

Biomechanical variables associated with tibial and third metatarsal stress fractures in Royal Marines recruits

Submitted by Michael Parnell levers Nunns to the University of Exeter as a thesis for the degree of Doctor of Philosophy in Sport and Health Sciences in September 2013.

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ABSTRACT

Due to their prevalence and associated high rehabilitation costs, this thesis aimed to better understand factors influencing the risk of tibial (TSF) and third metatarsal (MT3SF) stress fractures in Royal Marine recruit training. In Study 1, the standard issue combat assault boot and neutral trainer were assessed during running. Running in the boot caused restricted ankle motion, greater forefoot loading, greater ankle stiffness and a more laterally applied horizontal force vector at the instant of peak braking, suggesting that the risk of incurring MT3SF was greater in this condition.

In Study 2, bending stresses were modelled along the length of the third metatarsal of five participants, using individual bone geometry and dynamic gait data. Stresses were modelled for running when barefoot, and when shod in the standard issue footwear. Estimated peak bending stresses were significantly greater in the combat assault boot than the gym trainer, predominantly due to increased plantar loading. Individual bone geometry was however dominant in determining peak bending stresses.

In Study 3, a large (n=1065) prospective study was conducted to identify differences in baseline characteristics between recruits sustaining a TSF or MT3SF and those who complete training uninjured. Ten TSF and 14 MT3SF cases were compared to 120 uninjured legs. Results suggest that risk of TSF is greater in those recruits with reduced ability to resist loading and attenuate impact during gait. Results for MT3SF suggest that ankle and foot position at touchdown, and the timing and magnitude of forefoot loading, are important factors influencing risk of this injury. The observation of lower age and BMI in

both stress fracture groups was linked to lower bone strength and earlier fatigue mechanisms.

This thesis has increased the understanding of MT3SF in particular, and provides information on specific factors which may be associated with MT3SF and TSF in RM recruits during basic training.

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It is fair to say that this PhD has demanded much in the 6 years I have spent working towards its production. In addition to my own efforts, I have benefitted from the work and support of many individuals. It is therefore a difficult task to ensure that they all receive the credit they deserve, and I apologise now for the inevitable insufficiency with which I am about to credit these people.

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LIST OF ABBREVIATIONS, ACRONYMS AND SYMBOLS USED

| Abbreviation/acronym | Definition |
|-------------------------|---|
| σ | Stress |
| σ_{ax} | Axial stress |
| σ_{be} | Bending stress |
| σ_c | Peak compressive stress |
| σ_t | Peak tensile stress |
| σ_{tor} | Peak torsional stress |
| A | Area |
| ADF | Ankle dorsiflexion |
| BF | Barefoot |
| BMD | Bone mineral density |
| BMI | Body mass index |
| BW | Bodyweight |
| CA | Cross-sectional area |
| CAB | Combat assault boot |
| COP | Centre of pressure |
| COP _Y | Anterior-posterior location of the centre of pressure |
| CT | Computerised tomography |
| CTCRM | Commando Training Centre, Royal Marines Lympstone |
| CV% | Coefficient of variance |
| Deg | Degrees |
| DEXA | Dual-energy x-ray absorptiometry |
| DFP | The dorsiflexion phase |
| DLT | Direct linear transformation |
| EHR | External hip rotation |
| EMG | Electromyography |
| EVA | Ethylene-vinyl acetate |
| F | Force |
| F _H | Resultant horizontal force |
| F θ _H | Angle of application of the horizontal force vector |
| FFCP | Forefoot contact phase |
| F _x | Ground reaction force in the medial-lateral direction |
| F _y | Ground reaction force in the anterior-posterior direction |
| F _z | Ground reaction force in the vertical direction |
| GRF | Ground reaction force |
| GT | Gym trainer |
| HL | Lateral heel |
| HM | Medial heel |
| HROM | Passive range of internal-external rotation at the hip joint |
| Hz | Hertz |
| ICC | Intra-class correlation coefficient |
| IHR | Internal hip rotation |
| ISAK | International Society for the Advancement of Kinanthropometry |
| I _x | Area moment of inertia about the x-axis |
| I _y | Area moment of inertia about the y-axis |
| I _z | Polar moment of inertia about the z-axis |
| M1-5 | Metatarsal head 1-5 |

| | |
|--------------------|--|
| M_{Be} | Bending moment |
| MF | Midfoot |
| MoD | Ministry of Defence |
| MRI | Magnetic resonance imaging |
| MT | Metatarsal |
| MT3 | Third metatarsal |
| MT3SF | Third metatarsal stress fracture |
| MTP | Metatarsophalangeal joint |
| MTP1-5 | First to fifth metatarsophalangeal joint |
| PADF | Passive ankle dorsiflexion |
| $PCOP_Y$ | Mean position of centre of pressure y-coordinate during stance, expressed as a percentage of truncated foot length |
| PF | The plantarflexion phase |
| R | The radius of the outer surface of a cross-section of bone |
| RM | Royal Marine |
| SD | Standard deviation |
| T1 | Hallux |
| T2-5 | Lesser toes |
| $TCOP_Y$ | The time at which the centre of pressure y-coordinate first became level with the metatarsophalangeal joint, in the anterior-posterior direction |
| TSF | Tibial stress fracture |
| $VO_2 \text{ max}$ | Maximum rate of oxygen consumption |
| y | Maximum distance of the cross-section from the centroid |

LIST OF EQUATIONS

- Direct Linear Transformation:

$$\begin{aligned}x_{ij} &= \frac{a_{1j}x_i + a_{2j}y_i + a_{3j}z_i + a_{4j}}{a_{9j}x_i + a_{10j}y_i + a_{11j}z_i + 1} \\y_{ij} &= \frac{a_{5j}x_i + a_{6j}y_i + a_{7j}z_i + a_{8j}}{a_{9j}x_i + a_{10j}y_i + a_{11j}z_i + 1}\end{aligned}\quad \text{Eq. 2.1}$$

Where x_i, y_i, z_i are the desired three-dimensional co-ordinates of point i ; x_{ij}, y_{ij} are the two dimensional co-ordinates of point i in object space, viewed by camera j ; a_{1j} to a_{11j} are constants for camera j , where each camera has 11 unique constants.

- Resultant horizontal force (F_H):

$$F_H = \sqrt{(F_x^2) + (F_y^2)} \quad \text{Eq. 3.1}$$

Where F_x is medio-lateral force and F_y is antero-posterior force.

- The angle of the application of F_H :

$$F\theta_H = \tan^{-1}(F_x \div F_y) \quad \text{Eq. 3.2}$$

- Heron's formula:

$$\text{Area of triangle} = \sqrt{(s(s-a)(s-b)(s-c))} \quad \text{Eq. 4.1}$$

Where s is half the perimeter of a triangle with sides a , b , c .

- Bending moment:

$$M_{be} = F(L - x) \quad \text{Eq. 4.2}$$

Where F is the applied force, derived from the vertical force time history acquired by plantar pressure analysis; L is the length of the metatarsal and x is the perpendicular distance from the section to the point of load application.

- The three stress parameters calculated in Study 3:

$$\sigma_{ax} = \frac{F}{A} \quad \text{Eq. 4.3}$$

$$\sigma_c = \sigma_{ax} + \sigma_{be} \quad \text{Eq. 4.4}$$

$$\sigma_t = \sigma_{ax} - \sigma_{be} \quad \text{Eq. 4.5}$$

Where σ_{ax} is axial stress; σ_c is compressive stress; σ_{be} is bending stress; σ_t is tensile stress; F is the force applied and A is the cross-sectional area of the bone.

- Maximal bending stress:

$$\sigma_{be}(\text{about } x\text{axis}) = \frac{(M_{be} \cdot y)}{I_x} \quad \text{Eq. 4.6}$$

$$\sigma_{be}(\text{about } y\text{axis}) = \frac{(M_{be} \cdot y)}{I_y} \quad \text{Eq. 4.7}$$

Where y is the maximal distance of the cross section from the neutral axis in the relevant direction and I_x or I_y is the area moment of inertia about the neutral axis in the relevant direction. The distance y was the mean distance of the outer 64 digitised points from the centroid.

- Maximal torsional stress:

$$\sigma_{tor} = \frac{(M_{be} \cdot R)}{I_z} \quad \text{Eq. 4.8}$$

Where I_z is the polar moment of inertia about the neutral axis, and is the sum of I_x and I_y and R is the radius of the outer surface of the bone.

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PUBLICATIONS&CONFERENCES

Published chapters

Nunns, M., Stiles, V. & Dixon, S. (2012). The effects of standard issue Royal Marine recruit footwear on risk factors associated with third metatarsal stress fractures. *Footwear Science*, 4(1), 59-70.

Publications not relating to thesis chapters

Nunns, M., House, Fallowfield, J., Allsopp, A. & Dixon, S. (in press). Biomechanical characteristics of barefoot footstrike modalities. *Journal of Biomechanics*. Available at <http://dx.doi.org/10.1016/j.jbiomech.2013.08.009i>

Rice, H., Nunns, M., House, C., Fallowfield, J., Allsopp, A. & Dixon, S. (in press). High medial plantar pressures during barefoot running are associated with increased risk of ankle inversion injury in Royal Marine recruits. *Gait and Posture*. Available at: <http://dx.doi.org/10.1016/j.gaitpost.2013.02.001>

Conference presentations relating to thesis chapters (as presenter)

Nunns, M. and Dixon, S. (April 2010). The effects of standard issue Royal Marine recruit footwear on risk factors associated with third metatarsal stress fractures. BASES Biomechanics Interest Group Meeting, Bath, UK (Poster Presentation).

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Dixon, S. Nunns, M., House, C., Rice, H. & Stiles, V. (August 2013). Ankle joint kinematics influence risk of third metatarsal stress fracture in military recruits. Footwear Biomechanics Symposium, Natal, BRA (Oral Presentation).

Other conference presentations

Nunns, M. & Dixon, S. (April 2008). The influence of running surface compliance on lower limb kinetic and kinematic variables. BASES Biomechanics Interest Group Meeting, Cardiff, UK (Oral Presentation).

Nunns, M., Dixon, S., Clarke, J. & Carré, M. (July 2009). Boot-insole effects on heel and fifth metatarsal pressures during running and turning in soccer. Footwear Biomechanics Symposium, Cape Town, SA (Oral Presentation).

Nunns, M., Dixon, S., House, C., Fallowfield, J. & Allsopp, A. (April 2011). The validity of instantaneous RSScan pressure software outputs to determine rearfoot movement in barefoot runners. BASES Biomechanics Interest Group Meeting, Chichester, UK (Oral Presentation).

Dixon, S., Nunns, M., House, C. & Fallowfield, J. (August 2011). Does limited ankle dorsiflexion increase forefoot loading during running in military recruits? Footwear Biomechanics Symposium, Tübingen, DE (Oral Presentation).

Nunns, M., House, C., Fallowfield, J., Allsopp, A. & Dixon, S. (April 2012). Kinematic Differences Between Barefoot Running Foot Strike Modalities. BASES Biomechanics Interest Group Meeting, Ulster, NI (Oral Presentation). *Winner of best abstract.*

Rice, H., Nunns, M., House, C., Fallowfield, J., Allsopp, A. & Dixon, S. (April 2012). High medial plantar pressures during barefoot running are associated with increased risk of ankle inversion injury in Royal Marine recruits. BASES Biomechanics Interest Group Meeting, Ulster, NI (Oral Presentation).

