrated to increase the magnitude of impact loading experienced during gait (Perry and Lafortune, 1995).

In addition to motion of the rearfoot at the subtalar joint, forefoot motion might be significant in determining the foot's response to impact loading. During early stance, the loading response of the foot is associated with the unlocking of the transverse tarsal joints. This allows for relative motion between the rearfoot and forefoot (Bojsen-Moller, 1979), particularly at both the talonavicular and calcaneocuboid joints for which significant motions have been reported during slow running (Arndt *et al.*, 2007). Furthermore, motion at these joints is closely associated with lowering of the medial longitudinal arch (Sammarco, 2004). These actions serve to increase the distance between the insertions of the plantar fascia on the calcaneus and metatarsals, resulting in greater stretch of the ligament. This tension allows for more effective utilization of the elastic structures in the foot to reduce shock (Freychat *et al.*, 1996). In support of this theory, greater forefoot abduction range of motion and reduced vertical forces have been reported in flat *(pes planus)* feet during barefoot running (Freychat *et al.*, 1996). Given this evidence, it is suggested that low-arched feet have greater forefoot joint excursions compared to high-arched feet.

In addition to expected differences in individual segment excursions, the coupling between segments may be significant when studying different foot types. The coupling of rearfoot eversion and shank internal rotation has previously been studied in high and low-arched individuals (Nigg *et al.*, 1993; Nawoczenski *et al.,* 1998). Both studies found differences in coupling between segments, with high-arched feet found to have a lower rearfoot to shank coupling ratio. More recent evidence has found that motion of the forefoot is strongly coupled to the rearfoot through joints of the midfoot (Cornwall and McPoil, 2002; Pohl *et al.*, 2006). Since differences are expected in joint motions associated with different foot types, it seems pertinent to explore the coupling between the forefoot, rearfoot and shank in high and low-arched populations. Differences in the coupling of segments throughout stance may result in non-optimal loading of bone or soft tissue structures and may represent a potential injury mechanism associated with the foot.

Within the foot, ligament laxity and muscle function help to govern not only the magnitude of joint excursions, but also the velocity of these rotations. Joint velocities have been suggested to be linked to increased injury risk (Smith et al. 1986). At the rearfoot, significantly higher eversion velocity has been reported in low-arched compared to high-arched runners (Williams *et al.*, 2001b). The rate of forefoot motion as the medial longitudinal arch deforms might be important in determining the shock attenuation capacity of the foot. Given the greater degree of forefoot and rearfoot excursions expected in low-arched individuals, higher joint velocities are also predicted.

Identifying differences in foot kinematics between high-arched and low-arched feet is important in understanding how structure and function interact. Clinically, an understanding of this interaction will allow for the development of injury prevention strategies specific to foot type. The purpose of this study was to investigate forefoot-rearfoot kinematics and tibial shock in high-arched and low-arched individuals. It was hypothesised that low-arched feet would demonstrate greater rearfoot eversion and lower tibial shock compared to high-arched feet. In addition, other variables of interest were forefoot inversion excursion, forefoot dorsiflexion excursion and forefoot abduction excursion, which were predicted to be greater in low-arched individuals. It was also hypothesised that low-arched individuals would demonstrate greater inversion velocity and forefoot abduction velocity than those with high arches. In addition to discrete segment variables, the coupling of these segments during stance was also a focus of this study.

4.2 Methods

4.2.1 Participants

After institutional ethics approval (Appendix D), 108 male participants gave informed consent to take part in the study. All participants were currently free from injury, had no lower extremity abnormalities, and were without a history of foot and ankle surgery. Measurements were taken on the right foot as participants were bearing 10% and then 90% of body weight (Williams and McClay, 2000), using a force platform (Kistler, 9286 AA, Winterthur, Switzerland) with visual display. Participants were seated in the 10% condition and stood with their hands on a countertop for stability in the 90% condition. The right foot was placed on the force platform and the left foot on an adjacent surface of the same height. Participants were asked to apply the desired weight evenly, and maintain it while measurements were taken using a custom built rig based on that of Zifchock et al. (2006b) (Figure 14). A measure of arch height index (AHI) was calculated for both weight bearing conditions, using dorsum height at 50% of foot length divided by truncated foot length (Williams and McClay, 2000) (Appendix E).

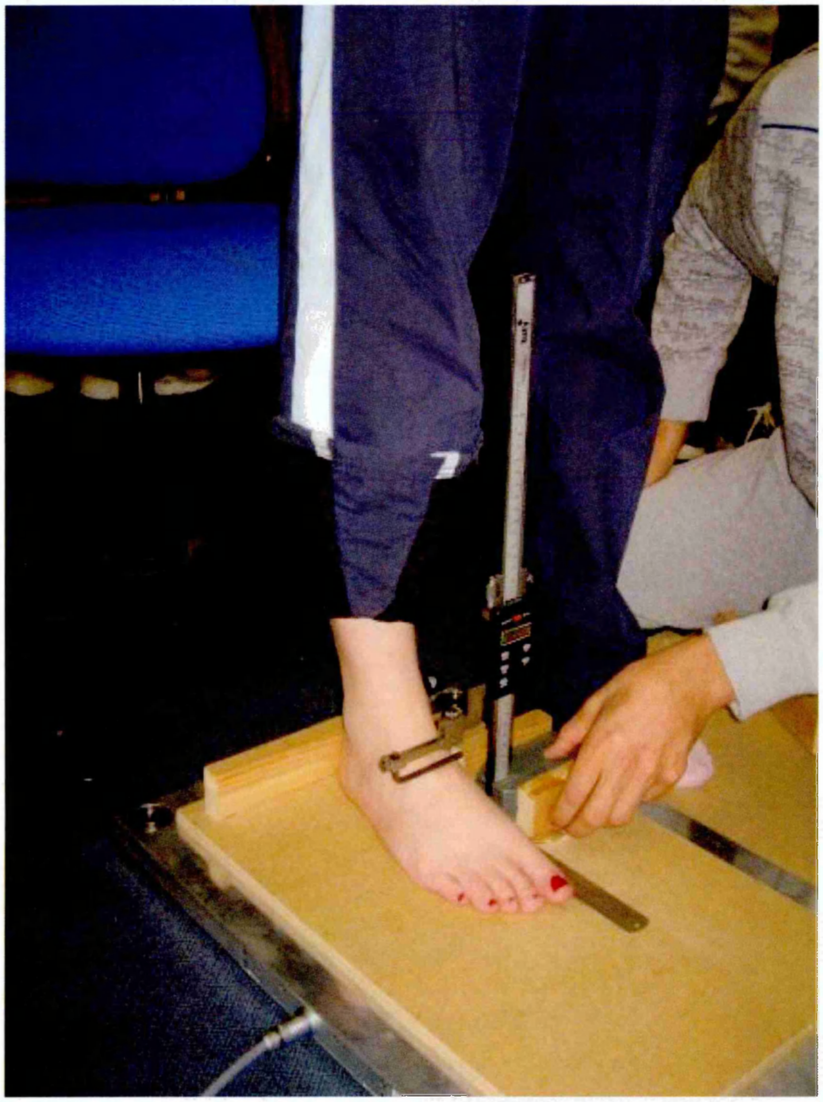


Figure 14. Arch height measurement system used. A custom built rig with load monitored using a portable force platform.

A measure of arch compliance was also calculated based on the relative deformation between 10% (AHI10%) and 90% (AHI90%) weight bearing conditions (Equation 1) (Nigg *et al.*, 1998). This measure is consistently referred to in the literature as 'arch stiffness' and in keeping with convention this term will be used in the present study. A low stiffness score indicates a more rigid arch and a high score indicates a more flexible arch. This method has been used previously to identify the subsets of a population which define low-arched and high-arched individuals (Nachbauer and Nigg, 1992).

$$\text{Arch Stiffness}\left(N^{-1}\right)=\left(\frac{\text{AHI10\%}-\text{AHI90\%}}{\text{AHI10\%}}\right)\frac{10^4}{\text{Bodyweight}} \qquad \text{Equation 1}$$

A sample power calculation (SPSS Inc., Chicago, IL, USA) was performed using pilot data for peak tibial shock and rearfoot eversion excursion. An alpha level of 0.05 and power of 80% were used to determine a clinically significant difference of 15% between groups. In order to detect a meaningful difference in both the variables of interest, a minimum of 15 participants per group was indicated. Therefore, 15 high-arched and 15 low-arched participants were invited back for analysis of their gait and used for comparison in the present study.

4.2.2 Procedure

Participants performed 10 over ground running trials after practice ensured they ran with a rearfoot strike pattern and could contact the centre of the force plate without apparent gait alteration - determined through visual inspection by the investigator. A consistent running speed of 3.5m/s (within 5%) was maintained and monitored using timing gates (Brower Timing Systems, Draper, UT, USA). All participants wore the same model of running sandal with a neutral midsole (Bite Orca, Orthosport, Anatom, Edinburgh, UK). All three-dimensional kinematic data were collected on the right foot and shank (as detailed in section 3.4.2) using an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, USA), sampling at 500Hz. A force plate (Kistler, 9281CA, Winterthur, Switzerland) and uni-axial accelerometer (PCB Piezotronics, Stevenage, UK) were sampled at 1000Hz and synchronised with the motion capture system.

4.2.3 Data Analysis

A rearfoot strike pattern was confirmed in all subjects through visual inspection of individual trial data. The raw coordinate data were filtered using a zero-lag fourth order low pass Butterworth filter (8Hz). Rearfoot motion was resolved relative to the shank and forefoot motion relative to the rearfoot, using a joint coordinate system adapted from Cole et al. (1993) (Section 3.2.2.4). Kinematic data were cropped to the stance phase using the ground reaction force data (15N threshold). Subsequently, rearfoot eversion excursion, forefoot dorsiflexion excursion, forefoot eversion excursion and forefoot abduction excursion were all defined as the difference between the angle at foot strike and the peak angle during stance. Joint velocities, defined as the maximum value between foot strike and peak joint angle, were also calculated for forefoot eversion, forefoot dorsiflexion and forefoot abduction. Before being cropped to stance, tibial acceleration data were filtered using a zero-lag fourth order low pass

Butterworth filter (50Hz) and corrected for the effects of angular motion and gravity (Lafortune and Hennig, 1991). The mean tibial acceleration over the stance phase was removed from the signal (Shorten and Winslow, 1992), before peak positive acceleration (tibial shock) was calculated.

All variables were calculated for each of the 10 trials and averaged within participants, before being averaged within groups. Independent samples t-tests were used to determine whether there were differences in participant characteristics (age, height, mass, AHI 90%, arch stiffness) and tibial shock values between groups. Independent t-tests were conducted on all kinematic variables to determine whether there were significant differences between high and low-arched groups in foot movement patterns. The alpha level for all statistical tests was set at 0.05, with all tests carried out using SPSS 17.0 (SPSS Inc., Chicago, IL, USA). In addition, Cohen's *d* effect sizes (ES) were calculated between groups to aid in the interpretation of these data. According to Cohen (1988), the following classifications were used to interpret effect size values: small effect 0.20, medium effect 0.50, and large effect 0.80.

To examine the coupling between adjacent segments of interest, a cross-correlation technique was used (Li and Caldwell, 1999). A cross-correlation coefficient (with zero phase shift) was calculated between the angular displacement curves of adjacent segments across the stance phase. Couplings of interest in the present study were: shank internal/external rotation and rearfoot eversion/inversion (Rearfoot IN/EV-Shank IR), rearfoot eversion/inversion and forefoot eversion/inversion (Rearfoot IN/EV-Forefoot IN/EV), rearfoot eversion/inversion and forefoot abduction/adduction (Rearfoot

IN/EV-Forefoot ADD/ABD), rearfoot eversion/inversion and forefoot plantarflexion/dorsiflexion (Rearfoot IN/EV-Forefoot DF/PF). Correlation coefficients greater than 0.7 or (less than -0.7) were considered to represent a strong coupling between segments. Correlation coefficients between 0.3 to 0.69 and -0.3 to 0.69 represented a moderate coupling, while coefficients between 0.3 and -0.3 suggested weak coupling.

To assess the similarity of joint excursions for the segment couples of interest, a vector coding technique was used to calculate a coupling angle between 0º and 360º (Chang *et al.*, 2008). Four patterns of coordination were identified whereby coupling angles of between 112.5º to 157.5º and 292.5º to 337.5º were considered to be anti-phase, while coupling angles between 22.5º to 67.5º and 202.5º to 247.5º were considered in-phase coupling. Angles between 0 º to 22.5 º, between 157.5 º to 202.5 º and 337.5 º to 360 º were considered a rearfoot phase and angles between 67.5 º to 112.5 º and 247.5 º to 292.5 º were considered a shank or forefoot phase.

4.3 Results

There were no differences between high and low-arched groups in height and mass (Table 8). The low-arched group was 2.6 years older than the high-arched group ($P = 0.02$). As expected, arch height (AHI 90%) was significantly lower in low-arched compared to high-arched individuals ($P < 0.01$). The means for high and low-arched groups fell 0.85 and 0.92 standard deviations respectively away from the database mean ($n = 108$).

Table 8. Participant characteristics for the normative database, as well as high-arched (HA) and low-arched (LA) groups; mean (SD).

	Database (*n*=108)	HA (*n*=15)	LA (*n*=15)
Age (years)	20.4 (3.3)	19.0 (1.2)	21.6 (2.8)*
Height (cm)	173.2 (30.3)	178.3 (4.3)	180.2 (5.5)
Mass (kg)	76.3 (9.0)	77.3 (11.3)	77.1 (6.5)
AHI 90%	0.352 (0.035)	0.383 (0.016)	0.320 (0.013)*
Arch Stiffness (N^{-1})	0.88(0.30)	0.65(0.13)	1.05(0.30)*

* Significant difference between HA and LA groups at P<0.05 level.

Table 9. Group mean (SD) dependent variables for high-arched (HA) and low-arched (LA) groups, as well as calculated effect sizes (ES).

Variables	HA (*n*=15)	LA (*n*=15)	*p*	ES
Tibial Shock (*g*)	7.2 (1.6)	6.3 (1.6)	0.24	0.56
Rearfoot eversion excursion (°)	13.2(3.3)	13.6(4.2)	0.81	0.12
Forefoot inversion excursion (°)	3.5(2.2)	2.9(1.5)	0.42	0.32
Forefoot abduction excursion (°)	4.7(1.3)	3.8(1.0)	0.037*	0.77
Forefoot dorsiflexion excursion (°)	8.3 (2.9)	5.8 (2.2)	0.016*	0.97
Forefoot inversion velocity (°/s)	85.4(23.6)	75.1(27.5)	0.30	0.40
Forefoot abduction velocity (°/s)	96.0(24.8)	69.3(13.3)*	0.002*	1.34
Forefoot dorsiflexion velocity (°/s)	114.1(28.0)	107.4(33.5)	0.57	0.21

* Significant difference between HA and LA groups at P<0.05 level.

A stiffer arch was observed in the high-arched group and a more flexible arch in the low-arched group ($P = 0.03$) (Table 8). Analysis of tibial shock indicated no statistically significant difference between groups ($P = 0.24$), but a medium effect size (ES = 0.56) was reported for higher tibial shock in the high-arched group (Table 9).

In the sagittal plane, high-arched individuals exhibited significantly greater forefoot dorsiflexion excursion than those with low-arches (large effect size) (Figure 15). However, no differences were seen between groups for forefoot dorsiflexion velocity. Frontal plane kinematics were also examined between high and low-arched groups. Analysis revealed no statistically significant differences between groups for rearfoot eversion excursion (Figure 15), forefoot inversion excursion (Figure 16) and forefoot inversion velocity. The small effect sizes for these variables further support these findings (Table 9). Transverse plane kinematic variables were also compared between groups. High-arched individuals exhibited significantly greater forefoot abduction excursion (medium effect size) (Figure 16) and forefoot abduction velocity than the low-arched group (large effect size). Further investigation of forefoot abduction indicated that the greater forefoot abduction excursion in the high-arched group was the result of a less abducted forefoot position at heelstrike (medium effect size). Peak forefoot abduction was similar between groups (small effect size) (Figure 17).

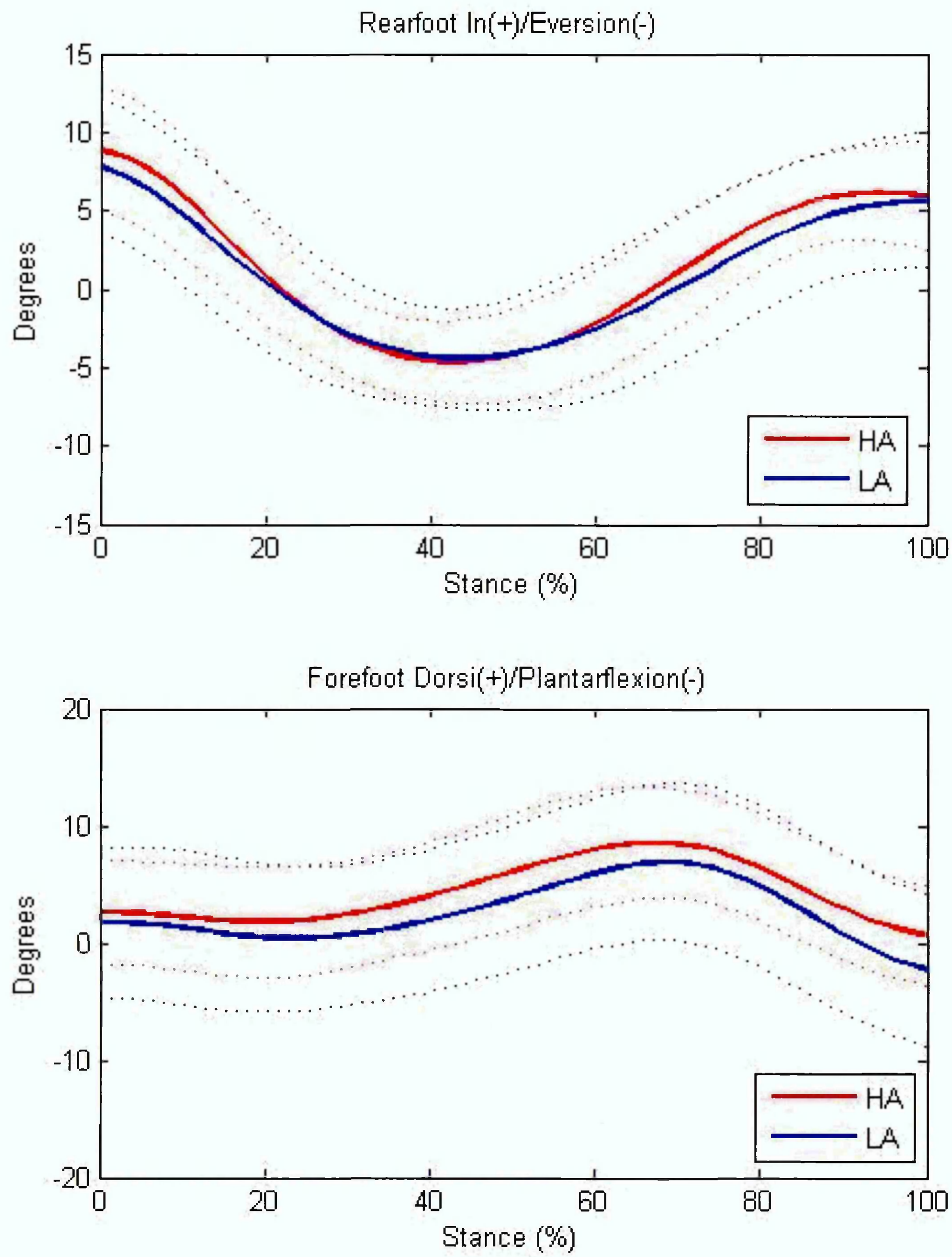


Figure 15. Mean (SD shaded) ensemble angular displacement curves of rearfoot in/eversion and forefoot plantar/dorsiflexion for high-arched (HA) and low-arched (LA) groups.

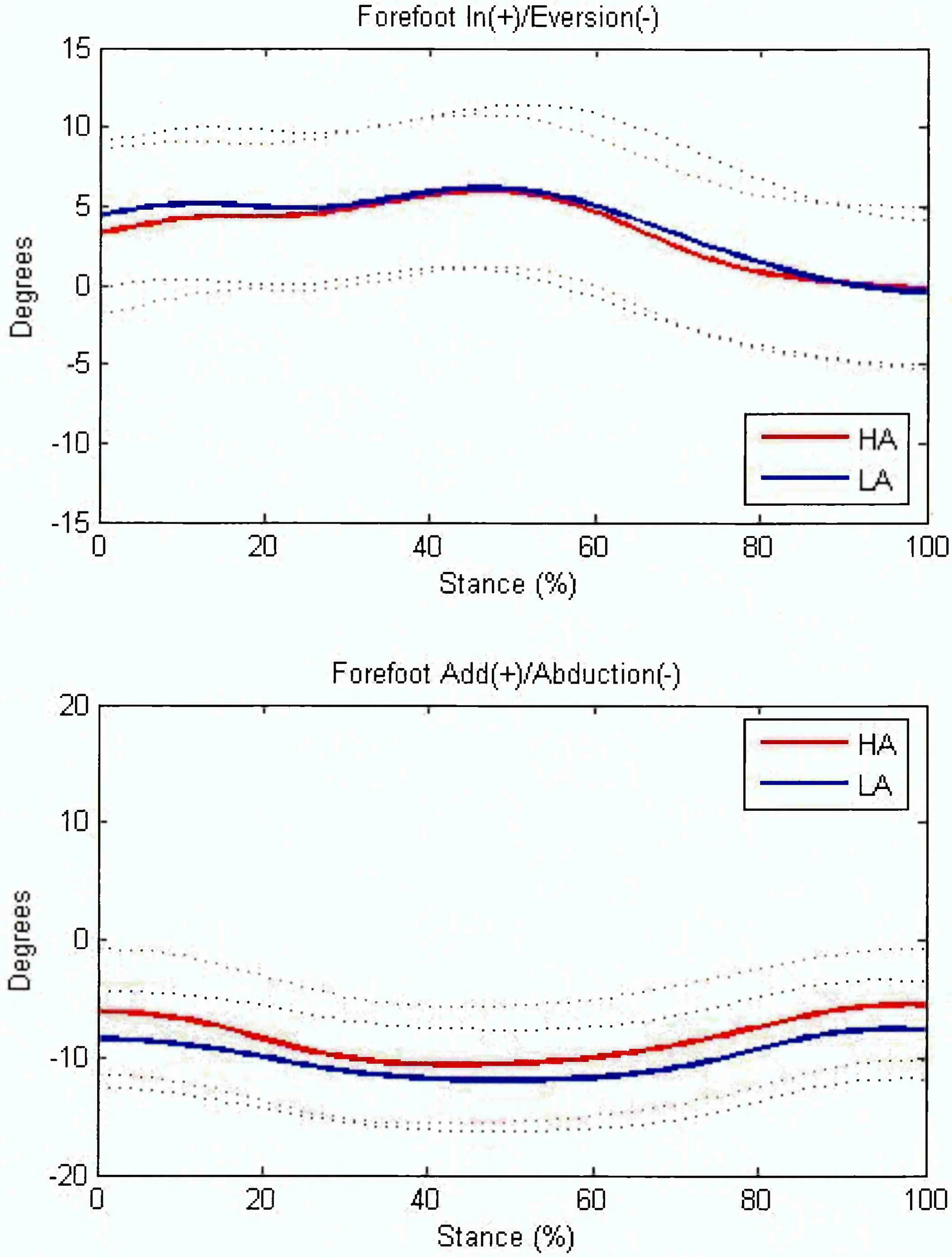


Figure 16. Mean (SD shaded) ensemble angular displacement curves of forefoot inversion/eversion and abduction/adduction for high-arched (HA) and low-arched (LA) groups.

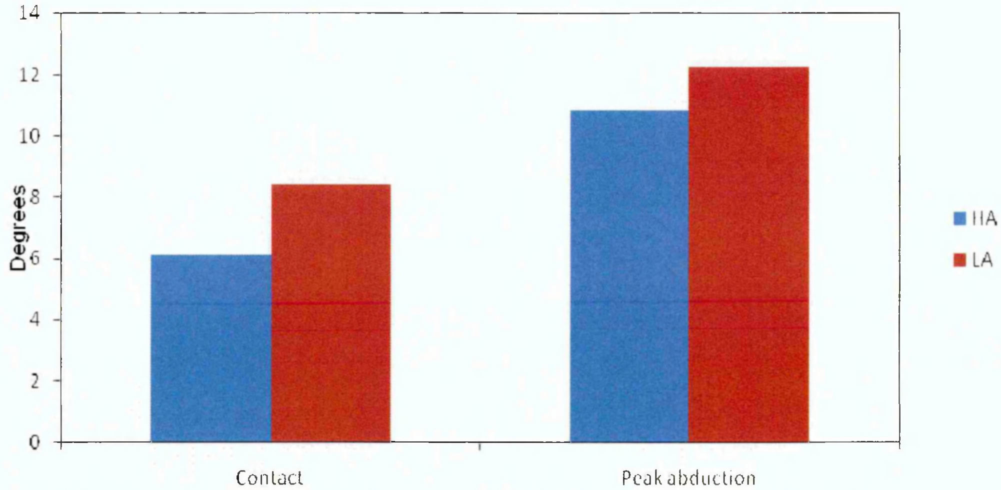


Figure 17. Mean forefoot add/abduction at contact and peak value for High-arched (HA) and low-arched (LA) individuals.

Table 10. Group mean (SD) cross-correlation values for joint couplings of interest in high-arched (HA) and low-arched (LA) groups.

Couplings	HA (*n*=15)	LA (*n*=15)
Rearfoot IN/EV-Shank IR	-0.839(0.095)	-0.867(0.098)
Rearfoot IN/EV-Forefoot IN/EV	-0.194(0.269)	-0.289(0.273)
Rearfoot IN/EV- Forefoot ADD/ABD	0.927(0.061)	0.695(0.457)
Rearfoot IN/EV-Forefoot DF/PF	-0.699(0.208)	-0.639(0.218)

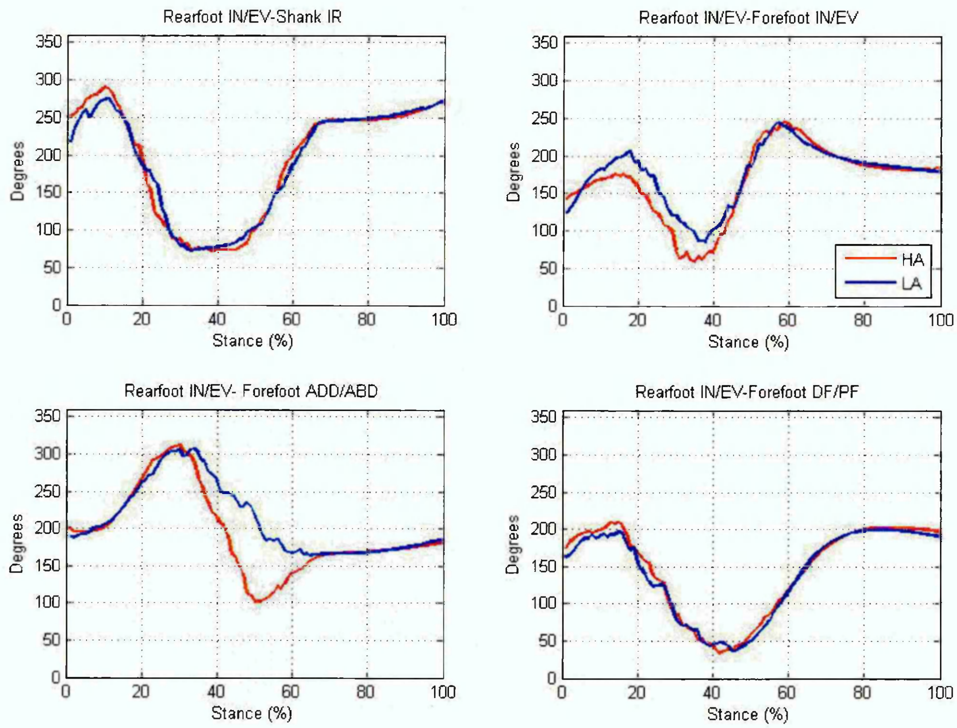


Figure 18. Vector coding curves for the segment couplings of interest in high (HA) and low-arched (LA) individuals. Ensemble means for all subjects are shown throughout stance.