CHAPTER 1: INTRODUCTION

Stress fractures, also known as 'march' or 'fatigue' fractures are well documented overuse injuries first reported by Breithaupt in 1855. They affect bone and occur as a result of the repeated application of submaximal loads without sufficient time for recovery. As such, individuals who engage in activities which frequently load the lower limb, such as running, are at risk of developing a lower limb stress fracture. Stress fractures may affect any bone, but are most commonly observed in the lower limb and are associated with dull localised pain, usually prohibiting exercise and potentially escalating to full frank fracture (Brukner, Bennell & Matheson, 1999). Incidences of stress fracture have been reported in a variety of activities including ballet (Albisetti, Perugia, De Bartolomeo et al., 2009), soccer (Woods, Hawkins, Hulse & Hodson, 2002) and tennis (Maquirriain & Ghisi, 2006), but are most frequently reported in distance runners and military recruits. In these populations, lower limb stress fracture incidence rates are well reported and range from 3.3 to 31% (Almeida, Williams, Shaffer & Brodine, 1999; Armstrong, Rue, Wilckens & Frassica, 2004; Beck, Ruff, Mourtada et al., 1996; Beck, Ruff, Shaffer et al., 2000; Bennell, Malcolm, Thomas et al., 1996b; Kaufman, Brodine, Shaffer, Johnson & Cullison, 1999; Khan, Khan, Ahman, Jeilani & Khan, 2008; Milgrom, Giladi, Stein et al., 1985; Ross & Allsopp, 2002; Shaffer, Rauh, Brodine, Trone & Macera, 2006). In military institutions, the cost of sustaining a lower limb stress fracture is significant.

At the Commando Training Centre, Royal Marines Lympstone, Devon (CTCRM), recruits undertake a 32 week infantry training programme, considered to be one of the longest and most gruelling continuous training programmes in the world. The training programme involves a range of exercise modalities, including marching, running, climbing, cross training, gym work and jumping – either unloaded or with load ranging from 4.5 kg (service rifle) to 35.1 kg (rifle, Bergen and full pack). A 2012 review of training by Rice, Davey, Dixon & Fallowfield (2012) details the weekly breakdown of scheduled activities in the programme, and highlights the progressive nature of the course. In the first 10 weeks of training, the majority of activities involve running or marching distances between 2.4 – 9.6 km. Trainers are used for the first 2 weeks and phased out completely by week 9. Load carriage (20.5 kg) is introduced to a 6.4 km march in week 5 and then gradually increased in magnitude, duration or frequency as training progresses. After week 8, all scheduled activities are load-bearing. The second half of the programme, after week 15, sees the introduction of more specialised training activities, including operational simulations and assessment runs. Several off-site exercises lasting 3-7 days take place during the final weeks of training and include coverage of unspecified distances carrying loads of at least 13.7 kg. In order to pass out of training, recruits are required to complete a battery of final tests in week 31, all carrying a 13.7 kg load: the 6 mile endurance course in under 72 mins; a 9 mile speed march in under 90 mins; the ‘Tarzan’ assault course in under 13 mins and a 30 mile march across Dartmoor in under 8 hours. During the 32 week programme a ‘clustering’ of stress fracture reports occurs during weeks 26 and 31. During these weeks, work done is greatest (summed load carriage in that week, multiplied by the total distance covered) (Rice et al., 2012). Given the

predominance of marching and running activities in the RM recruit training programme, investigation of movement patterns and associated loading during running may reveal relationships with stress fracture development.

Recruits who sustain a stress fracture at CTCRM will be forced to withdraw from their original training troop for rest and rehabilitation, a process which may enforce absence from full training for around 12-20 weeks (Ross & Allsopp, 2002). This setback may discourage the recruit from continuing as they will have to re-join training with a different troop and forge new relationships; and with the elevated risk of re-injury (Schneider, Begelow & Amoroso, 2000), the chances of the individual completing training are reduced. Given the expenditure on marketing and recruitment (approximately £10k per recruit, personal communication) every recruit who fails to complete training represents a financial loss to the MoD. Additionally, the bill that arises when a recruit undergoes rehabilitation is severe, with an estimated cost of £1500 per week spent in rehabilitation. With lower limb stress fractures causing the most training days lost in basic military training, the all round costs of this type of injury are highly significant (Jordaan & Schwellnus, 1994; Ross & Allsopp 2002). Furthermore, there is a duty of care incumbent upon the MoD to reduce injury risk in recruits.

Approximately 4% of recruits will sustain a lower limb stress fracture during Royal Marine (RM) training. In other military populations, stress fracture rates have been reported to be slightly higher in some institutions, for example reported stress fracture rates of 6% in males (Beck et al., 1996) and 5.3% in

females (Beck et al., 2002) have been reported in the US Marine Corps; 13.4% in naval recruits (Kaufman et al., 1999); and 24% in the Israeli Defence Force (Milgrom, Finestone, Shlamkovitch et al., 1994). In populations of recreational runners, varying rates of stress fracture incidence have been reported, for example 21.6% (Bennell et al., 1996) and 9% (Yagi, Muneta & Sekiya, 2013). A 2011 review of stress fracture incidence in military and athletic populations by Wentz, Liu, Haymes & Illich demonstrated that approximately 3% of males sustain a stress fracture in military settings, with this figure around 6.5% in athletic populations. For females, the incidence rate was around 9% in both settings. Although lower than some published examples, these figures show that the incidence of stress fracture in RM recruits is typical of military training settings.

Recent figures suggest that the rate of lower limb stress fracture incidence at CTCRM has remained steady since the Ross & Allsopp report in 2002 (House, Reece & Roiz de Sa, 2013). Throughout the period covered by these reports, the most common stress fracture site has been the third metatarsal (MT3), accounting for approximately 40% of all stress fractures. The tibia is the second most common site of stress fracture at approximately 30% (Ross & Allsopp, 2002; House et al., 2013). In other military and athletic populations the tibia is most frequently affected, followed by the metatarsals – usually grouped to include all metatarsals (Armstrong et al., 2004; Beck et al., 1996; Bennell et al., 1996; Iwamoto & Takeda, 2003; Milgrom et al., 1985). These figures suggest that certain necessary requirements of RM training may expose recruits to loading that affects the MT3 in particular, therefore given the unique distribution

of injuries in this population, an investigation of risk factors (variables associated with increased risk of injury) for third metatarsal stress fracture is warranted. Furthermore, in light of the dearth of literature regarding this injury, an investigation of risk factors for this injury would provide valuable insight into its aetiology.

Accounting for around 70% of lower limb stress fractures between them, third metatarsal and tibial stress fractures present a significant burden on RM recruits and the MoD and should be investigated with a view to identifying risk factors for their occurrence and developing interventions to reduce their incidence. The aim of this thesis was therefore to identify risk factors for tibial and third metatarsal stress fractures in recruits undergoing Royal Marine basic training, with specific focus on the influence of standard issue footwear on metatarsal loading, and aspects of individual gait and anthropometry which may predispose recruits to both types of stress fracture in this training environment.

Chapter 2 provides an extensive literature review of the background of stress fractures, methods for analysis of risk factors and current understanding regarding the aetiology of the two injuries of interest. A large body of previous research has been carried out in order to try to understand the aetiology of lower limb stress fractures, with discussion broadly focusing on two areas: the ability of bone to withstand load, and the nature of loads applied to the bone. The former category is largely determined by intrinsic factors such as bone geometry and density, while the latter is affected by numerous factors which may be extrinsic or intrinsic in nature. Extrinsic factors are external or

environmental factors such as equipment used and the training environment. Intrinsic risk factors are numerous and may be described as any characteristic of the individual in question, including age, flexibility, limb alignment and fitness. One advantage of investigating the RM recruits involved in training at CTCRM is that a number of these factors are controlled. Age, gender, nutritional intake, footwear, load carriage, exercise patterns and terrain are all controlled to some extent. As such, the training programme experienced by all recruits should be similar, facilitating the identification of individual intrinsic factors affecting injury risk in this population.

In light of the burden of tibial and MT3 stress fractures upon RM recruits and the MoD, this thesis sought to identify factors predisposing RM recruits to increased risk of these types of injury. Three studies were designed in order to achieve this, based on existing aetiological evidence, gaps in the literature and the requirements of CTCRM. Given that RM recruits suffer an unusually high incidence of third metatarsal stress fractures, specific aspects of training which all recruits experience may be implicated in putting this particular structure at risk. The ability of footwear to influence loading of the foot has been extensively documented, therefore this aspect of training was identified as an area for consideration in Study 1 (Chapter 3). Study 2 (Chapter 4) investigated the influence of both individual bone geometry and external loading and foot position on estimated MT3 bone stress through the development of a mathematical model. This model was applied to the footwear assessed in Study 2 with the aim of quantifying its influence on MT3 bending loads. The first two studies examine both the influence of standard issue footwear on risk factors for

MT3 stress fracture, and links between external loading and estimated MT3 bone stress. Study 3 used a prospective design to identify baseline characteristics of RM recruits that predispose them to MT3 or tibial stress fracture. There have been no prospective cohort studies in military populations utilising running gait analysis to assess stress fracture risk. The prospective approach was taken in order to negate the limitations of the more common retrospective study design, where gait characteristics may change following injury, masking the mechanisms which may have caused a stress fracture to occur.

Through the implementation of the three studies described in this thesis, it was intended that substantial gains be made in the knowledge and understanding of third metatarsal and tibial stress fractures in RM recruits. This insight will potentially inform future strategies for identifying at-risk recruits at the start of training, and offering suitable interventions to prevent injury. In addition to the direct benefit to the MoD and CTCRM, development of knowledge about third metatarsal stress fracture in particular will contribute to addressing a large gap in the research.

CHAPTER 2: REVIEW OF THE LITERATURE

2.1. Introduction

The nature of basic infantry military training programmes leads to a certain degree of injury risk, with stress fractures posing a significant burden on recruits and the entities involved with the funding of such programmes. The two injuries of specific focus in this thesis are TSF and MT3SF, which account for a large proportion of the total stress fractures incurred by RM recruits undertaking basic training. In order to contribute to the understanding of why such injuries occur, it is important to review existing etiological theory and identify areas for further investigation. This review of literature examines current theory on the suggested mechanisms of exercise-induced stress fractures, focusing on biomechanical risk factors and the measurement approaches used to quantify them. At the end of the review, areas for future investigation and the aims of this thesis are presented.

2.2. Background to stress fractures

2.2.1. Structure and function of bone

Bone is a connective tissue which performs both mechanical and physiological roles within the body. A primary mechanical function of bones is to act as levers

which transmit the forces produced by muscles to cause movement. Long bones, which are the main levers for gross movements, are characterised by a long cylindrical shaft (diaphysis) and wider, denser ends (epiphyses). As the tibia and metatarsals are examples of long bones, this type of bone will be explored in further detail.

Long bones are composed grossly of cortical and trabecular bone. Cortical, also known as 'compact' bone, is resistant to bending, and much denser than the inner trabecular or 'cancellous' bone. Trabecular bone is named after the thin fibres called trabeculae that form it. These fibres align themselves parallel to the direction most commonly loaded and are strongest under compression (Nigg & Grimston, in Nigg & Herzog, 1999). Bone is composed primarily (around 95%) of type 1 collagen (Schaffler & Jepsen, 2000) with the remaining material consisting of a non-organic matrix of calcium and phosphates in the form of hydroxyapatite. This dual arrangement gives bone its strength and flexibility.