Coupling between Rearfoot IN/EV-Shank IR was consistently high in both high-arch and low-arched groups (Table 10). Moderate but similar correlations were seen for the coupling between Rearfoot IN/EV-Forefoot DF/PF in both groups. The coupling of Rearfoot IN/EV-Forefoot IN/EV was poor for both groups with correlation values suggesting only a weak relationship. A key difference was observed between groups for the rearfoot and forefoot add/abduction coupling with a strong correlation observed for the high-arch group compared to a moderate correlation for the low-arched group. The mean vector coding curves may be seen in Figure 18, in general the curves were similar between groups. A key difference between groups was seen for Rearfoot IN/EV-Forefoot ADD/ABD during the middle portion of stance. During this period the high-arched group was moving in a more anti-phase motion compared to the low-arched group.

4.4 Discussion

The purpose of this study was to investigate forefoot-rearfoot kinematics and tibial shock in high and low-arched individuals. Rearfoot eversion excursion and tibial shock were not significantly different between groups. Differences in forefoot abduction excursion, forefoot dorsiflexion excursion and forefoot abduction velocity between high and low-arched individuals were found in the present study. Differences in the coupling of rearfoot in/eversion with forefoot add/abduction were also noted between foot types. Clear evidence of forefoot-rearfoot motion as a shock attenuation mechanism was not apparent.

As expected, a stiffer arch structure was observed in high-arched individuals and a more flexible arch in low-arched individuals, when measured statically. This supports the findings of Butler et al. (2006), and lends support to the notion that high-arched feet are relatively stiff and low-arched feet are relatively flexible. In the present study, tibial shock values (low-arched = 6.3*g* vs. high-arched = 7.2*g*) were similar to the 6.3*g* (low-arched) and 6.9*g* (high-arched) reported previously in runners wearing running shoes (Butler *et al.*, 2006). It was hypothesised that low-arched individuals would display lower tibial shock values than those with high arches. A trend was noted (medium effect) towards a higher tibial shock in those with high-arches, but differences between groups were not significant. Rather, this study found foot kinematic differences between high and low-arched individuals.

It was hypothesised that low-arched feet would exhibit greater forefoot and rearfoot motions than high-arched feet, reflected in larger joint excursions during early stance. Previous studies have found rearfoot eversion excursion to be a distinguishing variable between high and low-arched individuals (Williams *et al.*, 2001b; Butler *et al.*, 2006). When comparing high and low-arched individuals, these authors reported 2.9º (Butler *et al.*, 2006) and 2º (Williams *et al.*, 2001b) greater rearfoot eversion excursion in low-arched runners than in those with high-arches. The present study failed to find a difference between groups for this variable.

Contrary to the stated hypotheses, less forefoot dorsiflexion excursion and forefoot abduction excursion was observed in the low-arched group than in the high-arched group. Freychat et al. (1996) suggested that greater forefoot abduction increases the distance between the insertions of the plantar fascia on the calcaneus and metatarsals, resulting in greater stretch and more effective utilization of the elastic structures in the foot. Results of the present study indicated that peak forefoot abduction was similar between groups. This suggests that both groups have a similar end range of motion point and does not support the notion of increased distance between forefoot and rearfoot segments in those with low-arches. The smaller forefoot abduction excursion observed in the low-arched group was due to a more abducted position at heelstrike, compared to the high-arched group. This is supported by Hunt and Smith (2004), who reported a more abducted forefoot position at initial contact in *pes planus* feet during barefoot walking. A more abducted foot position at initial contact will reduce the range of motion through which the forefoot can pass before reaching the end range of motion point, potentially altering how loads within the foot are dissipated. This theory may be extended to the sagittal plane where high-arched individuals were seen to have a greater forefoot dorsiflexion excursion. A higher arch structure would allow a greater available range of motion through which the forefoot can pass as the foot is loaded and medial longitudinal arch collapses.

In addition to discrete kinematic variables, the coupling between segments was also investigated using correlation and vector coding techniques. The strength of the correlations for couplings of interest were generally in agreement with those previously reported (Pohl *et al.*, 2006, 2007). Strong couplings were seen for Rearfoot IN/EV-Shank IR and Rearfoot IN/EV-Forefoot ADD/ABD and a moderate coupling for Rearfoot IN/EV-Forefoot DF/PF. In support of previous findings (Pohl *et al.*, 2006, 2007) a weak relationship was seen for the coupling of Rearfoot IN/EV-Forefoot IN/EV. In general no differences in the coupling of Rearfoot IN/EV-Shank IR, Rearfoot IN/EV-Forefoot DF/PF and Rearfoot IN/EV-Forefoot DF/PF segments were seen between high and low-arched groups. This is contrary to the findings that the coupling of the rearfoot and shank differ between high and low-arched individuals (Nigg *et al.*, 1993; Nawoczenski *et al.*, 1998). Key differences were noted for Rearfoot IN/EV-Forefoot ADD/ABD with the high-arched group displaying a strong compared to a moderate relationship in the low-arched group. Furthermore, vector coding analysis revealed more anti-phase motion in the high-arched group during the middle portion of stance. This finding is supported by a greater forefoot abduction excursion in high-arched individuals and suggests a greater counter rotation of the rearfoot and forefoot in high-arched feet as they are loaded. Importantly, these differences in forefoot-rearfoot coupling between foot types do not appear to have been transferred to the proximal segment of the shank.

It was also hypothesised that low-arched individuals would demonstrate greater forefoot joint velocities than those with high-arches. This hypothesis was not supported by the present data. Although forefoot inversion velocity was similar between groups, forefoot abduction velocity was significantly lower in low-arched compared to high-arched individuals. Effective shock absorption is likely determined not only by joint excursion, but also the rate of joint rotation. A smaller abduction excursion occurring over a longer period of time in low-arched individuals resulted in reduced forefoot abduction velocity compared to high-arched individuals. This lower joint velocity could result in reduced average forces during loading through a more gradual attenuation of the shock (Hetsroni *et al.*, 2008). Therefore, while peak tibial shock was not significantly different between groups, differences in foot kinematics during early stance may exist between high and low-arched feet.

Relatively small differences in dependent variables were observed between high-arched and low-arched individuals in the present study. This may be related to the sample examined. In accordance with Nachbauer and Nigg (1992) participants were sampled from the upper and lower quartiles of the normative database. However, when comparing arch height values in the present study with those measured by Butler et al. (2006), mean scores for the high-arched group appeared similar between studies (0.383 ± 0.016 vs. 0.390 ± 0.015). Arch height in the low-arched group in the present study (0.320 ± 0.013) was greater than the low-arched population studied by Butler et al. (0.291 ± 0.018). These differing group characteristics may be the result of methodological differences between studies. We measured arch height at 90% weight bearing (Williams and McClay, 2000), while Butler et al. (2006) used 50% weight bearing. Furthermore, we took measurements on a flat surface while Butler et al. (2006) supported the ball and heel of the foot to allow the arch to maximally lower when standing. As a result of the sampling method, arch height measures for the high-arched and low-arched groups fell 0.85 and 0.92 standard deviations respectively, outside the mean of the normative database. Williams and colleagues (2001b) sampled extremes of arch height were such that high-arched and low-arched groups fell 1.9 and 1.7 standard deviations respectively, outside the mean. Even when sampling such extremes, Williams et al. (2001) only reported a 2º difference in rearfoot eversion excursion between groups. More pronounced differences between groups may have been seen if more extremes of foot type (very low-arched and very high-arched) were sampled. We suggest that future studies of arch type in runners sample those with very high and very low arches.

4.5 Conclusions

In summary, these results provide some insight into forefoot-rearfoot kinematics and tibial shock in high and low-arched individuals. Dependent variables including rearfoot eversion excursion and tibial shock were not significantly different between groups. Clear evidence of forefoot-rearfoot motion as a shock attenuation mechanism was not found. In accordance with previous research (Butler *et al.*, 2006), low-arched feet were more flexible (lower arch stiffness) than high-arched feet. Differences between high and low-arched feet were also observed for some dynamic variables. Forefoot abduction excursion, forefoot dorsiflexion excursion and peak forefoot abduction velocity was greater in high-arched than low-arched individuals. Furthermore, the coupling of rearfoot in/eversion with forefoot add/abduction was also found to differ between foot types. However, importantly this difference did not appear to be transferred proximally to the shank. Given the observed differences in some variables between foot types, forefoot-rearfoot kinematics during stance warrant further investigation to establish any specific link to injury.

CHAPTER V

5 Forefoot-rearfoot kinematics in recreational runners with a history of tibial stress injury

5.1 Introduction

Understanding how structure and function interact is important in identifying injury risk factors. Clinically, an understanding of this interaction will allow for the development of injury prevention strategies. Chapter IV of this thesis investigated how structure and function of the foot interact during running. Results highlighted some key differences in dynamic forefoot variables between foot types. Given this, forefoot-rearfoot motions should be studied in injured populations in an attempt to better understand risk factors. Tibial stress injuries have been associated with repetitive loading of the lower extremities through cyclic activities such as running (Beck, 1998; Bennell and Brukner 2005). There is some evidence to support the notion that those with a history of tibial stress injuries exhibit a unique loading pattern compared to an uninjured population. Individuals who have sustained a tibial stress fracture have been shown to exhibit higher tibial shock (Davis *et al.,* 2004; Milner *et al.,* 2006b). However, factors such as lower extremity alignment and dynamic variables have been suggested to contribute to the risk of stress fracture (Bennell and Brukner, 2005).

The motion permitted by some joints of the foot when loaded may be important in determining injury risk, in particular, the motion of subtalar eversion. At present, evidence is contradictory regarding the role of rearfoot motion as a risk factor for tibial stress injuries. Published data from an on-going study have found greater peak rearfoot eversion during the loading phase of gait in those with a history of tibial stress fracture (Pohl *et al.*, 2008; Milner *et al.*, 2010). Milner et al. (2010) reported a 2.7º greater peak rearfoot eversion in a population of female runners with a history of tibial stress fracture. Several theories have been proposed to explain the tibial stress injury risk associated with greater rearfoot eversion. The orientation of the subtalar joint in the frontal plane and its function as a mitered hinge means that rearfoot eversion is transferred into internal rotation of the tibia (Inman et al., 1981). Chapter IV found a strong coupling relationship between the rearfoot and shank in both high and low-arched individuals. These findings are further supported by previous literature on this coupling relationship (Pohl *et al.*, 2006, 2007; Eslami *et al.*, 2007).

Excessive rearfoot motion may result in increased rotation of the shank and potentially altered loading on the tibia. By contrast, Hetsroni et al. (2008) failed to establish a link between either peak rearfoot eversion or eversion excursion and risk of stress fracture, when conducting a large scale prospective study on military recruits. Despite not identifying any spatial risk factors associated with rearfoot eversion, these authors did find a temporal risk factor associated with reduced time to peak eversion. Hetsroni et al. (2008) reported a reduced stress fracture risk associated with a relative eversion time of greater than 42% of stance. Given that findings have proved conflicting regarding the role of rearfoot motion as a risk factor for tibial stress injury (Pohl *et al.*, 2008; Milner *et al.*, 2010), rearfoot variables are worthy of further investigation in injured populations.

Forefoot kinematics during gait have not previously been investigated in those with a history of tibial stress injury. In addition to the rearfoot, forefoot motion might be significant in determining the foot's response to loading. As previously noted (section 2.1.6.2), significant motion occurs between the rearfoot and forefoot during early stance, particularly at the joints of the midfoot (Arndt *et al.*, 2007; Bojsen-Moller, 1979). The foot has been suggested to act as a twisted plate model, which produces counter rotations of the forefoot with respect to the rearfoot during loading (Sarrafian, 1987). It is well established that the rearfoot everts from heel strike through to midstance (Reinschmidt *et al.*, 1997), the twisted plate model suggests that this motion is accompanied by forefoot inversion. In addition, rearfoot eversion is also accompanied by dorsiflexion and abduction of the forefoot. These motions of the forefoot have been found to be coupled with motions of the rearfoot (Cornwall and McPoil, 2002; Pohl *et al.*, 2006). Results from Chapter IV of this thesis suggest forefoot-rearfoot motion are coupled during stance, in particular, rearfoot eversion with forefoot dorsiflexion and forefoot abduction. Given that previous research (Pohl *et al.*, 2008; Milner *et al.*, 2010) has found those with a history of tibial stress injury exhibit a higher peak rearfoot eversion compared to a matched control group, it is suggested that greater separation of the midfoot joints will occur resulting in and higher peak forefoot dorsiflexion and forefoot abduction angles in the injury group. Therefore, peak forefoot segment angles during stance were considered primary variables of interest in the present study.

Previous research has suggested that motions of the forefoot a rearfoot are linked through the joints of the midfoot. Given the strong coupling of the rearfoot and shank, it is suggested that motions of the forefoot may have some influence on the transfer of movement between the rearfoot and shank. Evidence from looking at high and low-arched groups suggests that some forefoot-rearfoot coupling relationships change between foot types. Since there is a reported association between rearfoot eversion and tibial stress fracture risk (Pohl *et al.*, 2008; Milner *et al.*, 2010), it appears pertinent to include the coupling mechanics of the forefoot-rearfoot and their potential influence on more proximal segments.

Within the foot, ligament laxity and muscle function help to govern not only the magnitude of peak joint angles but also the velocity of joint rotations. Analysis of joint velocities may provide insight into potential risk factors associated with tibial stress injuries. High rearfoot eversion velocity has been found to be a risk factor associated with shin splints (Messier and Pittala, 1988). These authors found greater eversion velocity in the injured compared to control groups. However, it should be noted that the injury group in this study contained subjects with undefined shin splints. By contrast, Hetsroni et al. (2008) failed to establish rearfoot eversion velocity as a risk factor for tibial and femoral stress fractures. However, these authors did find a reduced time of rearfoot eversion excursion in those with a history of tibial stress injuries (Hetsroni *et al.*, 2008). Given that temporal variables are key in determining the rate of joint motion, a greater rearfoot eversion velocity may be expected in those with a history of tibial stress injury. Separation at the joints of the midfoot is determined by soft tissue structures in the foot; in particular, the spring ligament and plantar fascia (Van Boerum and Sangeorzan, 2003). In addition to the degree of separation between the forefoot and rearfoot, the velocities of these rotations might be important in determining the loading response of the foot. Key differences in forefoot abduction velocity were noted between foot types in the present thesis. Therefore, in addition to peak joint angles, peak joint velocities of the selected rotations were also analysed.