Strain on the bone above a certain threshold encourages a process of regeneration called remodelling, which results in a structure that exhibits anisotropic behaviour, as summarised by Wolff's Law (1870). This process sees bone become stronger in the most frequently loaded direction (Whiting & Zernicke, 1998), which in long bones is the longitudinal direction. The mechanotransduction model describes the response of bone to mechanical loading in four distinct stages: mechanocoupling; biochemical coupling; transmission of signal and effector cell response (Burr & Martin, 1992; Duncan & Turner, 1995). To summarise these stages, mechanical loading (particularly

dynamic loading) of the bone causes cellular movement within the structure which in turn promotes the transmission of electrical signals. These signals trigger the activation of osteoblasts and osteoclasts in the remodelling process. The magnitude, rate and direction of loading will all influence the amount of bone formation that occurs (see section 2.2.2).

The remodelling process involves the removal (resorption) of damaged bone and replacement (formation) of new tissue that is structurally stronger. The resorption phase involves the removal of damaged tissue by osteoclasts, which leaves tunnels typically 0.2 mm in diameter and a few millimetres in length in the surface of the bone (Schaffler & Jepsen, 2000). The formation phase involves the filling of these tunnels. Osteoblasts place a matrix within the tunnels and later deposit hydroxyapatite to form mineralised bone. The gradual concentric formation of mineralised bone around this osteoblast creates a secondary osteon. Secondary osteons have larger Haversian canals than primary osteons, and are surrounded by distinctive cement lines, which separate them from the surrounding bone tissue (Nigg & Grimston, in Nigg & Herzog, 1999). The material properties of cement lines are not well understood, and there is debate over their effect, with some evidence suggesting that they reduce the shear strength of bone (Frasca, 1981) and general agreement that bone stiffness may be reduced (Raeisi Najafi, Arshi, Eslami, Fariborz & Moeinzadeh, 2007). However, some experimental evidence indicates that resistance to shear forces may be increased and microcrack propagation inhibited by the microscopic gaps around secondary osteons (O'Brien, Taylor & Lee, 2003, 2007). Donahue, Sharkey, Modanlou, Sequeira & Martin (2000)

indicated that microcrack development was limited by the presence of cement lines and osteons, which may be a reason for the increased strength of secondary bone. Further research is required to understand the consequences of cement line development. Microcracks, or small fissures in the bone structure, were first reported by Frost (1960) in human rib samples, and are widely considered to be a significant step in the development of stress fractures, although they are not yet fully understood.

2.2.2. Stress fracture aetiology

Stress fractures may occur in settings where the demand for bone repair exceeds the capacity of the remodelling response to meet it following repeated sub-maximal loading. Excessive demand for repair may occur due to either an increase in bone damage, or the inhibition of the remodelling response. The remodelling response may be inhibited for reasons such as an imbalance of cytokines and prostaglandins (Raisz, 1999), or due to the presence of disease such as osteoporosis or Paget's disease. In the case of these chemical influences on remodelling, a detrimental interaction of resorption and formation usually occurs to reduce bone mass, density or strength (Raisz, 1999). Military recruits who sustain stress fractures are likely to experience an interaction between greater loading and/or the insufficiency of the musculoskeletal system to withstand such loads that results in excessive damage to the bone. The chemical influences on stress fracture aetiology are important, however it is beyond the scope of this thesis to investigate them, instead a focus on the relationship between load and bone resistance is maintained.

The mechanostat model of bone response to loading, developed by Frost throughout a series of papers (1983; 1987; 1990a; 1990b) and summarised by Burr & Martin (1992), suggests that the balance between resorption and formation is dependent on the magnitude of bone loading. The mechanostat model describes four mechanical usage windows, each framed by upper and lower limits of minimum effective strain (MES). The disuse window, when bone is largely unloaded, describes a status where resorption exceeds formation. Above this, with strain exceeding between 50-200 microstrain (change in the length of bone per unit length of bone, in parts per million) but not more than 1500-2500 microstrain, is described as the physiological window. Within the physiological window, a balance between resorption and formation is maintained, however strains exceeding the upper MES of this window are proposed to promote formation over resorption in order to increase bone mass and thus reduce local strains. Strains of above 4000-5000 microstrain may be considered 'pathological overload' and cause the formation of woven bone (Duncan & Turner, 1995).

Despite these proposed MES thresholds, applied stresses and bone properties may vary widely at different bone locations, therefore the load required to trigger remodelling is likely to vary between individuals and sites. Additionally, despite these guidelines for MES, it has been proposed that there is a linear relationship between bone formation and increases in strain magnitude above 1000 microstrain (Rubin & Lanyon, 1985). The rate and frequency characteristics of load contribute to the varied response of bone to external

load, and explain why the MES model does not ultimately predict the response of bone to external loads. For example, by maintaining the magnitude of load and the number of cycles per day, but manipulating the frequency of load application (number of load cycles per second), Quin, Rubin & McCleod (1998) showed that the strain threshold for bone maintenance reduced from 1200 to 100 microstrain. Similarly, Mosley & Lanyon (1998) demonstrated significantly greater bone formation in rat tibia subjected to a load at a high rate, compared to those loaded with the same magnitude but lower rate. In circumstances where applied strain triggers remodelling, bone is susceptible to the accumulation of fatigue damage. When considering the nature of load and whether it may trigger remodelling, evidence suggests that higher magnitude, frequency and rate of loading are important.

Damage in response to sub-maximal loading occurs initially in the form of microdamage, or fatigue damage, commonly referred to as microcracks. Microcracks result from the inability of the bone to withstand applied load and are a manifestation of excessive localised strain or deformation experienced by the bone. O'Brien et al. (2003; 2007) found that during fatigue testing of bone (repeated application of submaximal loads *in vitro*, usually until the point of failure), microcracks developed rapidly at the beginning of testing and then stopped occurring until the period immediately prior to failure, while Donahue et al. (2000) found similar levels of microcrack propagation at sites frequently (second metatarsal) and rarely injured (fifth metatarsal). It seems apparent that microcracks develop initially as a response to relatively low levels of loading, and that they mark a stage in the process leading to stress fracture

development. This may be as a result of affecting the response of bone to loading by altering its elastic modulus. The relationship between stress and strain in bone is linear while stress levels remain below the elastic modulus. When stress exceeds the elastic modulus, the deformation of the structure is no longer linear, and furthermore the structure will act inelastically, no longer returning to its original shape upon unloading. At this stage, fracture occurs. Microcrack accumulation has been shown to decrease the elastic modulus, reducing the failure threshold (force required to break) of bone (Keaveny, Pinilla, Crawford, Kopperdahl & Lou, 1997; Burr, Turner, Naick et al., 1998; Garrison, Gargac & Niebur, 2011).

The ultimate strain required for full fracture of healthy bone is in the region of 10,000 microstrain, which is around four to five times more than has been reported to occur in human locomotion. Strain magnitudes have been investigated *in vivo* using tibial strain gauges by Burr, Milgrom, Fyhrie et al. (1996), who reported that activities typical in Israeli military recruit training did not elicit peak strain magnitudes above 2000 microstrain. Even at this strain level, *in vitro* loading studies suggest that it would theoretically require over 10 million load cycles before bone failure (Carter, Caler, Spengler & Frankel, 1981), and even then failure may not occur (Schaffler, Radin and, Burr 1989; 1990). Evidently, stress fractures occur at lower exercise thresholds and load repetitions than this; at loading levels where microcracks occur. Indeed, fatigue testing of bone has caused significant microcrack accumulation (quantified by stiffness loss) without ultimate failure (Schaffler et al., 1989; 1990; Carter et al., 1981; Patten, Caler & Carter, 1996). The development of microcracks promotes

the remodelling response, and reduces the elastic modulus of bone, potentially increasing its susceptibility to future damage and highlighting the cyclical nature of stress fracture development.

One criticism of *in vitro* studies such as Carter et al. (1981) is that they do not account for the shear stress experienced during dynamic loading *in vivo*. This is likely to be significant, as long bones are mechanically less strong when loaded away from the 'preferred' direction. The anisotropic nature of bone means that it is strongest in the most commonly loaded direction, which for long bones is the longitudinal direction. Examination of the failure thresholds of cortical and trabecular bone highlights this phenomenon. Cortical bone, which plays a much greater role in load bearing than trabecular bone, requires far greater stresses to induce failure than trabecular bone. In the longitudinal direction, trabecular bone has an ultimate failure threshold of around 10 MPa in both tension and compression (Nigg & Herzog, 1999), while cortical bone is approximately 1.4 times stronger under compression than tension; approximately 200 MPa under compression and 150 MPa under tension (Nordin & Frankel, 2001). Both types of bone are weaker under transverse loading, for example the failure threshold of cortical bone is reported to be in the region of 130 MPa under compression and 50 MPa under tension (Martin, Burr & Sharkey, 1998).

It is no surprise that microcrack propagation has been shown to increase in severity following changes in loading modality towards less frequently loaded directions (Wang & Niebur, 2006), highlighting the potential role of 'unaccustomed' loading in stress fracture aetiology. Initial stimulation of the

remodelling response in bone may be hormonal or chemical, but in exercise settings is usually due to physical loads. When unaccustomed loading is experienced, the trigger threshold of stimulation is lower (Nigg & Grimston, in Nigg & Herzog, 1999) because the bone is less well adapted to stress it does not normally experience.

It has been suggested that the complete process from initial stimulation to increased structural integrity of bone takes around 90 days (Li, Zhang, Chen, Chen & Wang, 1985), meaning that it is potentially weakened for this entire period. However, the time between the resorption and formation phases of remodelling, usually two to four weeks after stimulation (Schaffler & Jepsen, 2000), leaves bone in its most weakened state, and thus more susceptible to further accumulation of microdamage (Popovich, Gardner, Potter, Knapik & Jones, 2000). It is during this period, known as the period of 'transient weakness' (Li et al., 1985), where further loading of the structure is most likely to result in rapid accumulation of microdamage and subsequent stress fracture. For example, in rabbit tibiae, where loading was repeated to induce stress fracture, evidence of bone lesions occurred in weeks 2-6 following loading, with some healing occurring between weeks 6 and 9 after loading (Burr, Milgrom, Boyd et al., 1990). Given the cyclic nature of the bone remodelling process, the provision of sufficient rest is of importance in the management of stress fracture risk during continuous training programmes. Figure 2.1 briefly summarises the cycle of overuse injury development, highlighting the risk period which occurs after the stimulation of the remodelling response.

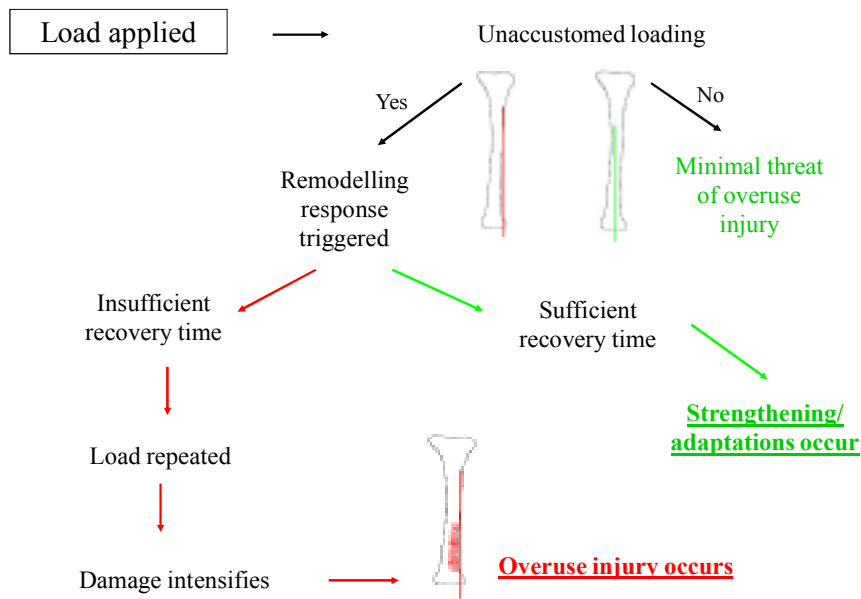


Figure 2.1. The cycle of overuse injury development.

As rest periods during RM training are prescribed by the training programme, an understanding of the structure and function of bone points to two broad areas of interest regarding the aetiology of stress fractures in this population. These are the nature of loads being applied to the lower limb, and the ability of the bone to withstand such loads. Methods for quantifying these loads are discussed below.

2.3. Biomechanical approaches to the estimation of bone loading

Biomechanical research utilises a variety of approaches to the quantification of kinetics, kinematics and human characteristics involved in movement. When investigating the causes of stress fractures in the tibia and metatarsals, loads

acting upon these structures must be measured or estimated. This section discusses the approaches to the measurement of these variables.

2.3.1. Direct measurement

The ideal scenario when investigating bone strain is to be able to directly measure the forces in question. The use of strain gauges makes this a possibility. Originally used *in vivo* in human tibiae by Lanyon, Hampson, Goodship & Shah in 1975, bone-mounted strain gauges were surgically attached to the anteromedial midshaft of the tibia, allowing measurement of strain in that area whilst the subject performed certain activities. Strain gauges provide an electrical signal proportional to the deformation of the gauge, which is interpreted to provide information on the level of strain experienced by the bone. The sensitivity of the gauge is crucial to its function, because the strain values it detects are so small. Furthermore, the force required to register a reading must be small enough that it does not interfere with the displacement of the bone.

Different types and arrays of strain gauges have been attached to bone sections, providing information on strain in different directions and over varying regions of the bone surface area. For example, strain gauges bonded to the surface of the bone were used historically (Perry & Lissner, 1962, cited in Milgrom, Finestone, Hamel et al., 2004) before the development of instrumented staples by Buttermann, Janevic, Lewis et al. (1994), which require

considerably less invasive surgical procedures in their use. Lanyon et al. (1975) used a rosette arrangement (several bonded strain gauges aligned at different angles) which revealed a non-uniform pattern of tibial loading. Since this ground breaking study, a number of other authors have investigated bone strain in a similar way, largely also focusing on the tibia and second metatarsal. However, reported values have provided no consistent evidence of loading characteristics. This may be due to variations in the placement of the strain gauges, but study strength may also suffer from ethical, financial and logistical difficulties. For example, Lanyon et al. (1975), along with studies by Burr et al. (1996), Milgrom, Finestone, Levy et al. (2000), Milgrom, Finestone, Segev et al. (2003) and Milgrom, Radeva-Petrova, Finestone et al.(2007) used members of their research team, and no more than three subjects to produce results. Despite the limitations of such studies, they provide unique insight into bone loading, and have produced some interesting findings.

Lanyon et al. (1975) demonstrated the multi-directional loading that the tibia experiences during a variety of activities; the large differences in peak strains between running and walking; and distinct loading phases present during the gait cycle. Burr et al. (1996) further highlighted the large increases in principal strain and strain rate magnitudes with increasingly vigorous activities, with sprinting and downhill running eliciting the highest magnitudes. Milgrom et al. (2000) demonstrated that drop jumps from heights of up to 52 cm do not elicit greater tibial strains than running at 17 km.h⁻¹. Three significant studies looking at *in vivo* second metatarsal strains have found greater magnitudes in a military boot associated with increased incidence of stress fracture at that site (Arndt,

Westblad, Ekenman & Lundberg, 2003); greater strains following fatigue and with the addition of load carriage (Arndt, Ekenman, Westblad and Lundberg, 2002); and that strain magnitudes in the metatarsal are greater than the suggested threshold for cortical fatigue failure (Milgrom, Finestone, Sharkey et al., 2002). Finally, Milgrom, Agar, Ekenman et al. (2011) explored the effects of various orthotics on tibial and second metatarsal strains, finding that only a custom polyurethane composite orthotic caused a reduction in peak strains at both sites when worn in army boots, as well as having some effect at the second metatarsal when used in trainers. The reliability of studies using staple gauges has been described as poor (Milgrom et al., 2004), and these studies are all limited by small sample sizes and examination of small site areas, however they provide some useful information that will be referred to later in this review.

Cadaver studies provide another opportunity to measure directly the forces experienced by lower limb structures during simulated gait. Although an inevitable weakness of such studies is the difficulty in replicating lifelike dynamic movements, they have produced useful findings. For example, Sharkey & Hamel (1998) simulated walking using five cadaveric feet, and modelled compressive force acting on the tibia, as well as measuring dynamic plantar pressures and ground reaction forces. Donahue et al. (2000) used cadaver feet to simulate walking, and measured bone strain and microcrack parameters at the second and fifth metatarsals. While peak bone strain was more than twice as large in the second metatarsal than the fifth, there was no difference in microcrack parameters between sites. Nester Liu, Ward et al. (2007) used a

similar dynamic walking model with cadaver feet to study the kinematics of the foot structures in response to applied loads. These authors demonstrated the important contribution of movement of the metatarsals relative to their proximal attachments in deflecting loads.

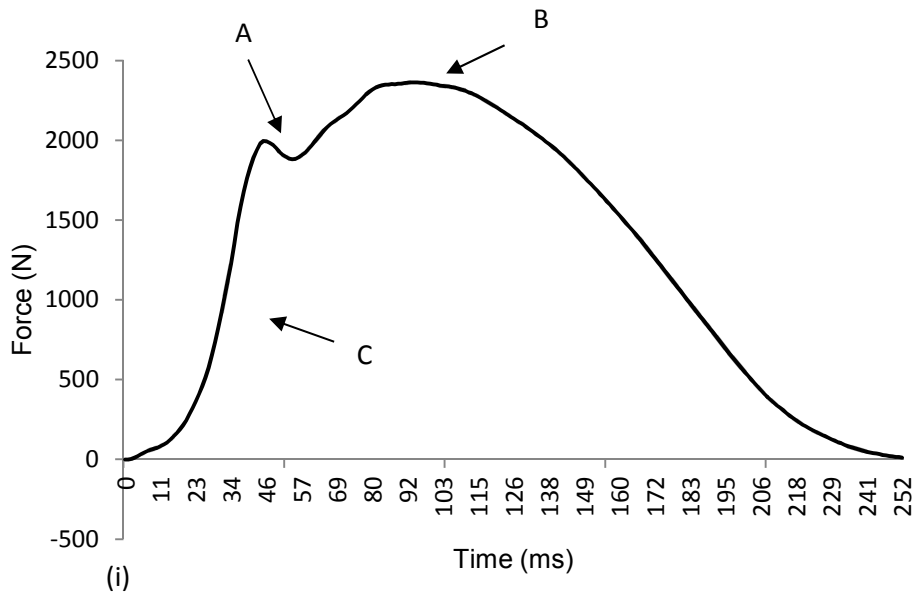
Direct measurement approaches provide useful and potentially precise and sensitive measurements of bone loading, however there are some significant limitations which prevent the widespread use of these techniques. There are alternative methods which have been more widely used in the estimation of factors contributing to stress fracture risk.

2.3.2. Ground reaction forces

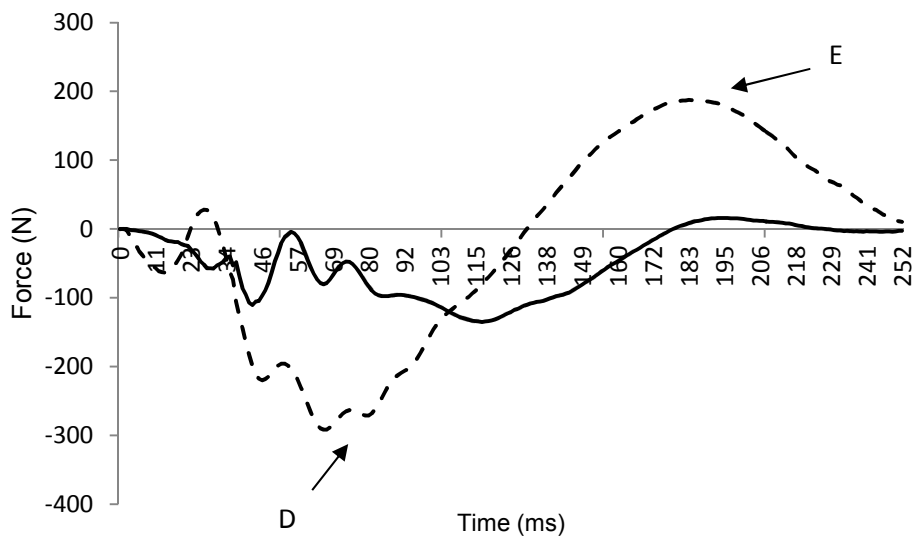
Force plates have become a commonly used tool in the estimation of loads acting on the lower limb. Newton's third law states that every action has an equal and opposite reaction, therefore if we wish to examine the force transmitted up the lower limb during stance, analysis of ground reaction forces will permit this. Ground reaction forces (GRF) can be obtained using a force plate, which is usually set flush within a laboratory floor. Data obtained from a force plate include the magnitudes of reaction forces in the vertical (F_z), anterior-posterior (F_y) and medial-lateral (F_x) planes, centre of pressure location and the horizontal (free) moment acting on the plate. Centre of pressure location indicates the point at which the resultant GRF vector can be

considered to be acting, while free moment is the rotational moment applied to the plate about the vertical axis.

A typical vertical GRF time history for a heel-toe shod runner is presented in Figure 2.2(i). The vertical component of GRF has been widely used to make inferences about lower limb loading. The use of tibial-mounted accelerometers by Lafortune, Lake & Hennig (1995) revealed that GRF cause tibial accelerations during the impact phase in running in particular. The authors calculated transfer functions to accurately infer tibial loading from GRF data, and while a general similarity between acceleration and force data was shown for their five participants, individual adjustments to the transfer term provided improved data from which to infer loading. This highlights not only the relationship between GRF and lower limb loading, but that individual variations in this relationship exist, which are likely to be due to the way each individual dampens impact forces. Indeed Scott & Winter (1990) had previously suggested that tibial strain may be more influenced by activation of the surrounding muscles than GRF. Nevertheless, the main point of interest on the vertical force time history has been the magnitude of the vertical impact peak (if one is visible) and the gradient of the loading slope.



(i)



(ii)

Figure 2.2. (i) Typical vertical impact force (F_z) time history. Peak impact force (A), peak active force (B) and the gradient of the loading slope (C) are highlighted. (ii) Typical anterior posterior (F_y , dashed line) and medial-lateral (F_x , solid line) time histories. Peak braking force (D) and peak propulsive force (E) are highlighted on the F_y curve.