The purpose of this study was to investigate forefoot-rearfoot kinematics in relation to tibial stress injury risk. In particular, it aimed to establish if differences exist in forefoot-rearfoot mechanics between recreational runners with a history of tibial stress injury and a control group. It was hypothesised that those with a history of tibial stress injury would demonstrate greater peak joint angles than a control group, in particular; peak rearfoot eversion, peak forefoot inversion, peak forefoot dorsiflexion and peak forefoot abduction. Further variables of interest included selected joint velocities of interest. It was hypothesised that those with a history of tibial stress injury would demonstrate greater joint velocities than a control group. In addition to discrete segment variables, the coupling of these segments during stance was also a focus of this study.

5.2 Methods

5.2.1 Participants

After institutional ethics approval (Appendix F), all participants gave written informed consent before starting the study. Male and female recreational runners between the ages of 18 and 45 years who reported running at least 10 miles per week on average were recruited into the study. Six participants (4 males and 2 females) with a history of tibial stress injury were compared to six control participants with no history of bone related lower extremity injury. Control participants were recruited from local running clubs to represent a group of non injured runners who adopted a rearfoot strike pattern from which individuals were paired with injured subjects for comparison. The tibial stress injury and control participants were matched in terms of gender, age and weekly mileage.

The tibial stress injury group comprised two subjects with a previously reported tibial stress fracture, confirmed by a medical professional using bone scans. The remaining four participants were diagnosed through assessment by the principle investigator as having a tibial stress injury within the last 12 months. A detailed training history was taken to provide insight into the nature of the injury (Fredericson and Wun, 2003). Participants were questioned about the presence of pain and type of pain they experienced after activity, during activity, and during daily ambulation. Assessment parameters included presence or absence of focal tenderness and swelling at the site of pain, direct or indirect percussion tenderness, and pain with resisted manual muscle testing. These symptoms have previously been outlined in the literature and validated using diagnostic imaging techniques (Fredericson *et al.*, 1995, Batt *et al.*, 1998). The same researcher performed the interview and assessment on all prospective subjects to ensure consistency in the approach taken and the diagnosis. All participants had no history of lower extremity abnormalities and were without a history of foot and ankle surgery. At the time of testing participants were free from injury and had been pain free and active for at least eight weeks.

Static foot measurements were taken on the involved limb of the injury group and the right limb of the control group. Measures were taken in 10% and 90% weight bearing conditions and used to calculate measures of AHI (Williams and McClay, 2000), as detailed in section 4.2.1. Further, a measure of arch stiffness was also calculated based on the relative deformation between 10% and 90% weight bearing conditions (Nigg *et al.*, 1998) (Section 4.2.1).

5.2.2 Procedure

Sixteen reflective markers were attached to the skin of the shank, rearfoot and forefoot as described previously in section 3.2.2. A uni-axial accelerometer (PCB Piezotronics, Stevenage, UK) was attached to the anterior medial aspect of the tibia. All participants wore the same model of running sandal with a neutral midsole (Bite Orca, Orthosport, Anatom, Edinburgh, UK). Participants performed 10 over ground running trials after practice ensured they could contact the centre of the force plate without apparent gait alteration - determined through visual inspection by the investigator. A consistent running speed of 3.5m/s (within 5%) was maintained and monitored using timing gates (Brower Timing Systems, Draper, UT, USA). All three-dimensional kinematic data were collected using an eight camera digital motion capture system (Motion Analysis Corporation, Santa Rosa, USA), sampling at 500Hz. A force plate (Kistler, 9281CA, Winterthur, Switzerland) and the accelerometer were sampled simultaneously at 1000Hz and synchronised with the motion capture system. Data were collected from the involved limb in the injury group and the right limb in the control group, since there was no reason to prefer a particular side.

5.2.3 Data Analysis

A rearfoot strike pattern was confirmed in all subjects through visual inspection of individual trial data. The raw coordinate data were filtered using a zero-lag fourth order low pass Butterworth filter (8Hz). Rearfoot motion was resolved relative to the shank, and forefoot motion relative to the rearfoot, using a joint coordinate system adapted from Cole et al. (1993) (section 3.2.2.4). Kinematic data were cropped to the stance phase using the ground reaction force data (15N threshold). Subsequently, peak rearfoot eversion and peak forefoot abduction were defined as the maximum negative value during stance, while peak forefoot inversion and peak forefoot dorsiflexion were both defined as the maximum positive values during stance. Joint velocities, defined as the maximum value between foot strike and peak joint angle, were also calculated for forefoot inversion, forefoot dorsiflexion and forefoot abduction. Before being cropped to stance, tibial acceleration data were filtered using a zero-lag fourth order low pass Butterworth filter (50Hz) and corrected for the effects of angular motion and gravity (Lafortune and Hennig, 1991). The mean tibial acceleration over the stance phase was removed from the signal (Shorten and Winslow, 1992), before peak positive acceleration (tibial shock) was calculated.

All variables were calculated for each of the 10 trials and averaged within participants, before being averaged within groups. Independent t-tests were conducted on all kinematic variables to determine whether there were significant differences between tibial stress injury and control groups in foot movement patterns. The alpha level for all statistical tests was set at 0.05, with all tests carried out using SPSS 17.0 (SPSS Inc., Chicago, IL, USA). Cohen's *d* effect sizes (*ES*) were calculated for all dependent variables to aid interpretation of these data. According to Cohen (1988), the following classifications were used to interpret effect size values: small effect 0.20, medium effect 0.50, and large effect 0.80.

To examine the coupling between adjacent segments of interest, a cross-correlation technique was used (Li and Caldwell, 1999). A cross-correlation coefficient (with zero phase shift) was calculated between the angular displacement curves of adjacent segments across the stance phase. Couplings of interest in the present study were: shank internal/external rotation and rearfoot eversion/inversion (Rearfoot IN/EV-Shank IR), rearfoot eversion/inversion and forefoot eversion/inversion (Rearfoot IN/EV-Forefoot IN/EV), rearfoot eversion/inversion and forefoot abduction/adduction (Rearfoot IN/EV-Forefoot ADD/ABD), rearfoot eversion/inversion and forefoot plantarflexion/dorsiflexion (Rearfoot IN/EV-Forefoot DF/PF). Correlation coefficients greater than 0.7 or (less than -0.7) were considered to represent a strong coupling between segments. Correlation coefficients between 0.3 to 0.69 and -0.3 to 0.69 represented a moderate coupling, while coefficients between 0.3 and -0.3 suggested weak coupling.

To assess the similarity of joint excursions for the segment couples of interest, a vector coding technique was used in accordance with Chang et al. (2008). Coupling classifications used were consistent with Chang et al. (2008), whereby coupling angles of between 112.5º to 157.5º and 292.5º to 337.5º were considered to be anti-phase, while coupling angles between 22.5º to 67.5º and 202.5º to 247.5º were considered in-phase coupling. Angles between 0 º to 22.5 º, between 157.5 º to 202.5 º and 337.5 º to 360 º were considered a rearfoot phase and angles between 67.5 º to 112.5 º and 247.5 º to 292.5 º were considered a shank or forefoot phase. Stance was subdivided into three time intervals, early (1-33%), mid (34-66%) and late stance (67-100%), with the percentage of time spent in each of the four phases calculated for these three periods.

5.3 Results

A comparison of participant characteristics between the tibial stress injury and control groups can be seen in Table 11. Static arch height (AHI 90%) was found to be similar between groups (small effect size). Mean arch height values for both groups studied were close to the group mean previously reported in a sample of 108 participants (Chapter IV, : AHI 90% = 0.3520). Similar arch stiffness values were observed between tibial stress injury and control groups, supported by a small effect size (small effect size). Analysis of tibial shock indicated groups were similar, although a moderate trend was noted towards a higher tibial shock in the tibial stress injury group (medium effect size) (Table 11).

The comparison of dependent variables between runners in the tibial stress injury and control groups is presented in Table 12 and Figures 19-20. With the exception of forefoot abduction velocity, none of the examined kinematic variables were found to be significantly different between groups. However, a strong trend was noted towards a greater peak rearfoot eversion in those with a history of tibial stress injury, compared to controls (large effect size). No differences were seen between groups for rearfoot eversion velocity (small effect size). Frontal plane analysis of forefoot motion revealed no differences between groups for either peak forefoot inversion or peak forefoot inversion velocity (small effect sizes). In the sagittal plane, while peak forefoot dorsiflexion was not different between groups, a strong trend was noted towards a greater forefoot dorsiflexion velocity in those with a history of tibial stress injuries (large effect size).

Analysis of transverse plane variables revealed that groups were similar for peak forefoot abduction (small effect size). However, forefoot abduction velocity was found to be greater in the tibial stress injury compared to the control group (large effect size). Further investigation of forefoot abduction indicated that an increased excursion occurring over a shorter period of time resulted in a greater forefoot abduction velocity in the injury group. Forefoot abduction excursion was found to be greater in the tibial stress injury compared to the control groups (tibial stress injury = 5.5º, control group = 3.8º large effect size). Furthermore, peak forefoot abduction was found to occur earlier in those with a history of tibial stress injury compared to controls (tibial stress injury = 43%, control group = 48%, medium effect size) (Figure 20).

Rearfoot IN/EV-Shank IR correlations suggested a strong relationship between segments and rearfoot IN/EV-Forefoot IN/EV indicated a weak relationship between segments. No differences were observed for these couplings between groups (Table 13).For rearfoot IN/EV-Forefoot ADD/ABD, a strong relationship was observed for both groups, with the tibial stress injury group displaying marginally more coupled movement of the segments. A key difference was observed between groups for Rearfoot IN/EV- Forefoot DF/PF coupling with a strong correlation observed for the control group compared to a moderate correlation for the tibial stress injury group. Further analysis of these joint couplings using a vector coding method revealed some key differences between groups (Figure 21).

The Rearfoot IN/EV-Shank IR coupling in the first third of stance revealed that the tibial stress injury group spent a greater percentage of time in a shank phase (tibial stress injury = 36%, control group = 14%) while the control group spent a greater portion in a rearfoot phase (tibial stress injury = 13%, control group = 44%) (Table 14a). During the first third of stance, a difference in in-phase motion was seen between groups for Rearfoot IN/EV- Forefoot DF/PF coupling. The control group spent a larger portion of this time moving in-phase (22%) compared to the tibial stress injury group (4%) (Table 14d).

Table 11. Participant characteristics for the control (CON) and tibial stress injury (TSI) groups; mean (SD).

	CON (*n*=6)	TSI (*n*=6)
Age (years)	27.8 (7.4)	28.0 (11.2)
Height (cm)	173.4 (0.03)	181.2 (0.12)
Mass (kg)	71.6 (8.2)	77.2 (15.1)
Mileage (mil week)	14.5 (4.6)	12.8 (2.6)
AHI 90%	0.347(0.022)	0.354 (0.011)
Arch Stiffness (N^{-1})	0.90 (0.24)	0.79 (0.28)
Tibial Shock (*g*)	6.5 (1.7)	7.2 (0.4)

Table 12. Group mean (SD) dependent variables for control group (CON) and tibial stress injury group (TSI), as well as calculated effect sizes (ES (confidence interval)).

Variables	CON	TSI (*n*=6)	*p*	ES
Peak rearfoot eversion (°)	4.4(2.9)	6.3(1.3)	0.17	0.85(0.2-1.5)
Rearfoot eversion velocity (°/s)	224.0(96.4	245.0(45.3)	0.66	0.28(-4.0-4.6)
Peak forefoot inversion (°)	8.6(2.6)	7.5(3.3)	0.55	0.38(-0.2-1.0)
Forefoot inversion velocity (°/s)	88.1(29.0)	82.1 (11.3)	0.65	0.27(-1.1-1.6)
Peak forefoot abduction (°)	11.5(5.8)	13.7 (5.3)	0.53	0.38(-0.3-1.1)
Forefoot abduction velocity (°/s)	71.2(10.8)	103.6 (25.6)	0.01*	1.60(-5.0-8.2)
Peak forefoot dorsiflexion (°)	7.2(2.1)	8.6(4.3)	0.47	0.41(-0.2-1.1)
Forefoot dorsiflexion velocity (°/s)	97.3(22.0)	124.0(35.0)	0.15	0.91(-4.6-6.4)

* Significant difference between CON and TSI groups at *P*<0.05 level.

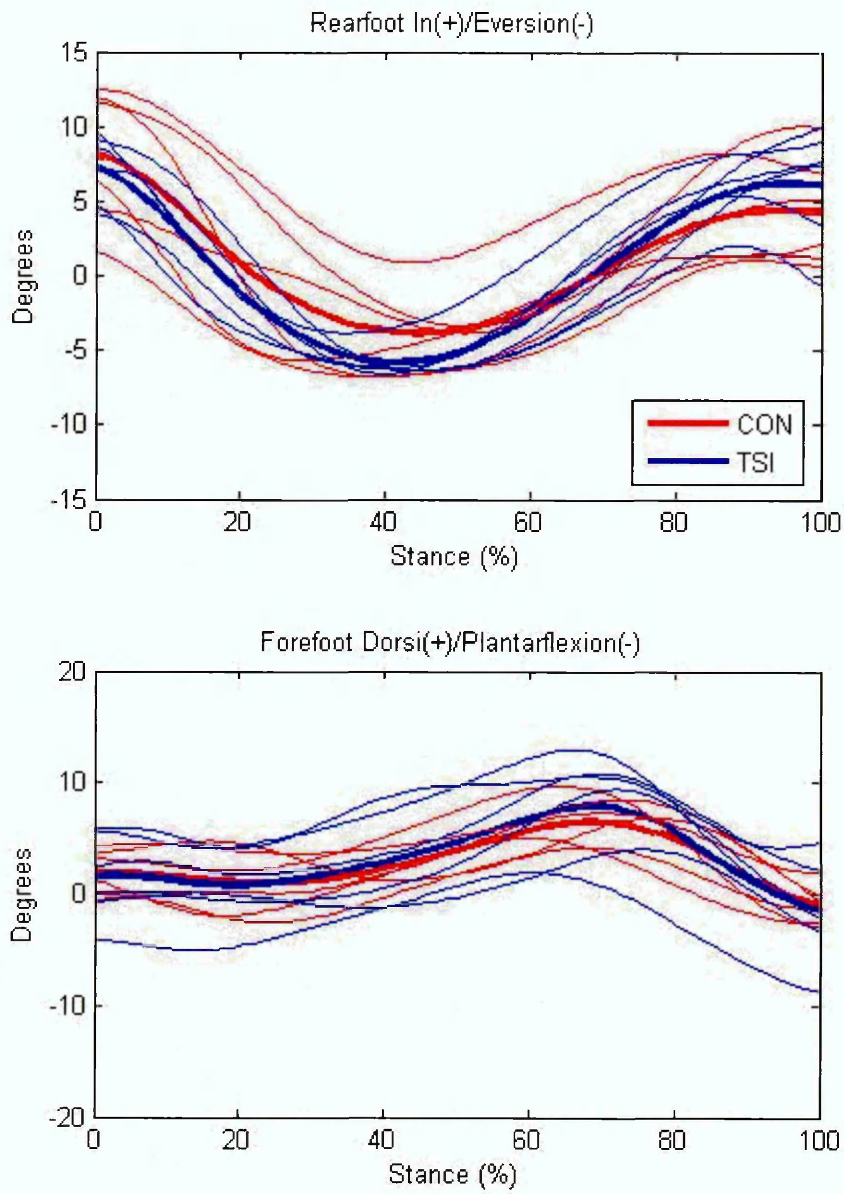


Figure 19. Rearfoot in/eversion and forefoot plantar/dorsiflexion curves for subjects in the tibial stress injury (TSI) and control (CON) groups. Group mean curves appear in bold.

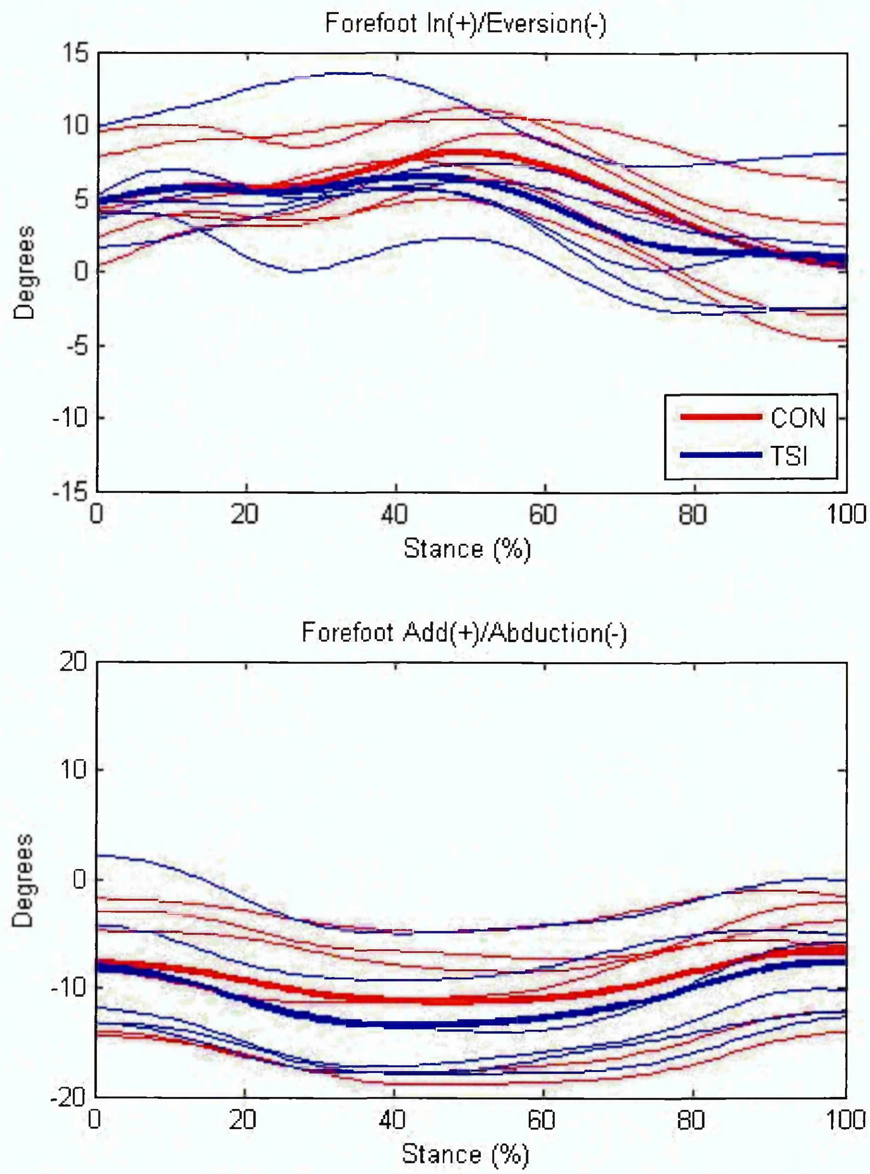


Figure 20. Forefoot in/eversion and add/abduction curves for subjects in the tibial stress injury (TSI) and control (CON) groups. Group mean curves appear in bold.

5.4 Discussion

The purpose of this study was to investigate forefoot-rearfoot kinematics in individuals with a history of tibial stress injury and a matched control group. Hypotheses focused on key variables within the foot which define its motions during loading. Key differences were found in peak rearfoot eversion, forefoot abduction velocity and forefoot dorsiflexion velocity between groups. Small differences between subjects were also noted for coupling relationships between segments, results may provide some insight into potential injury mechanisms.

It was hypothesised that individuals with a history of tibial stress injury would exhibit greater peak rearfoot eversion compared to controls. This hypothesis was supported with the tibial stress injury group found to have 1.9° greater peak rearfoot eversion than the control group. These findings support those of previous studies which have highlighted peak eversion to be a risk factor associated with tibial stress fractures (Pohl *et al.*, 2008; Milner *et al.,* 2010). Milner et al. (2010) found a 2.7º greater peak rearfoot eversion in those with a history of tibial stress fracture compared to controls. At the forefoot, hypotheses were not supported for peak forefoot inversion, peak forefoot dorsiflexion and peak forefoot abduction. These variables were found to be similar between groups. These data do not support the notion of increased end range of motion resulting in an increased torsional association between rearfoot eversion and forefoot abduction during loading (Pohl *et al.,* 2006).

Table 13. Group mean (SD) for cross-correlation joint couplings of interest in tibial stress injury (TSI) and control (CON) groups.

Couplings	CON (*n*=6)	TSI (*n*=6)
Rearfoot IN/EV-Shank IR	-0.828(0.075)	-0.838(0.112)
Rearfoot IN/EV-Forefoot IN/EV	-0.112(0.354)	-0.207(0.361)
Rearfoot IN/EV- Forefoot ADD/ABD	0.889(0.068)	0.948(0.032)
Rearfoot IN/EV-Forefoot DF/PF	-0.983(0.579)	-0.698(0.183)

Table 14. Segmental joint couplings of interest during the three phases of stance. Five coordination patterns were considered: in-phase (IP), anti-phase (AP), Shank phase (SH), Rearfoot phase (RF) and forefoot phase (FF). Values represent the percentage of time spent in each pattern.

a. Rearfoot IN/EV-Shank IR

	0-33%				34-66%				67-100%			
	IP	AP	SH	RF	IP	AP	SH	RF	IP	AP	SH	RF
CON	19(11)	23(10)	14(7)	44(18)	42(8)	5(2)	5(4)	48(7)	47(27)	1(1)	0(0)	52(27)
TSI	25(9)	26(16)	36(19)	13(6)	41(29)	8(12)	3(1)	48(9)	28(18)	2(3)	0(1)	70(17)

b. Rearfoot IN/EV-Forefoot IN/EV

	0-33%				34-66%				67-100%			
	IP	AP	FF	RF	IP	AP	FF	RF	IP	AP	FF	RF
CON	13(8)	17(10)	7(4)	63(14)	25(15)	10(4)	10(5)	56(17)	16(13)	0(0)	1(2)	83(14)
TSI	17(12)	16(6)	6(4)	62(13)	22(9)	11(4)	10(4)	56(15)	11(6)	0(0)	0(1)	89(6)

c. Rearfoot IN/EV- Forefoot ADD/ABD

	0-33%				34-66%				67-100%			
	IP	AP	FF	RF	IP	AP	FF	RF	IP	AP	FF	RF
CON	18(13)	11(4)	7(4)	65(15)	5(3)	13(8)	2(1)	80(9)	0(0)	3(5)	0(0)	97(0)
TSI	20(8)	17(4)	9(3)	54(13)	5(3)	9(7)	2(1)	83(8)	0(0)	2(2)	0(1)	98(2)

d. Rearfoot IN/EV-Forefoot DF/PF

	0-33%				34-66%				67-100%			
	IP	AP	FF	RF	IP	AP	FF	RF	IP	AP	FF	RF
CON	22(15)	9(7)	7(5)	61(16)	35(9)	36(17)	6(4)	15(6)	21(23)	6(6)	1(1)	72(24)
TSI	4(3)	7(3)	4(3)	67(18)	37(17)	37(5)	7(5)	15(11)	30(21)	5(5)	1(1)	65(19)

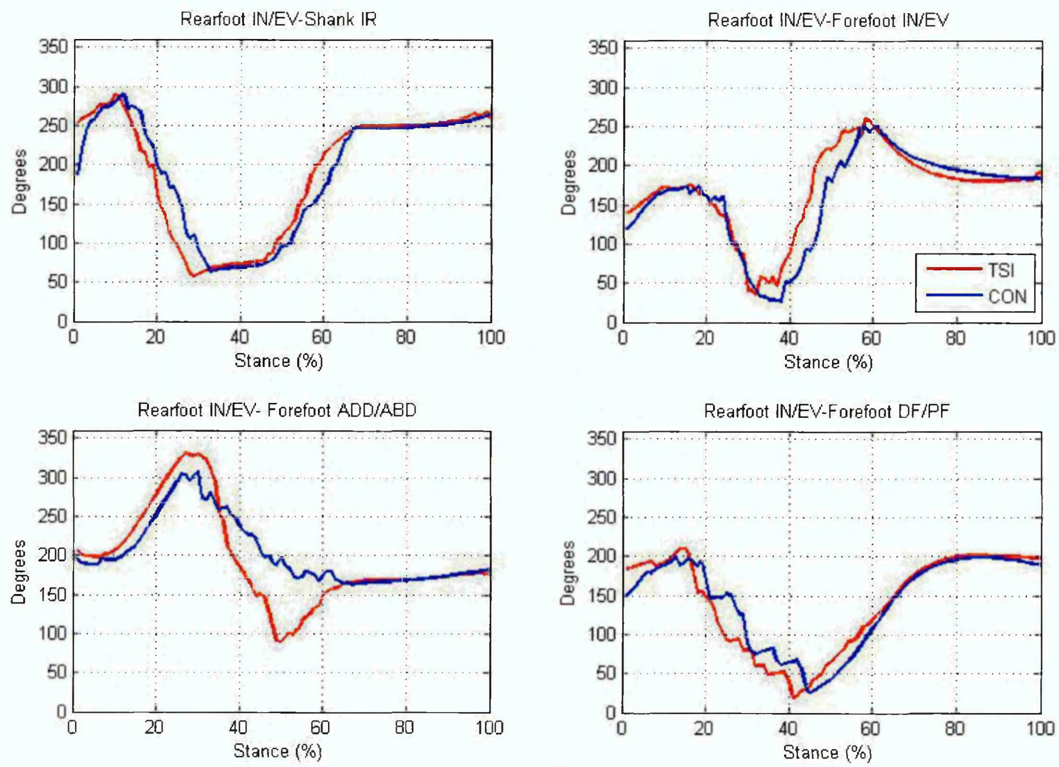


Figure 21. Vector coding curves for the segment couplings of interest in tibial stress injury (TSI) and control (CON) groups. Ensemble means for all subjects are shown throughout stance.

Therefore, greater separation between the joints of the midfoot in those with a history of tibial stress injury was not found. Further exploration of the data revealed a greater forefoot abduction excursion in the tibial stress injury group. These findings suggest that while peak abduction was similar between groups, the forefoot of those with a history of tibial stress injury passes through a greater range of motion during loading.

In addition to a trend towards a difference in peak rearfoot eversion, the present study also identified key differences between groups in rearfoot and shank coupling during early stance. The tibial stress injury group spent a greater percentage of time in a shank phase (tibial stress injury = 36%, control group = 14%) while the control group spent a greater portion of time in a rearfoot phase (tibial stress injury = 13%, control group = 44%) (Table 14a). This suggests that in those with a history of tibial stress injury, greater motion is transferred from the rearfoot to the shank. Therefore, this may result in increased rotation of the shank and potentially altered loading on the tibia. This findings offers partial support to the theory that greater peak rearfoot eversion may be related to increased torsional load on the tibia. In support of increased tibial torsion as an injury mechanism, free moment has been found to be greater in those with a previous tibial stress fracture (Milner *et al.,* 2006a; Pohl *et al.*, 2008). However, any relationship between peak rearfoot eversion and free moment remains speculative. This potential mechanism is though worthy of further study.

A second potential injury mechanism associated with greater rearfoot eversion relates to the phase angle of tibial loading and its relationship to bone fatigue. A difference in phase angle between axial and torsional loading has been reported to have a significant impact on bone fatigue by up to a factor of seven (George and Vashishth, 2005). Significantly more bone fatigue was observed when axial and torsional loading was not synchronised (in phase). The phase angle seen during gait is related to the body's loading response, specifically, the kinematics associated with the stance phase. Overpronation during running has been suggested to alter the phase angle and act as an injury mechanism for stress fracture (George and Vashishth, 2005). Data from this thesis found greater peak rearfoot eversion to be a risk factor for tibial stress injury, this relationship may exist through the mechanism of altered phase angle during loading.