Force plates are a relatively accurate (given adequate calibration) and sensitive tool able to capture at high frequencies, typically around 1000 Hz, however

there are potential sources of error to consider when using them. Although the absolute force magnitudes produced by the plate are dependent on load cell performance, research suggests that free moment and centre of pressure data is more susceptible to error. The accuracy of centre of pressure location data has been shown to vary with vertical loads and location on the plate, with minimum thresholds required for accuracy which may deteriorate towards the edges of the plate (Lees & Lake, in Payton & Bartlett, 2007). Bobbert & Schamhardt (1990) performed an analysis of the accuracy of centre of pressure values provided by a Kistler force plate. The authors applied static loads to a wooden board placed over a Kistler force plate at 117 different locations, at a range of 0-2000 N in magnitude. The board was supported by a stylus in a known location, nestled in a drill hole in the surface of the plate. Dynamic loads were applied by a participant running over the plate. The errors reported in centre of pressure were greater towards the corners of the plate (up to ± 20 mm) reducing to $\pm 3-5$ mm in more central areas. Although errors were demonstrably reduced by around half with the application of a correction algorithm, these results indicate that where possible, foot strike should be aimed as centrally as possible when centre of pressure data are assessed.

In addition to potential reliability issues with the plate, there may be considered to be certain validity issues with the use of this measurement technique, dependent on its size and location. It is desirable to replicate 'typical' gait performance as closely as possible in a lab setting, but asking a participant to target a force plate has the potential to cause an unnatural stride. In studies of the effect of targeting in runners, Grabiner, Feuerbach, Lundin & Davis, (1995)

and Patla, Robinson, Samways, & Armstrong, (1989) reported no effect of targeting, although this is in contrast to the findings of Cuddeford, Yack, Jensen, Peterson, Simonsen & Eichelberger (1998). It may be possible to overcome targeting through the use of larger or multiple adjacent force plates, or by increasing the number of trials collected. Although variations in acquired force plate data may occur for reasons other than having to target the plate, Bates, Osternig, Sawhill & James (1983) demonstrated that once 10 trials are acquired, the addition of further trials will not result in a difference of more than 0.25 of the standard deviation of a criterion value (obtained from 15 trials). It is also commonly recognised that the tester should observe the participant's running gait to judge visually that no unnatural stride deviation occurs when collecting force plate data for running (Challis, 2001).

Force plates provide information on the magnitude and direction of the resultant force vector acting at the point of contact between the foot and the ground. Although inferences can be made about lower limb loading, impact attenuation by soft tissues will influence the extent to which this force is transmitted to lower limb structures. The resultant force vector is heavily influenced by the acceleration of the whole body centre of mass, and therefore in order to examine the distribution of forces acting at the interface between foot and surface, it may be beneficial to utilise plantar pressure analysis.

2.3.3. *Plantar pressure*

Since Marey's use of shoes with instrumented air chambers in the early 19th century (Nigg & Herzog, 1994), and Beely's use of plaster of Paris impressions in 1882 (Elftman, 1934), there have been further attempts to quantify the areas of high and low pressure on the plantar surface of the foot. Recent iterations of these attempts include pressure plates/mats and pressure insoles, although flexible pressure arrays that can be cut to size and used in most conceivable situations have also been developed. Pressure, defined as the distribution of force over an area, can be analysed using one of these devices in order to provide information on foot function in particular. Pressure measurement devices differ from force plates in two fundamental ways: they give the distribution of force over a contacting surface; they only detect forces normal to the sensor bed. As such, there are benefits and drawbacks to their use.

Pressure analysis systems have frequently been described and reviewed identifying key functional characteristics such as spatial resolution (e.g. number of sensors per unit area), sampling frequency and sensor type (Rosenbaum and Becker, 1997). RSScan insoles and pressure plates utilise similar sensor beds containing an array of resistive sensors at a typical arrangement of 10 sensors per 4 cm². Pressure insoles can fit within footwear to provide in-shoe pressure data over a number of consecutive steps. Sampling at 500 Hz, data are transmitted through cables to a data logger, worn on a belt, which has capacity for several seconds (around 7) of data collection. Pressure plates are produced in sizes ranging from 0.4 m x 0.5 m/1 m/2 m. In each case, sampling frequency

can reach 500 Hz, but with the increasing plate size, resolution must be compromised. The available bandwidth for data is restricted to 4 megabytes per second, therefore a compromise must be made with larger plates. The 1 m and 2 m long RSScan pressure plates are constructed by arranging multiple 0.5 m plates together. As each constituent plate scans progressively through lines of sensors, starting at the leading edge of the plate, this may result in errors with regard to timing information. Therefore, while a 2 m pressure plate may allow for data collection of more than one stride per trial, there are issues regarding data quality which must be considered. Pressure insoles allow the investigator to collect data over a number of steps and remove the influence of targeting a plate in the ground, however the devices are quite fragile and have a relatively brief lifespan. It is recommended that frequent calibration of insoles is performed during use, preferably against data from a force plate, because absolute magnitudes of values are reliable but may be inaccurate, underestimating vertical reaction force by as much as 50% compared to force plate data (Low & Dixon, 2010).

When conducting plantar pressure analysis, the foot is usually divided into regions of interest using either zone definitions or mask placements. Masks can be adjusted in size, and are free to be placed on any region of the foot, whereas zones are more rigidly defined based on the foot anatomy, although they can be manually adjusted. A sample plantar pressure profile is shown below (Figure 2.3), demonstrating typical adjusted mask (A) and zone (B) placements. Image C of Figure 2.3 shows the automatically defined zones of the foot, highlighting the need to check these before extracting data.

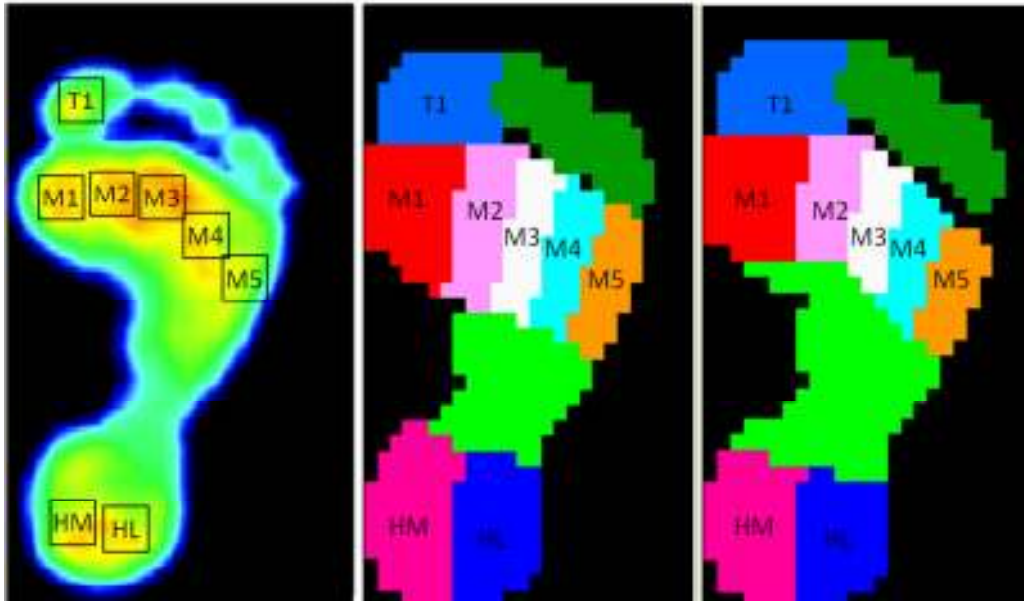


Figure 2.3. Example plantar pressure profile showing adjusted mask (left: A) and zone (middle: B) definitions. An example of automatically assigned zones is shown in (right: C).

Following the definition of regions of interest, there are a multitude of analyses that can be performed. For each region, the basic exports of peak pressure, impulse (the force-time integral), contact time and loading rate are available. Other available measurements include centre of pressure location referenced both to the foot in contact with the plate and the whole plate; ground contact time; resultant vertical force; calculation of the angle between the longitudinal foot axis and the longitudinal plate axis; estimations of subtalar joint motion; foot size and timings of key events such as first metatarsal contact and foot flat time. Further interpretation of raw data may involve the calculation of relative pressure or medial-lateral balance, or ratios between regions of interest, such as between the hallux and first metatarsal head, to estimate hallux stiffness. Data can be exported from accompanying software, such as *footscan* software (RSScan, Belgium), or exported for analysis in other programmes such as Matlab and Microsoft Excel. The choice of mask or zone analysis may be

dependent on the type of data the user wishes to obtain. For example, if the point of interest is directly above the third metatarsal head, then a mask placed in this region may be more appropriate than looking at the whole zone. If one wishes to obtain peak pressure data from the MT3 region, then either approach would provide this because it should appear where the bony head of the metatarsal protrudes. However the analysis of impulse above the MT3 would be less accurate with the use of a zone, as the defined area includes regions of the foot not directly above the MT3 head. Furthermore, the estimation of landmarks on the foot is subjective but could be improved with the use of an accompanying MRI scan, as used in Study 2. Where this approach is not possible, the use of standard placements published in the literature (e.g. De Cock et al., 2005) can reduce random error due to zone definition or mask placement.

2.3.4. Kinematics

In addition to the interpretation of external forces acting on the lower limb, it is important to consider the influence of kinematics upon the loads experienced during locomotion. It has been demonstrated that changes in lower limb kinematics can influence GRF, for example Dixon, Collop & Batt (2005) demonstrated that participants adjusted their ankle kinematics in order to maintain peak impact forces on two surfaces of differing hardness, while McMahon, Valiant & Frederick (1987) highlighted the role of knee flexion in impact attenuation through the reduction of vertical stiffness. In addition to modifying the magnitude of vertical loads, limb rotations may change the direction of loads applied to the bones of the lower limb. For example, high

levels of eversion of the foot during stance may cause torsional stresses to be applied to the tibia (McKeag & Dolan, 1989), or place excessive demands on muscles such as the tibialis posterior (Pohl, Mullineaux, Milner, Hamill & Davis, 2008) which are responsible for controlling such motion. Given their potential to influence lower limb loading characteristics, kinematic variables should be studied, usually through the calculation of joint angles derived from markers attached to body landmarks and referenced to an origin. The kinematic quantification of human gait is concerned with the absolute and relative rotations of segments and joints, which may occur through more than one plane of movement. Therefore the use of a three-dimensional model to calculate joint angles is usually preferable.

This thesis quantifies the kinematics of the knee and ankle. These joints facilitate varying degrees of motion in three planes, about three axes, motion which can be determined using a joint coordinate system. In simple terms, joint coordinate systems use markers to define local segmental axes, which can be related both to a global (laboratory) axis system and to an adjoining segment axis system. For each segment, (usually) three joint markers are used to define two 'body fixed' axes, with the third axis being mutually perpendicular to these. For example the z-axis of the thigh is defined by a unit vector created between markers placed on the greater trochanter and the lateral epicondyle of the knee, the x-axis by markers on the medial and lateral epicondyle of the knee, and the y-axis by the cross product of these unit vectors. Once the Cartesian coordinate system is similarly defined for the shank, relative knee angles can be calculated using three unit coordinate vectors, one being the x-vector of the proximal

segment, one being the z-vector of the proximal segment, and the final being the cross product of these, also referred to as the 'floating axis'. Relative motion between the segments is then quantified by calculating the angles between the body fixed axes and the floating axis. Joint coordinate systems can be modified for any articulating rigid bodies and may provide accurate data with alternative marker placements. Since the introduction of a joint coordinate system for the calculation of three-dimensional knee angles by Grood & Suntay (1983), adaptations of this model have been designed to examine the lower limb, most notably by Soutas-Little, Beavis, Verstraete & Markus (1987).