In addition to motion of the rearfoot, differences in forefoot loading may serve to alter the phase angle. While discrete forefoot peak angles were not found to differ between groups, some differences in the coupling of forefoot dorsi/plantarflexion and rearfoot in/eversion were noted between groups. During the first third of stance, the control group spent a larger portion of time moving in-phase (22%) compared to the tibial stress injury group (4%) (Table 14d). This suggests a greater degree of synchrony between rearfoot eversion and forefoot dorsiflexion in the control group. Given the strong coupling between rearfoot in/eversion and forefoot dorsi/plantarflexion (Table 13), differences in this coupling relationship may have resulted in changes in more proximal segments (shank). Therefore, observed differences between rearfoot-shank coupling may in part be related to differences in the relationship between forefoot-rearfoot mechanics.

It was also hypothesised that those with a history of tibial stress injuries would demonstrate higher joint velocities than those in the control group. No differences between groups were seen for both forefoot inversion velocity and rearfoot eversion velocity. Differences in forefoot frontal plane variables may have been difficult to detect given the questionable reliability previously reported for this rotation. However, the lack of difference seen for rearfoot eversion velocity supports the findings of Hetsroni and colleagues (2008), who failed to establish rearfoot eversion velocity as a risk factor for tibial and femoral stress fractures. Hypotheses were partially supported by the finding of greater forefoot abduction velocity and forefoot dorsiflexion velocity in those with a history of tibial stress injury. A greater abduction excursion occurring over a shorter time period in the injury group resulted in a higher peak forefoot abduction velocity compared to the controls. A reduced excursion time associated with forefoot abduction may result in higher average forces during the loading period, thereby inflicting greater stress on the skeletal structures of the lower extremity (Hetsroni *et al.*, 2008).

In addition to dynamic kinematic variables associated with the forefoot and rearfoot, other descriptive variables including static arch height, arch stiffness and tibial shock were measured in injury and control groups. Both static arch height (AHI 90%) and arch stiffness were found to be similar between groups (Table 11). This small scale study support the previous conclusions made in Chapter II, which failed to establish a link between tibial stress injuries and static or quasi static measures of the foot. Also, these data further support the study of dynamic kinematic variables as potentially more relevant to tibial stress injury risk. Greater tibial shock has been found to be a risk factor for tibial stress fracture in both retrospective (Milner *et al.*, 2006b) and prospective studies (Davis *et al.,* 2004). Although a trend was noted (medium effect) towards a higher tibial shock in those with a history of tibial stress injury, data from the present study do not provide strong evidence to support this relationship. However, key differences found in kinematic variables suggest that even in the presence of similar shock, the body's response to loading may predispose it to tibial stress injuries.

Some important limitations to this study should be noted. The study analysed a limited sample of subjects with only six in each group. Recruitment for the study was limited by the strict criteria outlined for inclusion in the injury group. During assessment, extreme care was taken to ensure there was a history of bone related tibial stress injury and to distinguish this from other injuries in the same region, including; medial tibial stress syndrome, compartment syndrome and soft tissue injury. Volunteers, for whom this could not be confidently determined, were not permitted entry into the study. Therefore, while subject numbers were low, it was determined that those included had a similar type of injury; either a tibial stress reaction or tibial stress fracture.

The present study used a process of matching control participants with injured ones based on gender, age and weekly mileage. This approach of matching injury and control participants is one which is consistent with previous retrospective research investigating risk factors for tibial stress injuries (Pohl *et al.*, 2008). However, results revealed that not all control subjects had lower values than tibial stress injury subjects for the variables identified as different. An alternative approach would have been to compare the injury group to a larger group of controls which may better represent the characteristics exhibited by a normal runner. This approach has also been used in previous research where the control group was double the size of the injury group (Creaby and Dixon, 2008). Future use of this sampling method may help to identify the unique differences associated with those at risk of a tibial stress injury compared to the average runner.

The statistical approach in the present study used independent t-tests coupled with a measure of effect size to aid interpretation of meaningful differences between groups. This approach was chosen given the low subject numbers. A greater emphasis in this analysis was placed on the use of effect sizes since they are useful in identifying trends in data. For example, the large effect size observed for forefoot dorsiflexion velocity, suggest a trend towards differences that might be significant in a study with a larger sample. An alternative approach to this analysis may have been the use of binary logistic regression which could be used to assess the combined influence of the predictor variables on the risk of having a tibial stress injury. This data analysis approach has previously been used in larger scale retrospective studies which have sought to identify risk factors associated with tibial stress injuries (Milner *et al.,* 2006b; Pohl *et al.,* 2008)

Finally, the present study employed a retrospective research design. It was assumed that measured variables remained unchanged after the incidence of injury. Some limited evidence does exist to suggest retrospective and prospective studies produce similar results when identifying risk factors relating to tibial stress fracture (Hamill and Davis, 2006). However, further prospective studies are needed to indicate if identified risk factors are present prior to these individuals sustaining a tibial stress injury.

5.5 Conclusions

In summary, these results provide some insight into forefoot-rearfoot kinematics in runners with a history of tibial stress injury. Key differences were found in forefoot-rearfoot kinematics with peak rearfoot eversion, forefoot dorsiflexion velocity and forefoot abduction velocity greater in those with a history of tibial stress injury compared to a control group. Furthermore, key differences between the coupling of the rearfoot-shank and forefoot-rearfoot were noted between groups. These differences appear to represent potential risk factors for tibial stress injuries which are worthy of further investigation. Based on this study, the mechanisms of tibial stress injury related to forefoot-rearfoot kinematics remain speculative.

CHAPTER VI

6 Summary and discussion

Over the past decade, the development of multisegment foot models has allowed for the measurement of foot kinematics during gait. An important application of these foot models has been to investigate clinical pathological conditions. However, multisegment foot models also have an application in the investigation of overuse injuries and dynamic risk factors. Previous *in vivo* research has identified significant motions in the joints of the midfoot during gait. To date, no studies have investigated forefoot-rearfoot kinematics in relation to overuse running injuries. The overall purpose of this thesis was to answer the question, are forefoot-rearfoot kinematics risk factors for tibial stress injuries?

This chapter provides an overall discussion and summary of this thesis and is divided into four sections. Firstly, a brief summary of chapters II, III, IV and V is provided. Secondly, the implications of findings in the thesis are discussed in the context of current and future research in the area of tibial stress injuries. Thirdly, the limitations of the thesis are acknowledged. Finally, a thesis conclusion is presented.

6.1 Summary of individual chapters

6.1.1 Chapter II

This chapter provided a comprehensive review of previous literature relevant to the programme of study. This involved a review of possible tibial stress injury mechanisms and risk factors. The foot was considered in detail and mobility within the joints of the foot highlighted. Methodological considerations associated with the use of multisegment foot models were highlighted with a view to developing a model for use in the series of studies for the programme of research. Forefoot-rearfoot motion has been discussed and the idea of forefoot-rearfoot motions as risk factors for tibial stress injuries introduced.

At present, literature regarding the relationship between foot type and injury is somewhat contradictory. Therefore, a further aim of Chapter II was to establish if any specific association between foot type and tibial stress injuries through a systematic review. Of the 479 initially identified, nine studies were deemed relevant and of sufficient quality for inclusion in the review. The close scrutiny of the included studies failed to provide sufficient evidence to indicate a definitive link between foot type and tibial stress injuries. Some limited evidence was found to support an increased risk associated with extremes of foot type. Importantly, this review highlighted a need for dynamic measures of foot type to investigate the relationship between foot function and tibial stress injury risk.

6.1.2 Chapter III

Given the highlighted need for studies investigating the relationship between dynamic foot characteristics and tibial stress injuries, Chapter III sought to examine methodological considerations associated with the measurement of forefoot-rearfoot kinematics. Specifically, the Chapter sought to develop and present a multisegment foot model for use with gait sandals. Sandals were used because they allow the easy application and tracking of skin markers on the whole foot and offer similar outsoles and midsoles to running footwear. To help support the use of this model in future research, a study to assess the use of gait sandals as a means of analysing forefoot-rearfoot motions when wearing running shoes was conducted. Although sandals have been used as an alternative to running shoes (Branthwaite *et al.*, 2004, Eslami *et al.*, 2007; Morio *et al.*, 2009), no study has compared running kinematics in gait sandals and running shoes to support their use in future research. The findings from Chapter III (section 3.3) revealed rearfoot and shank excursions were similar between sandals and running shoes. Evidence presented justifies the use of sandals as a means of measuring shod kinematics and supports the model presented in section 3. 2 for future studies examining shod running mechanics.

A further aim of Chapter III was to assess the between session reliability of selected kinematic variables associated with the forefoot and rearfoot. Selected kinematic variables focussed on forefoot-rearfoot characteristics which define the motions of the foot during loading, these include peak joint angles, joint excursions and joint velocities. Reliability analysis revealed mixed results with good to excellent reliability reported for frontal plane motions of the rearfoot and triplanar motions of the forefoot. It should be noted that questionable reliability was reported for frontal plane forefoot kinematic variables. In the main, these findings support the use of the present model in future experimental studies.

6.1.3 Chapter IV

Chapter IV sought to present an application of the developed multisegment foot model. No clear association between static foot type and tibial stress injuries was identified in Chapter II. High-arched feet have been suggested to be more rigid and low-arched feet more flexible (Matheson *et al.*, 1987; Simkin *et al.*, 1989). Chapter IV sought to investigate the relationship between dynamic foot function and foot type. Specifically, the study sought to answer the question, do forefoot-rearfoot kinematics differ in high and low-arched individuals? Tibial shock has previously been found to be a risk factor for stress related injuries. Measures of tibial shock were also taken to explore the relationship between forefoot-rearfoot mechanics and shock attenuation.

Static measures of the foot suggested a stiffer arch structure in high-arched individuals and a more flexible arch in low-arched individuals. These data support the previously suggested notion that high-arched feet are relatively stiff and low-arched feet are relatively flexible (Matheson *et al.*, 1987; Simkin *et al.*, 1989). However, a similar relationship was not observed in the present thesis when studying selected dynamic forefoot-rearfoot variables during running. It was hypothesised that a flexible low-arched foot would display greater joint excursions than a stiffer high-arched foot during running. In the main, joint excursions were found to be similar between groups. These data support previous research which has failed to establish a clear link between static foot structure and dynamic foot function (McPoil and Cornwall, 1994; Cavanagh *et al.*, 1997; Nachbauer and Nigg, 1992). Despite this, forefoot abduction excursion, forefoot dorsiflexion excursion and forefoot abduction velocity were found to be greater in high-arched individuals. Differences may represent an increased available range of motion through which joints can pass in those with high arches. Key differences between groups were also noted in the coupling between the forefoot and rearfoot, but these differences did not appear to be transferred proximally to the shank. This application of the developed model has highlighted some differences in dynamic forefoot variables between foot types.

6.1.4 Chapter V

Previous research of dynamic risk factors for tibial stress injuries associated with the foot has focused on rearfoot kinematics (Pohl *et al.*, 2008; Milner *et al.*, 2010). Chapter V sought to investigate forefoot-rearfoot kinematics as potential risk factors in the development of tibial stress injuries. Specifically it sought to answer the research question of the thesis, are forefoot-rearfoot kinematics risk factors for tibial stress injuries? For dynamic forefoot-rearfoot variables, it was hypothesised that those with a history of tibial stress injury would demonstrate higher peak rearfoot and forefoot joint angles than a matched control group. Since some differences in coupling had been identified between foot types, forefoot-rearfoot coupling was also investigated in relation to tibial stress injury risk.

Given the low subject numbers used, results represented some initial findings for further discussion. The tibial stress injury group was found to display a greater peak rearfoot eversion than the control group. This finding supports those of recent larger scale retrospective studies who found greater peak rearfoot eversion as a risk factor for tibial stress fractures in female runners (Pohl *et al.*, 2008; Milner *et al.*, 2010). Investigation of forefoot variables revealed greater peak dorsiflexion velocity and abduction velocity in those with a history of tibial stress injury compared to controls. In addition, key differences were noted for the coupling of the rearfoot and shank, as well as the rearfoot and forefoot. These differences appear to represent potential tibial stress injury risk factors which warrant further investigation.

6.2 Implications of findings

The findings of the present thesis have significant implications in furthering knowledge of this research area. Findings from Chapter II found a lack of relationship between foot type and tibial stress injuries. Further, it highlighted few studies which attempt to use dynamic measures of foot type to investigate the relationship between foot function and tibial stress injury risk. It is hoped that these findings will result in a greater focus on dynamic foot function rather than static measures in future research.

Multisegment foot modelling is becoming increasingly common place in routine biomechanical analysis. The application of such models to the study of running injuries is an important step in better understanding risk factors associated with the foot. To date, this thesis represents the first time a multisegment foot model has been used to investigate risk factors associated with overuse injuries. It is hoped that the methods used will be adopted by other researchers to further explore this area of research. Specifically, data in Chapter III supports the use of gait sandals in future studies of running. The justification for their use has important implications for future work in furthering our understanding of dynamic foot function. Given the relative ease with which gait sandals can be used in conjunction with a two segment foot, it is hoped this type of analysis will become commonplace when analysing the foot during gait.

The findings of Chapter IV provide insight into the complex nature of the interaction between structure and function in the foot. An understanding of this interaction has clinical implications and may allow future development of injury prevention strategies specific to foot type. Key differences were found for the transverse plane variables of forefoot abduction excursion and forefoot abduction velocity. Higher forefoot abduction excursion and forefoot abduction velocity were found in high-arched compared to low-arched individuals. These findings appear to characterise the differences in the loading response of high and low-arched feet. Furthermore, these observations serve to highlight the importance of exploring forefoot loading patterns in specific relation to injury risk.

The findings of Chapter V have significant implications on our understanding of risk factors associated with tibial stress injuries. A greater peak rearfoot eversion was found in those with tibial stress injuries. This finding lends support to recent literature which implicates peak rearfoot eversion as a strong risk factor for tibial stress injury (Pohl *et al.*, 2008; Milner *et al.*, 2010). However, further insight has also been provided in this thesis with the inclusion of forefoot motions. New risk factors associated with the forefoot have been identified, these include, forefoot dorsiflexion velocity and forefoot abduction velocity. This finding supports the notion that those with a history of tibial stress injury display a unique forefoot loading pattern during stance.

In both Chapters IV and V, transverse plane forefoot motions were found to be significant joint rotations. Specifically, both high-arched individuals and those with a history of tibial stress injury were found to have greater forefoot abduction excursion and higher forefoot abduction velocities. Given that these variables have been found to be risk factors for tibial stress injury (Chapter V), a high-arched foot may represent a risk factor for this type of injury. The forefoot loading pattern seen in those with high-arches may predispose them to an increased risk of tibial stress injury. Indeed, this finding supports previous research which has found high-arched individuals are at an increased risk of shock related injuries such as stress fractures (Sullivan *et al.*, 1984; Giladi *et al.*, 1985; Matheson *et al.*, 1987; Williams *et al.*, 2001a). However, since the high-arched population studied in this thesis were not injured, any specific relationship between these variables and injury risk remains speculative.

Work from this thesis represents an initial exploration of forefoot-rearfoot kinematics. Findings have highlighted the importance of forefoot motions in relation to injury. This is significant for future work attempting to development specific injury prevention strategies. The manufacture of running shoes has predominantly focused on rearfoot motion control and seeking to limit rearfoot eversion in the frontal plane. Future work should give more consideration to the forefoot and should consider triplanar motions of the foot, particularly the transverse plane. In addition to footwear, the present thesis may have future implications regarding running technique. All studies examined forefoot-rearfoot kinematics in rearfoot strikers. It is likely that risk factors are modified with different styles of running and the coupling between segments changed. While forefoot risk factors were identified in rearfoot strikers, a forefoot strike pattern may place greater emphasis on the kinematics of the forefoot. Therefore, subject specific footwear which considers structure and function of the foot, as well as running style may become the norm in attempting to minimise the risk of tibial stress injuries.

6.3 Future directions

The results of the present thesis provide a basis for future research in this area of biomechanics. It is clear that more detailed multisegment foot models are needed to quantify the discrete movements of the joints in the foot within various footwear conditions. However, future studies using skin mounted markers will remain constrained by the errors associated with skin marker movement on the foot. Future imaging technologies may provide a detailed means of assessing foot motion (Nester, 2009). However, at present dynamic imaging methods do not offer the detail or sufficient sampling rates to make their use valid.

Future challenges are also concerned with the measurement of dynamic forefoot motion within running shoes, while seeking to maintain the integrity of the footwear. Chapter III (section 3.3) presents data to support the use of gait sandals in replicating the rearfoot motions seen when wearing running shoes. At present, no data exists on forefoot motion within running shoes. Future studies should seek to establish if gait sandals may be used as a means of replicating the forefoot motions seen when wearing running shoes.

Additionally, a more detailed investigation of the research question formulated in Chapter V is warranted. A study of greater statistical power is required to explore these selected kinematic variables. This increased power would allow significant detection of any clinically meaningful differences for the variables of interest and would allow firm conclusions to be drawn regarding the role of forefoot-rearfoot kinematics and tibial stress injury risk.

The dependent variables explored in the present thesis represent a small portion of those which may be important in identifying risk factors associated with the foot. These variables were selected as ones which characterised the motions of the foot during loading. Given the complex nature of the interactions of the tissues in the foot when loaded, other potentially important variables should be explored. This may include kinetic as well as further kinematic variables. In addition, further investigation of the coupling between foot segments should be conducted.

The present thesis has provided some initial evidence of forefoot abduction velocity as a risk factor for tibial stress injuries. This suggests that those with tibial stress injuries may display unique forefoot loading patterns during stance. A greater understanding of these loading patterns may be found through the study of the relationship between forefoot kinematics and ground reaction forces. In particular, the relationship between forefoot abduction variables and free moment warrants investigation. Free moment represents the torque about the vertical axis due to friction between the foot and the ground during stance. Free moment has previously been found to be a risk factor for developing tibial stress injuries (Milner *et al.*, 2006a; Pohl *et al.*, 2008). Milner et al. (2006a) found absolute free moment and peak adduction free moment were significantly higher in female runners with a history of tibial stress fracture than in a control group. Adduction free moment acts to resist toeing out. Chapter V of this thesis reported findings to suggest a greater forefoot abduction excursion and peak abduction velocity in those with a history of tibial stress injury. Throughout the loading phase of stance, the forefoot abducts with respect to the rearfoot. Greater abduction excursion and abduction velocity may result in a larger free moment resisting (adduction free moment acting on the runner) this action during loading. However, given that the present thesis did not investigate free moment, this relationship remains purely speculative. Future studies should explore the relationship between free moment and transverse plane kinematics of the forefoot.

The results of the present thesis represent initial exploration of the dynamic forefoot-rearfoot motions as risk factors for tibial stress injury. Future work in this area will improve our understanding of potential injury mechanisms associated with the foot. An improved understanding of these mechanisms will assist in the early identification of those individuals at risk and help to develop preventative interventions and rehabilitation strategies.

6.4 Limitations

There are various limitations which could have influenced the results of the included studies. Firstly, the multisegment foot model developed and employed in Chapters IV and V consisted of only two segments (forefoot-rearfoot). This meant that motion of the midfoot joints was modelled as the forefoot segment relative to the rearfoot. As previously noted (section 2.1.6.2), significant motions are present in numerous joints of the midfoot. Furthermore, contradictory evidence has been presented concerning the direction of these rotations at specific joints. The present model does not discriminate movements that occurred at different joints within the midfoot. Therefore, caution should be exercised when attempting to use the present findings to infer movement about specific joints of the foot.

A further limitation of the model relates to the assumption of rigid segments. At the forefoot, this assumes that all five metatarsals are acting together as a single rigid segment. However, as previously discussed, significant motion between the metatarsals has been reported (section 2.1.6.2). Okita et al. (2009) reported a maximum error of 4.1º associated with the assumption of a five metatarsal forefoot segment. In the forefoot, the first three metatarsals appear to function as a stable unit, while the fourth and fifth metatarsals function as a separate unit (Lundgren *et al.*, 2008). Furthermore, when assessing errors associated with rigid body assumptions, recent evidence supports the modelling of the forefoot as two segments (Metatarsal 1-3 and metatarsals 4,5) (Nester *et al.*, 2010). Modelling the forefoot as more than one segment was beyond the scope of the present thesis. Therefore, error associated with this assumption should be realised when interpreting findings of this work.

Limitations specific to Chapter V have previously been outlined (section 5.4). However, certain of these are noteworthy of further discussion given their importance in relation to the overall findings of the thesis. Chapter V analysed only a limited sample of subjects (n = 6 in each group). Therefore results of this study represent only initial findings and caution should be exercised when drawing conclusions regarding risk factors for tibial stress injuries. Furthermore, the sample examined comprised both males and females (4 males and 2 females). Research to date investigating tibial stress fractures has focused mainly on females given the increased injury risk associated with this population (Milner *et al.*, 2006ab, 2007, 2010, Pohl *et al.*, 2008). Previous research has indicated that female runners exhibit different frontal and transverse plane mechanics at the hip and knee compared with males (Ferber *et al.*, 2003). Therefore, similar differences may be expected for gait mechanics at more distal joints including the ankle and joints of the foot. Given these expected gender differences, injury mechanism associated with tibial stress injuries may also differ between males and females.

Finally, the research study in Chapter V employed a retrospective design. Some initial evidence exists to suggest retrospective and prospective studies produce similar results when risk factors to tibial stress fracture (Hamill and Davis, 2006). However, it is widely acknowledged that large scale prospective studies form the gold standard for identification of overuse injury risk factors. Despite this, challenges with this type of approach prevail. The measurement and processing of complex biomechanical variables on large cohorts of participants is both time consuming and costly. Future advances in the capture and processing of dynamic variables may reduce data collection times and make large scale prospective studies viable in the future.

6.5 Conclusion

The purpose of this thesis was to investigate forefoot-rearfoot motion in relation to tibial stress injuries. To investigate this aim, four studies were reported in the present thesis. A summary of key findings within these studies are outlined in sections 6.1.1, 6.1.2, 6.1.3, 6.1.4. Findings of this thesis present some initial data to suggest forefoot-rearfoot kinematics may be important risk factors in the development of tibial stress injuries. In particular, peak rearfoot eversion, forefoot dorsiflexion velocity and forefoot abduction velocity were highlighted as a key during loading. However, further larger scale studies are required to confirm these findings. Included chapters have also further highlighted the multifactorial nature of running injuries. Thesis findings have helped further understanding of a relatively new area in biomechanics. It is hoped that the results presented in this thesis will be a useful basis for future studies investigating foot characteristics as risk factors for tibial stress injuries.

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Appendix A

Quality assessment scoring scheme

Name of Study/Author...

Item	Criteria	Classification	Score
A	Were the inclusion/exclusion criteria clearly defined?	3= Yes clearly/ well defined 2= Adequately defined 1= Poorly/inadequately defined 0= Not defined at all.	
B	Were there sufficient subject numbers included?	3= excellent subject numbers/large cohort. 2= Reasonable subject numbers on which good quality statistics could be based. 1= Low subject numbers/inadequate power	
C	What was the quality of the research design?	3= Longitudinal Prospective cohort study (Surveillance for a period of 12 months or longer). 2= Prospective study (Surveillance for at least a period of 3 months, but less than 12 months) 1= The prospective study had a surveillance period of less than 3 months or the study is retrospective	
D	Comparability of injury and control groups?	3= unconfounded/ good comparability or matching of groups/ confounding factors adjusted for. 2= confounding small/comparability reported in text without supporting data. 1= No matching of groups/confounding factors mentioned but not adjusted for. 0= not discussed	
E	The methods used to diagnose Injury	3=Tibial stress fracture/ reaction diagnosed using radiological methods (or other comparable imaging techniques). 2= Tibial stress fracture/reaction diagnosed by experienced clinician through physical exam. 1= self diagnosis/through injury history questionnaire. 0= unknown	
F	The methods used to assess medial longitudinal arch height.	3=Validated quantitative methods for measuring arch height (i.e. x-ray) 2= Reliable quantitative methods for measuring arch height (i.e. navicular drop)	

		1= Subjective assessment and classification of foot type.	
G	Were multiple comparisons and subgroup analyses carried out?	3= good quality statistical analysis/multiple comparisons 2= adequate statistical methods were employed for comparisons 1= Basic data analysis methods (i.e. multiple t-tests) 0= None carried out or discussed	
		Total Score	

Appendix B

Data Extraction form for assessing the relationship between foot type and risk of tibial stress injuries.

General Information

Date of Extraction	
Name of Reviewer	
Author Surname	
Year of Publication	

Methods

Participants

Recruitment period	
Subject numbers	
Gender of subjects	
Study setting/population	
Subject demographics	
Subject inclusion/exclusion criteria	

Study Design

The research design (quality)	
The follow up period	
Comparability of groups	
Statistical Methods	

Procedures

The diagnosis/definition of Injury	
Classification of arch height/foot type	

Results

Drop outs	
Injury Incidence	
Foot classification	
Correlation/risk factor between factors	
Conclusion	

Comments

General comments/ Notes	

Appendix C

Chapter III

Participant information sheet and informed consent forms

Sheffield Hallam University

Faculty of Health and Wellbeing Research Ethics Committee

Sport and Exercise Research Ethics Review Group

Participant Information Sheet

Project Title	The use of gait sandals for measuring rearfoot and shar motion during running.
Supervisor/Director of Studies	Professor Ian Maynard
Principal Investigator	Andrew Barnes
Telephone number	07773239263

Purpose of Study and Brief Description of Procedures
(Not a legal explanation but a simple statement)

The purpose of the study is to examine the suitability of gait sandals for routine biomechanical analysis of running. To be suitable, you must not have a history of surgery on your legs/feet and currently be injury free. If you are suitable and willing to participate, you will be required to visit the biomechanics laboratory at Sheffield Hallam University for the testing procedure. You will be required to visit the biomechanics lab once, with all testing taking approximately 90 minutes. When you arrive at the lab, you will be asked to change into suitable sports clothes (Shorts and t shirt) Note: you will not need to bring any footwear, however, you must be either a size 7,8, 9 or 10 shoe to participate in this study. Reflective markers will be attached to your right lower leg and foot while wearing running shoes. Subsequently, you will be familiarised with running down the lab within given speed constraints, and contacting a force plate which is mounted within the runaway. After practicing, you will be required to complete a series of short runs (approx 15 meters, 10 repeat trials), contacting the centre of the force plate (the running in this study is not strenuous and is a relatively intermediate pace). This process will be repeated when running barefoot and also when wearing gait sandals. You may experience some minor discomfort associated with the application and removal of markers and sensors to the skin. If you have any questions I will be happy to answer them before you agree to participate. You are under no obligation to participate in the study and you are free to withdraw at any time, without r explanation. All data collected will be treated confidentially and your name will not be identified in any reports or publications resulting from this study.

It has been made clear to me that, should I feel that these Regulations are being infringed or that my interests are otherwise being ignored, neglected or denied, I should inform Professor Edward Winter, Chair of the Faculty of Health and Wellbeing Research Ethics Committee (Tel: 0114 225 4333) who will undertake to investigate my complaint.

Sheffield Hallam University

Faculty of Health and Wellbeing Research Ethics Committee

Sport and Exercise Research Ethics Review Group

INFORMED CONSENT FORM

TITLE OF PROJECT: The use of gait sandals for measuring rearfoot and shank motion during running.

The participant should complete the whole of this sheet himself/herself

Have you read the Participant Information Sheet?	YES/NO
Have you had an opportunity to ask questions and discuss this study?	YES/NO
Have you received satisfactory answers to all of your questions?	YES/NO
Have you received enough information about the study?	YES/NO
To whom have you spoken? .... Do you understand that you are free to withdraw from the study: • at any time • without having to give a reason for withdrawing • and without affecting your future medical care	YES/NO
Have you had sufficient time to consider the nature of this project?	YES/NO
Do you agree to take part in this study?	YES/NO

Signed ....   Date ....

(NAME IN BLOCK LETTERS)...