There are a number of assumptions made when using kinematic analysis. Joint coordinate systems rely on the premise that rigid bodies are articulating and rotating relative to each other. In fact, not only are bones not rigid, but sub dermal soft tissue movement and skin-mounted marker movement is likely to occur during dynamic movements. Reinschmidt, van den Bogert, Nigg, Lundberg & Murphy (1997) investigated differences between bone and skin-mounted markers in the estimation of knee kinematics during running, reporting between 20-70% errors associated with skin marker estimates depending on the mode of rotation. This finding built on reports of estimated differences of between 7 and 10 mm between skin and bone mounted markers during walking (Capozzo, 1991, cited in Manal, McClay, Stanhop, Richards & Galinat, 2000). Skin marker error is likely to be exacerbated when placed at areas where subcutaneous fat is stored, or if placed on clothes, as calculations assume a representation of an underlying bone location. Joint markers should normally be placed in order to represent functional axes of interest of the segments, such as

the long axis of a long bone. As such, care should be taken when placing these to avoid clouding interpretation of reported movements. One strategy to reduce the influence of marker placement error is to refer dynamic data to a static trial collected with the participant in a neutral stance (Reinschmidt et al., 1997). This is a common practice and assumes that all joints are aligned to a known value (e.g. neutral or ninety degrees), and therefore any non-neutral joint angles indicated by the standing trial occur as a result of marker placement errors, and should be subtracted from dynamic trial data. This approach does not take into account any structural misalignment of the individual's lower limb, and in some cases it is arguably preferable not to reference to static values unless completely neutral alignment is possible. However, referencing to a static trial removes the relatively large errors that may occur due to marker placement errors, and will not influence range of motion data, even if errors in absolute values arise due to alignment issues.

In addition to systematically providing a poor estimation of an underlying landmark, there is potential for random error to affect data quality due to marker movement independent of the segment. This is particularly likely in dynamic movements, when using large markers or markers placed on clothing. The influence of random marker oscillations can be reduced with smoothing or filtering of data. Signal data (such as raw marker coordinates) is comprised of both the true signal and noise introduced by data collection techniques, therefore filtering techniques aim to remove noise from the signal by eliminating components outside a desired frequency range, while smoothing techniques use piece wise polynomial curve fitting procedures (Wood & Jennings, 1979).

Both techniques achieve similar results and are used interchangeably in biomechanical analysis. The most common smoothing technique is the quintic spline, which has been shown to be a superior method in biomechanical research (Challis & Kerwin, 1988). Winter, Sidwall & Hobson (1974) proposed the use of Butterworth digital filters in kinematic analysis, and these types of filter are now commonly used as an alternative to smoothing. A Butterworth filter is typically low-pass, meaning that any data components occurring at a frequency above a set frequency are removed. The cut-off frequency should ideally be determined by residual analysis (Winter, 1990) as an incorrect threshold has the potential to affect the observed range of motion at a joint. Typically, cut-off frequencies between 6 and 12 Hz are reported. A further consideration of the Butterworth filter is that it introduces a lag effect, therefore zero-lag filtering is achieved by using recursive processing (Winter, 1990).

Kinematic techniques are flexible and when combined with the latest technology may facilitate the collection of relatively accurate and precise metrics of human movement. As with kinetics, this field of study is common in biomechanical research, but is not without limitations.

2.3.5. Mathematical modelling

Despite the proliferation of biomechanical studies utilising the aforementioned techniques, there are significant limitations within each approach that restrict the ability to quantify loads acting on the tibia and third metatarsal. While direct

measurement techniques provide an opportunity to obtain the most valid and accurate loading data, they are not without limitation and are not suitable for use with large cohorts. One approach which carries the potential to increase the precision and validity of estimates of loading is mathematical modelling.

The two broad approaches to modelling are inverse dynamics and forward dynamics. Inverse dynamics utilises equations of motion, along with information about the properties of segments, movement of segments and forces within the system, to estimate joint forces and moments. This approach uses measured movements and forces to estimate unknown forces and moments (Van Den Bogert & Nigg, in Nigg & Herzog, 1999). A forward dynamics approach uses measured forces as inputs to estimate muscle forces that perform the movement in question. The latter approach demands that an indication of muscle activity must be obtained or estimated (e.g. using EMG), and used to estimate muscle forces.

In both the forward and inverse dynamics approaches, knowledge of muscle locations, including lines of action of force and moment arm lengths are used to calculate joint moments. The inverse dynamics approach is the less complex and more common of the two approaches, and involves some inherent limitations. For example, net muscle moments are calculated meaning that assumptions are made about co-contracting muscles acting about a joint. Forward dynamics solutions, while potentially more accurate and powerful than their inverse counterpart, are complex and still have a number of limitations. At each stage, there are limitations to the approach, such as the difficulty and

inaccuracy involved in the calculation of muscle force from muscle activation estimates (Buchanan, Lloyd, Manal & Besier, 2005). Furthermore, regardless of the modelling approach, some assumptions have to be made about the contributions of individual muscles, and of structures such as supporting ligaments. The inverse dynamics approach, being less complex in nature, has been utilised more frequently in the literature.

The tibia was first modelled by Scott & Winter (1990) using estimates of muscle and bone forces to model tibial loading, with more recent models presented by Sasimontonkul, Bay & Pavol (2007) and Haris Phuah, Schache, Crossley, Wrigley & Creaby (2010). Scott & Winter (1990) obtained kinematics and kinetics to calculate joint reaction forces and sagittal plane bending moments using inverse dynamics, demonstrating the considerable contribution (80%) of plantarflexor muscular contraction to loads acting on the tibia. Sasimontonkul et al. (2007) and Haris Phuah et al. (2010) reported similar findings regarding the contribution of musculature to tibial loading, both finding an important effect of lower limb position on loading. Similar approaches were used in these three studies, which considered muscle forces, joint moments calculated using GRF data and inverse dynamics solutions. Modelling of bending moments at nine equidistant points on the tibia by Haris Phuah et al. (2010), highlighted a predominance of tensile loads acting on the posterior tibia, which increased in magnitude towards the distal end. This finding provided a suggestion as to why the distal end of the tibia is the most frequent site of stress fracture.

Despite a very small number of relevant studies, some of the biggest advancements in the understanding of MT3SF aetiology have arisen due to mathematical modelling. Research in this area has estimated the geometry of the third metatarsal and modelled the stress or strain acting on the bone as a result of loads experienced during gait. Arangio, Beam, Kowalczyk & Salathe (1998) built upon earlier work focusing specifically on the fifth metatarsal (Arangio, Xiao & Salathe, 1997), in which beam theory was used to calculate stress acting on slices of a plastic mould of the bone taken at 1.5 mm intervals. Points along the inner and outer surfaces of each slice were obtained and used to determine a local coordinate system for each cross-section, the centroid, and subsequently the moments acting about the three axes of the bone for a given load. Their study on the third metatarsal (Arangio et al., 1998) used computed tomography scans to calculate the cross-sectional area at 5 mm intervals of metatarsal bones two to five. For each cross-section, they obtained coordinates for the inner and outer surfaces of the bones, and simulated the shear and normal stresses experienced when loads were applied at the metatarsal head at 15 degree intervals between the horizontal and the vertical directions. The model indicated that stress in the third metatarsal was highest 3.5 cm from the proximal end of the bone, under a laterally-applied horizontal load, encouraging future research to consider the effect of these forces in MT3SF aetiology.

Gross & Bunch (1989) used inverse dynamics to estimate sagittal plane bending moments at the metatarsal heads during gait, relying on reference data for bone characteristics. This model provided some interesting findings but was quite simplified, representing the metatarsals as simple uniform ellipses. The

work of Arangio et al. (1998) represents an improvement on the work of Gross & Bunch (1989) however there has yet to be a study which has combined accurate geometric, bone position and external load data to estimate third metatarsal bone strain. Further complications include the necessary consideration of the influence of connective tissues and muscle action, presenting a difficult task overall. The development of models of bone loading, while challenging, may provide significant advances in the understanding of stress fracture aetiology.

2.4. Risk factors for stress fractures

In the context of the current biomechanical research, risk factors are considered to be variables which have an association with greater likelihood of sustaining an injury. In considering risk factors for stress fracture, this section has been divided into two parts. The first section contains a review of factors associated with increased risk of lower limb stress fracture in general, including non-biomechanical factors. The second section focuses on biomechanical characteristics identified as risk factors for tibial and MT3SF specifically, and is sub-divided into anatomical and dynamic risk factors.

2.4.1. General risk factors for stress fracture

2.4.1.1. Training errors

As highlighted in section 2.2.4, for stress fractures to develop it is necessary for the remodelling process to be triggered, but fail due to insufficient opportunity to rest and remodel to meet the demands of repeated load application. When considering the influence of exercise on stress fracture development, if the loads experienced by bone do not stimulate the remodelling process, there is no opportunity for injury. Similarly, if sufficient rest is provided to allow remodelling, there will be no injury. As the following aspects of participation/activity which may cause altered loading or insufficient rest can be controlled with relative ease, they may be considered to be training errors. Training errors include the type, intensity and duration of activity undertaken, as well as the training terrain and the footwear worn.

Training is standardised for all RM recruits, from which it may be inferred that individual characteristics predispose certain recruits to stress fracture, although aspects of the training programme, such as the 32-week duration, may influence the risk of stress fracture overall. Finestone, Milgrom, Wolf et al. (2011) followed military recruits through a progressive military training programme, and found that the stress fracture site swung dramatically from 78% occurring in the tibia or femur in the first two stages of the programme (total 26 weeks), to 91% occurring at the metatarsals in the final section of training (a further 26 weeks). This evidence suggests that either the increased intensity or longer duration of training were linked to the development of metatarsal, rather than tibial or femoral stress fractures, perhaps linking with the prevalence of MT3SF in RM training. Whilst the RM training programme is both long and physically demanding, it is difficult to quantify this in comparison to

other military training programmes, and therefore adds to the justification of research conducted specifically on this population.

For a given base level of fitness, the risk of injury is increased if the intensity, duration or frequency of activity is increased (Woods et al., 2002; Saglimbeni, Fulmer & O'Connor, 2007). This may be because the loads experienced by the bones of the lower limb are new or 'unaccustomed' during the new activity, or perhaps because increased physical work leads to earlier fatigue, and the bones are exposed to greater loads without the protection of muscle contractions. The prior conditioning of recruits may therefore influence the degree to which the activities involved in RM recruit training impart 'new' loads to the lower limb. For new military recruits, changes in training and/or recovery times may combine with other stressors including nutrition, sleep patterns and illness to increase injury risk (Fry, Morton & Keast, 1991), therefore preparedness for the demands of training is desirable. This has been assessed previously using fitness assessment at the start of training. Lappe, Stegman & Recker (2001) determined that engagement in regular physical exercise prior to activity was protective against stress fracture in female military recruits. Studies by Beck et al. (2000, assessing the number of sit-ups and the time taken to run 2 miles at the start of training), Välimäki, Alfthan, Lehmuskallio, Löyttyniemi, Sahi et al. (2005, exercise questionnaire, max distance run in 12 min, isometric quadriceps strength) and Shaffer et al. (2006, timed run assessment) all report that poor performance in physical conditioning assessments was associated with greater risk of sustaining a stress fracture in military recruits. Hoffman, Chapnik, Shamis, Givon & Davidson (1999) found that activity levels prior to recruitment in basic training, as well as strength indices,

were key determinants of stress fracture risk, while Moran, Finestone, Arbel, Shabshin & Laor (2012) identified the duration and frequency of pre-enrolment aerobic training as significant predictors of stress fracture in young male military recruits. The indices of physical fitness used in these studies are inconsistent, therefore caution should be taken when attempting to apply these findings to wider populations, however the evidence presented here suggests that variations in physical fitness may be a confounding variable if not quantified alongside biomechanical variables.