Sheffield Hallam University

Faculty of Health and Wellbeing Research Ethics Committee

Sport and Exercise Research Ethics Review Group

Participant Information Sheet

Project Title	Reliability of selected kinematic variables and peak tibi shock during running
Supervisor/Director of Studies	Professor Ian Maynard
Principal Investigator	Andrew Barnes
Telephone number	07773239263

Purpose of Study and Brief Description of Procedures

The purpose of the study is to examine the reliability of selected biomechanical measures between testing session. To be suitable, you must not have a history of surgery on your legs/feet and currently be injury free. If you are willing to participate, you will be required to visit Sheffield Hallam biomechanics on two occasions, approximately one week apart. Each testing session should take approximately one hour When you arrive at the lab, you will be asked to change into suitable sports clothes (Shorts and t shirt) Note: you will not need to bring any footwear, however, you must be either a size 7,8, 9 or 10 shoe to participate in this study. 18 reflective markers will be attached to your lower leg and foot, as well as a small accelerometer which will be stuck to the front section of your lower leg. Subsequently, you will be familiarised with running down the lab within given speed constraints, and contacting a force plate which is mounted within the runaway. After practicing, you will be required to complete a series of short runs (approx 15 meters, 10 repeat trials), contacting the centre of the force plate (the running in this study is not strenuous and is a relatively intermediate pace). You may experience some minor discomfort associated with the application and removal of markers and sensors to the skin. If you have any questions I will be happy to answer them before you agree to participate. You are under no obligation to participate in the study and you are free to withdraw at any time, without r explanation. All data collected will be treated confidentially and you will not be identified in any reports or publications resulting from this study.

It has been made clear to me that, should I feel that these Regulations are being infringed or that my interests are otherwise being ignored, neglected or denied, I should inform Professor Edward Winter, Chair of the Faculty of Health and Wellbeing Research Ethics Committee (Tel: 0114 225 4333) who will undertake to investigate my complaint.

Sheffield Hallam University

Faculty of Health and Wellbeing Research Ethics Committee

Sport and Exercise Research Ethics Review Group

INFORMED CONSENT FORM

TITLE OF PROJECT: Reliability of selected kinematic variables and peak tibial shock during running

The participant should complete the whole of this sheet himself/herself

Have you read the Participant Information Sheet?	YES/NO
Have you had an opportunity to ask questions and discuss this study?	YES/NO
Have you received satisfactory answers to all of your questions?	YES/NO
Have you received enough information about the study?	YES/NO
To whom have you spoken? .... Do you understand that you are free to withdraw from the study: • at any time • without having to give a reason for withdrawing • and without affecting your future medical care	YES/NO
Have you had sufficient time to consider the nature of this project?	YES/NO
Do you agree to take part in this study?	YES/NO

Signed .... Date ....

(NAME IN BLOCK LETTERS)..

Appendix D

Chapter IV

Participant information sheet and informed consent forms

Sheffield Hallam University

Faculty of Health and Wellbeing Research Ethics Committee

Sport and Exercise Research Ethics Review Group

Participant Information Sheet

Project Title	Forefoot-rearfoot kinematics in high and low arched individuals.
Supervisor/Director of Studies	Professor Ian Maynard
Principal Investigator	Andrew Barnes
Telephone number	07773239263

Purpose of Study and Brief Description of Procedures

The purpose of the study is to examine dynamic foot characteristics in those with different types of feet. We are interested in the structure of your foot and how it moves when running. An initial assessment will involve the measurement of height and weight as well as several measurements on your right foot, in both standing and sitting positions. The whole procedure should take no more than 10 minutes. As a result of these measures, you may be contacted again via email and asked if you would like to participate in a biomechanical analysis.

You will be required to visit the biomechanics lab once, with all testing taking approximately one hour. When you arrive at the lab, you will be asked to change into suitable sports clothes (Shorts and t shirt) Note: you will not need to bring any footwear, however, you must be either a size 7,8, 9 or 10 shoe to participate in this study. 18 reflective markers will be attached to your lower leg and foot, as well as a small accelerometer which will be stuck to the front section of your lower leg. Subsequently, you will be familiarised with running down the lab within given speed constraints, and contacting a force plate which is mounted within the runaway. After practicing, you will be required to complete a series of short runs (approx 15 meters, 10 repeat trials), contacting the centre of the force plate (the running in this study is not strenuous and is a relatively intermediate pace). You may experience some minor discomfort associated with the application and removal of markers and sensors to the skin. If you have any questions I will be happy to answer them before you agree to participate. You are under no obligation to participate in the study and you are free to withdraw at any time, without r explanation. All data collected will be treated confidentially and your name will not be identified in any reports or publications resulting from this study.

It has been made clear to me that, should I feel that these Regulations are being infringed or that my interests are otherwise being ignored, neglected or denied, I should inform Professor Edward Winter, Chair of the Faculty of Health and Wellbeing Research Ethics Committee (Tel: 0114 225 4333) who will undertake to investigate my complaint.

Sheffield Hallam University

Faculty of Health and Wellbeing Research Ethics Committee

Sport and Exercise Research Ethics Review Group

INFORMED CONSENT FORM

TITLE OF PROJECT: Forefoot-rearfoot kinematics in high and low arched individuals.

The participant should complete the whole of this sheet himself/herself

Have you read the Participant Information Sheet?	YES/NO
Have you had an opportunity to ask questions and discuss this study?	YES/NO
Have you received satisfactory answers to all of your questions?	YES/NO
Have you received enough information about the study?	YES/NO
To whom have you spoken? .... Do you understand that you are free to withdraw from the study: • at any time • without having to give a reason for withdrawing • and without affecting your future medical care	YES/NO
Have you had sufficient time to consider the nature of this project?	YES/NO
Do you agree to take part in this study?	YES/NO

Signed .... Date ....

(NAME IN BLOCK LETTERS)...

Appendix E

Relationship between static arch height and arch stiffness

Introduction

Foot type and particularly arch height is an intrinsic injury risk factor which has received much attention within the literature (Cowan *et al.*, 1993). However, the relationship between foot type and injury is somewhat unclear. In addition to arch height, arch stiffness may be important in trying to relate foot type to injury. It is commonly thought that high arches tend to be more rigid, and lower arches more flexible. Furthermore, it is suggested that more flexible feet with lower arches may serve as more effective natural shock absorbers than more rigid foot types. A higher incidence of shock related bony injuries has been reported in high arched runners (Williams *et al.,* 2001). Zifchock et al. (2006) measured the foot structure of 145 individuals in both 10% and 50% of weight bearing, with relative arch deformation between conditions used as a means of assessing arch stiffness. Although a relationship between variables was observed, only 9% of the variance in arch stiffness could be explained by arch height (Zifchock *et al.*, 2006). During shod running, vertical ground reaction forces can reach 2-3 times body weight. Assessing arch characteristics in loaded conditions closer to those seen during running may provide insight into dynamic arch stiffness and its potential relationship to injury. The aim of the present study was to assess the relationship between arch height and arch stiffness in 10% and 90% of weight bearing. It is suggested that calculating arch stiffness between these conditions may give a better indication of foot function in loaded conditions, such as gait.

Methods

After ethics approval, 101 male participants (age 20.0±2.8, height 176±19 cm, mass 76.9±10.5 kg) gave informed consent to take part in the investigation. Measurements were taken on the right foot of participants in 10% and 90% of weight bearing, using a previously validated measurement system (Williams and McClay, 2000). A portable force plate was used to determine percentage weight bearing, with the 10% condition measured in a sitting and the 90% in a standing position. Dorsum height at 50% of foot length divided by truncated foot length was used as a measure of arch height index (AHI). Relative arch deformation between 10% and 90% of weight bearing was calculated using the equation described by Williams and McClay (2000). Relative arch deformation was normalised to body weight (BW) to give a measure of arch stiffness, with a lower arch stiffness score indicating a flexible arch structure, and a higher score indicating a more rigid arch. A one-tailed Pearson's correlation was used to examine the relationship between each AHI measure and arch stiffness. Significance was accepted at (p<0.05).

Results

Based on normality analysis, arch stiffness data were found to be positively skewed. A log transform ($\log_{10}$) was applied to all data, resulting in normal distribution of all variables. The relationship between arch stiffness and AHI in 10% and 90% weight bearing can be seen in Figures E.1 and E.2 respectively. No significant relationship was seen between AHI in 10% of weight bearing and arch stiffness (R^2= 0.024, p = 0.063). A relationship was observed between AHI in 90% of weight bearing and arch stiffness, although it was a relatively weak one (R^2=0.14, p= 0.0001). The observed trend suggests that a higher arch equates to a more rigid foot, and a lower arch a more flexible foot.

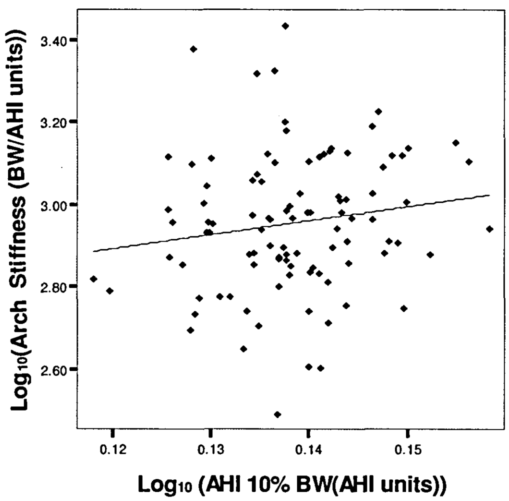

Figure E.1. Relationship between AHI 10% and arch stiffness (R^2= 0.024, p = 0.063).

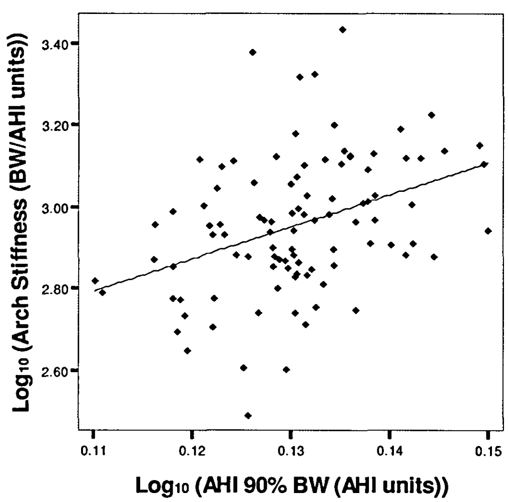

Figure E.2. Relationship between AHI 90% and arch stiffness (R^2=0.14, p= 0.0001).

Discussion

The findings of the present study suggest no relationship between AHI 10% and arch stiffness, and only a weak relationship between AHI 90% and arch stiffness. In support of the hypothesis regarding height and stiffness, a higher arch was suggestive of a more rigid foot, and a lower arch a more flexible foot. However, only 14% of the variance seen in arch stiffness can be explained by arch height. These findings do suggest a marginally higher association than the 9% variance previously reported (Zifchock *et al.,* 2006). Such differences may be due to the population measured, or the use of AHI at 90% rather than 50% weight bearing. Although 90% AHI represented an increased degree of loading, it was still below the level that would be expected during running. Evidence from this study and the study by Zifchcock and colleagues (2006) suggests static measures of arch height offer a limited indication of arch stiffness characteristics. Given the small amount of variance explained by arch height, other factors must be of importance in determining foot stiffness. Such factors might include foot mobility and the range of motion within the joints of the foot, particularly the midfoot. Significant rotations about the joints of the midfoot have been reported during walking (Lundgren *et al.,* 2008). The degree which bone and soft tissue structures permit motion between relative joints as they are loaded is likely a crucial factor in determining arch stiffness. Future research should consider the three dimensional motions at the joints of the foot as it is loaded. Analysis of dynamic situations such as running should be a priority, in an attempt to establish links between foot stiffness and injury.

Conclusion

A relationship between arch height and arch stiffness was only observed using AHI at 90% of weight bearing. Although this measure does provide some indication of arch stiffness, it is limited. The use of static measures may not reflect the dynamic function of the foot.

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Appendix F

Chapter V

Participant information sheet and informed consent forms

Sheffield Hallam University

Faculty of Health and Wellbeing Research Ethics Committee
Sport and Exercise Research Ethics Review Group
Participant Information Sheet

Project Title	Forefoot-rearfoot kinematics in recreational runners with a history of tibial stress fracture
Supervisor/Director of Studies	Professor Ian Maynard
Principal Investigator	Andrew Barnes
Telephone number	07773239263

Purpose of Study and Brief Description of Procedures

The purpose of the study is to examine dynamic foot characteristics in those who have suffered a tibial stress injury. We are interested in the structure of your foot and how it moves when running. The study will help to further our understanding of injury risk factors associated with stress injuries, and whether different foot types could predispose runners to risk of these injuries. In order to assess your suitability for the study, you will asked various questions concerning your injury history and physical activity profiles. If you are suitable and willing to participate, you will be required to visit the biomechanics laboratory at Sheffield Hallam University for the testing procedure. You will be required to visit the biomechanics lab once, with all testing taking approximately one hour. When you arrive at the lab, you will be asked to change into suitable sports clothes (Shorts and t shirt) Note: you will not need to bring any footwear, as this is provided. 18 reflective markers will be attached to your lower leg and foot, as well as a small accelerometer which will be stuck to the front section of your lower leg. Subsequently, you will be familiarised with running down the lab within given speed constraints, and contacting a force plate which is mounted within the runaway. After practicing, you will be required to complete a series of short runs (approx 15 meters, 10 repeat trials), contacting the centre of the force plate (the running in this study is not strenuous and is a relatively intermediate pace). You may experience some minor discomfort associated with the application and removal of markers and sensors to the skin. If you have any questions I will be happy to answer them before you agree to participate. You are under no obligation to participate in the study and you are free to withdraw at any time, without r explanation. All data collected will be treated confidentially and your name will not be identified in any reports or publications resulting from this study.

It has been made clear to me that, should I feel that these Regulations are being infringed or that my interests are otherwise being ignored, neglected or denied, I should inform Professor Edward Winter, Chair of the Faculty of Health and Wellbeing Research Ethics Committee (Tel: 0114 225 4333) who will undertake to investigate my complaint.

Sheffield Hallam University

Faculty of Health and Wellbeing Research Ethics Committee

Sport and Exercise Research Ethics Review Group

INFORMED CONSENT FORM

TITLE OF PROJECT: Forefoot-rearfoot kinematics in recreational runners with history of tibial stress fracture.

The participant should complete the whole of this sheet himself/herself

Have you read the Participant Information Sheet?	YES/NO
Have you had an opportunity to ask questions and discuss this study?	YES/NO
Have you received satisfactory answers to all of your questions?	YES/NO
Have you received enough information about the study?	YES/NO
To whom have you spoken? .... Do you understand that you are free to withdraw from the study: • at any time • without having to give a reason for withdrawing • and without affecting your future medical care	YES/NO
Have you had sufficient time to consider the nature of this project?	YES/NO
Do you agree to take part in this study?	YES/NO

Signed .... Date ....

(NAME IN BLOCK LETTERS)..