Muscle strength or rather muscle insufficiency has been linked with stress fractures. Scott & Winter (1990) found that the muscles of the shank act to reduce the bending stress on the tibia, therefore muscle insufficiency could expose bones to greater loads. The ability of muscles to provide this force may be compromised by fatigue. Mizrahi, Verbitsky, Isakov & Daily (2000b) investigated the effects of whole-body muscle fatigue on lower limb kinematics, showing increased impact acceleration of the shank after fatigue. Mizrahi, Verbitsky & Isakov (2000a) also examined the increased shock experienced in downhill running, attributing it to the increased demand on, and subsequent insufficiency of the eccentric action of lower limb muscles to attenuate shock. Whereas Mizrahi et al. (2000b) and other studies of a similar nature investigated the effects of whole-body fatigue, Flynn, Holmes & Andrews (2004) were able to isolate the gastrocnemius and tibialis anterior. In doing this, their findings were the opposite of those discovered in whole-body-fatigue experiments. Shock attenuation was shown to increase, most likely due to lack of tension in the fatigued muscles. In a similar study by Christina, White and Gilchrist (2001) the authors fatigued dorsiflexor and invertor muscles on the

participants' right foot, and found that this had a significant effect on the kinematics and kinetics of the ankle. In particular, fatigue of the dorsiflexors significantly increased the loading rate of peak impact force. Milgrom et al. (2007) conducted an investigation of tibial strain following fatigue using *in-vivo* strain gauges to measure bone strain following fatiguing exercise in Israeli military recruits. Their results showed that muscle fatigue had occurred and that tensile and compressive strain rates in bone were significantly greater post exercise compared to pre exercise.

Both Milgrom (1989) and Bennell et al. (1996) assessed the quantity of lean muscle mass in the lower limb by measuring calf girth and subtracting skinfold thickness. Milgrom (1989) observed lower corrected calf girth in males who experienced a TSF, while Bennell et al. (1996) observed the same finding for female runners but not males, and observed a positive correlation between lower limb muscle mass and bone mineral density (BMD). Not only does increased muscle mass function to reduce peak bending loads acting on the tibia, but its association with increased BMD suggests that the transmission of greater bending moments to the bone in stance and general locomotion may stimulate remodelling. For RM recruits with relatively low muscle mass, the infantry training programme may represent a potentially injurious increment in workload, and therefore either increased opportunities for rest, or a pre-training muscle strengthening programme would be required to reduce risk of stress fracture. Lower fitness upon entry to training or insufficient provision of rest in the programme may hasten or intensify the effects of fatigue, representing a training error that promotes unaccustomed loading of the lower limb.

While evidence points to pre-training conditioning as a significant risk factor for stress fracture, one would expect this to play less of a role with RM recruits, due to the strict minimum physical requirements for entry into training. In order for applicants to be eligible for the Potential Royal Marines Course (PRMC), a precursor designed to assess eligibility for RM recruit selection, recruits must pass a test comprising of two timed treadmill runs of 2.4 km each at a 2% incline. The first 2.4 km run must be completed in less than 12 minutes 30 seconds and immediately afterwards run the second 2.4 km in less than 10 minutes. Once selected for PRMC, potential recruits must successfully complete four physical tests over two days, which include timed push-up, sit-up and pull-up tests, VO_2 max assessment, a three mile run, a swimming test, an assault course and an endurance run (www.royalnavy.mod.uk). Given the rigorous nature of this selection process, it should be assumed that RM recruits have a relatively high level of physical fitness upon entry to training. Regardless of this, RM recruits certainly experience fatigue on a regular basis during training, potentially exposing their bones to damaging loads.

In conditions of reduced loading, bone resorption increases and BMD decreases, as shown in examples of prolonged bed rest (LeBlanc, Schneider, Krebs et al., 1987; LeBlanc, Schneider, Evans, Engelbretson & Krebs, 1990; Inoue, Tanaka, Moriwake et al., 2000) and space flight (Oganov, Grigoriev, Voronin et al., 1992; LeBlanc, Schneider & Shackleford, 2000). Sedentary women have been shown to possess significantly lower total body BMD than their athletic counterparts (Madsen, Adams and Van Loan, 1998), with a similar study showing that 30% of a sample of sedentary female students had low total body BMD (as defined by international bodies) compared to 16% of varsity

athletes (Hoch, Pajewski, Moraski et al., 2009). Due to the capacity of bone to adapt to loads and maintain suitable calcium levels, disuse has been shown to lead to significant reductions in bone mass, however the opposite can also be said to be true. Activity has been demonstrated as a powerful tool to increase BMD, even outweighing the effects of amenorrhea and oligomenorrhea in female gymnasts (Robinson, Snow-Harter, Taaffe et al., 1995). Shackleford, LeBlanc, Driscoll et al. (2004) demonstrated that resistance exercise performed during 17 weeks of horizontal bed rest helped to significantly negate loss of bone mass when compared with controls who only followed the rest protocol, while load bearing exercise is well established as a tool for increasing BMD (Rutherford, 1990) and even protecting against osteoporosis (Drinkwater, 1993; Howe, Shay, Dawson et al. 2011). Although reference to total body bone mass has been made, the response of bone to loading is local, and in the consideration of TSF and MT3SF, lower limb loading is of importance. Section 2.4.1.3 considers the effects of local bone characteristics in greater detail, however the evidence presented here suggests that, while successful applicants to RM recruit training must meet certain fitness requirements, the activity involved in developing this fitness should involve load-bearing exercise of the lower limb if bone health is to be suitable for the rigours of training.

In terms of protecting potential military recruits from experiencing damaging loads during training, certain types of load bearing exercise may be beneficial in pre-conditioning. Dynamic intermittent activities such as team sports, particularly soccer, have been shown to provide significant gains in BMD compared with controls in males (Seabra, Marques, Brito et al., 2012), and in females when compared with a control group and a group assigned to a

duration-matched jogging protocol (Helge, Aagaard, Jakobsen, et al., 2010). A caveat to exercise is that, as mentioned previously, in female athletes where training causes reduced oestrogen levels and subsequent amenorrhea, BMD can be reduced. In males there is the potential for extreme activity (as observed in males running at least 64 km per week) to suppress testosterone production (Wheeler, Wall, Belcastro & Cumming, 1984), which may lead to acquired hypogonadism – a risk factor for osteoporosis (Foresta, Ruzza, Mioni et al., 1984, cited in Smith & Rutherford, 1993). In the vast majority of cases however, participation in regular physical activity is a pre-requisite for having the necessary physical attributes for military training selection. Swissa, Milgrom, Giladi et al. (1989) found no evidence to support participation in pre-training sports as a protective factor against stress fracture risk in Israeli recruits, although the same research group later examined pre-recruit training by activity. They found that participation in ball sports (primarily basketball) in the two years prior to selection lowered the risk of stress fractures (Milgrom, Simkin, Eldad, Nyska and Finestone, 2000), and repeated this finding recently (Finestone et al., 2011). The intermittent, multi-directional nature of this weight bearing activity was proposed to cause bone remodelling to the point of increased stiffness. Running training was not protective against injury in these male recruits, however women who ran fewer miles prior to military recruitment were shown to be at greater risk of stress fracture in a study by Winfield, Moore, Bracker & Johnson (1997). Discrepancies between classification of activities, reporting methods and reporting accuracy may confound findings relating to this topic. However, overall, research considering the effect of intermittent exercise on bone strength suggests that this may be protective against stress fracture risk.

One factor frequently implicated in the aetiology of overuse injuries is the hardness of the surface upon which activities are performed (Nigg & Yeadon, 1987). In soccer, there has been a trend for greater injury rates in pre-season training compared to in-season training (Woods et al., 2002) which has been linked with surface hardness during the summer months (Inkelaar, 1994, Orchard, 2002). Woods et al. (2002) also implicated hard surfaces in Achilles tendon injury risk, while Sullivan, Warren, Pavlov & Kelman (1984) found an association between training on hard surfaces and stress fracture occurrence in runners. It has been speculated that injuries sustained from training on hard ground result from increased impact forces (Cavanagh & LaFortune, 1980; Light, MacLellan & Klenerman, 1980), but while there is epidemiological evidence to suggest that injuries are more likely on harder surfaces, there appears to be a lack of clarity regarding a cause-effect relationship.

Kerdok, Biewener, McMahon, Weyand & Herr (2002), comparing running on surfaces of different stiffness, and found that a stiffer surface resulted in reduced metabolic cost (perhaps due to the increased resilience of the surface) and changes in lower limb kinematics. Greater energy return may be a desirable characteristic of a surface, and yet harder surfaces are associated with more injury. Stiles & Dixon (2006, 2007) looked at the human response to different surface properties when performing a tennis-specific movement and found that peak impact force was found to be lowest on the hardest of the cushioned surfaces that were tested. Nigg & Liu (1999) found that vertical impact peaks could be reduced by altering muscle tuning in the leg, therefore the observed changes in kinematics when runners accommodate a seemingly harder surface and thus reduce impacts, may be linked to injury causation

through altered lower limb loading. For example, Dixon et al. (2005) demonstrated that participants adjusted their ankle kinematics in order to maintain peak impact forces on two surfaces of differing hardness; Ferris, Liang & Farley (1999) and Fiolkowski, Bishop, Brunt & Williams (2005) have demonstrated changes in leg stiffness in response to stepping onto a new surface and changes in plantar sensation respectively.

The response to changes in the perceived hardness of footwear has shown similar effects (Milani, Hennig & LaFortune, 1997). Even where participant perception has not been assessed, changes in function have been observed. Nigg, Stefanyshyn, Cole, Stergiou & Miller (2003) observed changes in oxygen uptake and lower limb kinematics with different shoe types; Nurse, Hulliger, Wakeling, Nigg & Stefanyshyn (2005) experimented with the texture of a shoe insert, finding that alternating between a rough and a smooth insert significantly affected muscle activation of the tibialis anterior and gastrocnemius during their most active stages of stance. Direct links have not been made between the adjustments an athlete makes to accommodate different surface or shoe hardness and injury risk, however it is evident that changes in peak impact force, for example, will alter the loads acting on the lower limb. These changes in loading may represent a trigger to the remodelling process. During RM recruit training, it may be necessary to attempt to control for exposure to climatic conditions when investigating risk factors for stress fracture. For example, recruits who enrol in autumn may carry out more of their training on natural surfaces that are wet, muddy or frozen, whereas a recruit enrolling in early spring may face exposure to dry, hard surfaces for the majority of their training.

In reality, this could prove difficult to do well without logging exposure to specific surfaces and knowing the characteristics of these surfaces.

All recruits are issued with the same footwear, therefore the level of cushioning provided should be the same for all recruits, although their perception of the footwear may differ. Cushioning, either as part of the sole of the shoe or through the provision of an insole, serves to alter ground reaction forces both through the increased contact area between the shoe or surface and the plantar surface of the foot, or through increasing the time component of impulse and thus reducing the force component as the foot decelerates on impact with the ground. The increased dissipation of forces across the plantar surface of the foot has been demonstrated with the use of plantar pressure devices (e.g. Wegener, Burns & Penkala, 2008), while the influence on the time taken to decelerate the foot has been shown with reduced loading rates of peak impact (e.g. Aguinaldo, Arnel & Mahar, 2003). In addition to the cushioning component, other design aspects of footwear may influence lower limb kinematics and kinetics, and these have been investigated extensively in sporting contexts. With respect to overuse injury prevention, military footwear has received relatively little attention in the literature. This may be due to the requirement for combat boots to protect the foot from sharp or heavy objects and the subsequent limitation to possible design changes, or perhaps the lack of choice in, or opportunity to change boot models in military institutions.

The footwear provided to RM recruits is standardised: a combat assault boot (CAB) and a neutral, cushioned gym trainer (GT). Typical of military combat boots, the CAB features a stiff, hard midsole, raised heel and above-ankle

leather upper. Research suggests that a raised heel may reduce Achilles tendon strain (Dixon & Kerwin, 1999) but increase forefoot loading (Mandato & Nester, 1999); a stiff boot shaft has been shown to reduce ankle dorsiflexion in walking boots (Cikajlo & Matjacic, 2007; Böhm & Hösl, 2010), a functional variable potentially associated with increased forefoot loads (Hughes, 1985) and a relatively hard midsole may potentially increase plantar loading (Weigerinck et al., 2009). Study 1 of the current thesis investigates the influence of the CAB and GT on variables associated with MT3SF in greater detail, as these aspects of footwear design seem likely to increase forefoot loading. If aspects of the standard issue footwear – in particular the CAB, which is exclusively worn after week 8 of training – could be manipulated to reduce risk factors for injury, this would be of interest. It is not currently known which aspects of lower limb movement may predispose RM recruits to TSF or MT3SF, but the potential influence of footwear on variables hypothesised to relate to these injuries is discussed where relevant in the following sections.

2.4.1.2. Gender and age

The most commonly identified demographic risk factors for stress fracture within the literature are gender and age. Being female has frequently been identified as a risk factor for overuse injuries, particularly with regard to stress fracture (Ireland, 2002; Milner, Davis & Hamill, 2006). Gemmell (2002) discussed the recruitment policy of the British Armed Forces, reporting that under 'gender free' conditions – males and females are required to undergo the same level of training – the rate of discharge due to injury in females was over 11%, compared to 1.5% in men. This may be due to factors such as greater relative

load carriage, assuming females are lighter than males, however in gender specific training and athletic populations, stress fracture incidence is higher overall in females. Arendt, Agel, Heikes & Griffiths (2003) reported much higher rates of injury in females compared to males, while Knapik, Montain, McGraw et al. (2012) reported stress fracture incidence rates of 19 and 80 cases per 1000 recruits in males and females respectively. The studies of Beck et al. looking at stress fracture reports in male (1994) and female (2000) military recruits actually reported higher incidence of stress fracture in males, however this appears to be the exception to the rule. The overall picture highlights significantly greater stress fracture risk in females when subjected to the same training loads.

Factors specific to female anatomy and physiology are cited as potential risk factors for stress fracture. For example, Jones et al. (2002) reported studies identifying female runners with amenorrhea and irregular menses to have greater risk of stress fracture, a finding also supported by a study of dancers by Kadel, Teitz & Kronmal (1992). The incidence of stress fracture was reported to be up to 3.3 times greater in these groups, although some methodological issues were raised. Similarly, structural differences at the hip – women possess wider hips in relation to other structural components of their bodies (Sloane, 2002) – and differences in BMD (Bennell, Mathesen, Meeuwisse & Brukner, 1999) have been linked to altered loading of the lower limb and increased risk of injury (Sahrmann, 2002). An important study by Looker, Beck & Orwoll (2001) identified that when males and females were matched for age and body mass, males still possessed characteristics indicative of stronger bones. Although only males are considered in the current thesis, the mechanisms for stress fracture

identified in females provide an indication that altered loading and bone characteristics are of importance in determining stress fracture risk.

Age has been identified as a risk factor for stress fracture at both ends of the spectrum. Bone undergoes a process of reduced regeneration with increased age which leads to reduced bone mass and reduced ability to cope with applied loads (Schaffler & Jepsen, 2000; Ural & Vashishth, 2007). Jones et al. (2002) identified a selection of studies reporting higher incidence of stress fracture in older military recruits, the effect presenting from the age of 21 and increasing susceptibility with each year of age (Shaffer, Brodine, Almeida et al., in Jones et al., 2002). There is also evidence in non-military settings linking increased age with increased risk of lower limb overuse injuries (Taunton, Ryan, Clement et al., 2002).

Growing bone is also more vulnerable to loading due to the presence of open physes (growth plates). There is evidence that adolescence is a risk factor in stress fracture, with damaging strain at the apophyses where tendons attach (e.g. heel, knee, elbow) most common (Adirim & Cheng, 2003). These sites are typically vulnerable due to the ossification process occurring at the ends of the bone, which is also where tendons attach. If high activity levels occur at the same time as growth periods, then apophysitis (inflammation of the apophysis) may occur (Soprano & Fuchs, 2007). Although these conditions are particularly common during puberty, bone may continue to grow until the mid-twenties and thus this may be a factor to consider when dealing with individuals of this age. The findings of Milgrom et al. (1994) suggest that each year of age above 17 at entry into training decreased the risk of stress fracture by 28% in Israeli military

recruits, again suggesting a link between age and bone strength. The findings of Milgrom et al. (1994) appear to contradict those reviewed by Jones et al. (2002), particularly as this statistic is applied to a cohort of recruits aged between 17 and 26. The predictive value of the logistic regression applied by Milgrom et al. (1994) may have been skewed by the fact that they saw no TSF in any recruit aged over 19 years. In the Israeli recruit population that was studied, these results point to a clear effect of age, however there was also an unusually high (24%) rate of TSF, which is not comparable to other populations. These results may therefore need to be interpreted with caution. It was also interesting to note that none of the 190 cases of TSF were shown to have open femoral or tibial physes, despite their age. The exclusive occurrence of TSF in recruits aged under 20 was attributed to the bones of these recruits not reaching full structural maturity: although the length of long bones may not be increasing, full cortical thickness does not occur until years later. RM recruits may be aged between 16 and 33 upon entry into training (www.royalnavy.mod.uk) therefore those at the lower end of this scale in particular may still have bones that are developing. Research into both gender and age as risk factors for stress fracture suggest that the quality of bone is an important factor in injury susceptibility.

2.4.1.3. Bone characteristics

The resistance of bone to bending, or its strength, is of interest because this essentially defines the load required to cause fracture. Determinants of bone strength may be considered as markers of its 'quality', and include consideration of its structural and material properties (Felsenberg & Boonen,

2005). Structural components include the geometry, alignment of fibres and cortical porosity, while material components include mineralisation, collagen composition and the level of microdamage present (Felsenberg & Boonen, 2005). Although BMD (the amount of mineral matter per square centimetre of bone) is commonly referred to as an index of bone strength, it is not the only feature which determines the bone's resistance to load. For example, Martin & Ishida (1989) found that the alignment of fibres was the strongest predictor of strength in primary cortical bovine bone samples. Mechanically, the distribution of bone mass has the potential to have a greater influence on strength than BMD alone (Felsenberg & Boonen, 2005).

BMD is calculated with relative ease using dual-energy x-ray absorptiometry (DEXA) scans of the wrist or femur, making it a popular reference tool. More accurate 3D estimates of volumetric BMD have recently been championed (Popp, Hughes, Smock et al., 2009), increasing its validity. There is also research to support a link between low BMD and stress fracture risk. In osteoporosis sufferers, a disease that leads to lower BMD, bones have been shown to be weaker (Hodgkinson & Currey, 1993) and research by Carter et al. (1981) links low BMD with more rapid microcrack propagation, therefore while BMD is not the sole determinant of bone quality, it is a common reference point when considering the strength of bone material.

Bone geometry may be estimated with scanning techniques such as magnetic resonance imaging (MRI) or computerised tomography (CT). Franklyn, Oakes, Field, Wells & Morgan (2008) estimated tibial geometry using measurements obtained from radiographs. Anthropometric measures may also be used to

estimate frame size, such as bimalleolar width (Himes & Bouchard, 1985). A number of measures of bone geometry may be considered including: the cross-sectional area; inertial properties; the section modulus; and bone width (Jones et al., 2002). The cross-sectional area of a bone determines its axial strength. The cross-sectional area moment of inertia is determined by the distribution of bone mass about axes perpendicular to the centroid, which is the neutral plane running through the geometric centre of any cross-section of bone. The resistance to sagittal or frontal plane bending can be determined by the distribution of mass about an axis relative to the centroid. The polar moment of inertia is the resistance to torsional bending and is determined by the distribution of mass about the longitudinal axis of the bone. The section modulus considers the maximum distribution of mass away from the centroid and is an indication of true bending strength (Franklyn et al., 2008). These components of bone geometry are independent of bone density or mineralisation, and highlight the complex nature of determinants of bone strength.

Franklyn et al. (2008) found a number of components of tibial geometry that differed both between males and females, and between those with and without TSF, including tibial width and section modulus. The prospective work of Beck et al. (1996, 2000) on military recruits measured components of bone geometry and compared those who sustained stress fractures with matched controls. In females (Beck et al., 1996), cortical thickness was lower at the most common site of stress fracture, a finding supported by Popp et al. (2009). In male military recruits (Beck et al., 2000) injuries were sustained by recruits with narrower tibiae, while cortical thickness and BMD were not significantly different. In

addition to the direct comparison of specific geometric properties, modelling studies have used multiple properties to estimate bending stress acting on a bone. A study by Gross & Bunch (1989) is an example of this approach being used at the metatarsals. These authors simplified the metatarsal cross-section as a hollow ellipse and estimated bending strains using reference data for bone geometry. Milgrom, Giladi, Simkin et al. (1989) used x-rays of Israeli military recruits to calculate the area moment of inertia at the narrowest point of the tibia. Their study revealed that those with a low area moment of inertia in the anterior-posterior direction in particular were at greater risk of TSF. Evidence supports the notion that either individual or combined components of bone geometry can influence bone strength, regardless of BMD.

There is evidence for the importance of either BMD (e.g. Hui, Slemenda & Johnston Jr, 1988; Myburgh, Hutchins, Fataar, Hough & Noakes, 1990; Bennell et al., 1996; Lauder, Dixit, Pezzin et al., 2000; Valimaki et al., 2005) or bone geometry (Giladi, Milgrom, Simkin et al., 1987a; Milgrom et al., 1989) in stress fracture risk in populations of military recruits and runners. Furthermore, evidence from Crossley, Bennell, Wrigley & Oakes (1999) indicates that tibial geometry is dominant over BMD in determining TSF risk in male runners. It is evident that both BMD and bone geometry are important determinants of bone strength, and that bone strength is a risk factor for stress fracture.

Given the importance of bone strength, those non-biomechanical factors (of which age and gender are two) which have been shown to influence the quality of bone should not be ignored. Amenorrhea or oligomenorrhea have been shown to lower BMD (Cook, Harding, Thomas et al., 1987; Myburgh et al.,

1993), contributing to greater risk of stress fracture in females. It has been suggested that inadequate calcium intake may increase the risk of stress fracture (Philipson & Parker, 2009) although this is not widely supported by research with a number of studies showing no correlation between calcium intake and stress fracture risk (Mustajoki, Laapio & Meurman, 1983; Schwellnus & Jordaan, 1992; Cline, Jansen & Melby, 1998). Low energy intake has however been linked with stress fracture risk. During physically demanding military training programmes, it is important to replenish lost energy, nutrients and minerals associated with high workload. Armstrong et al. (2004) concluded that an acute negative energy balance (more energy being used than taken up) may contribute to rapid weight loss in military recruits. This may lead to increased muscular fatigue and reduced bone collagen synthesis, which in turn may expose the bone to greater loads and reduce its capacity to withstand them. Smoking may also increase the risk of stress fracture by reducing bone quality. Altarac, Gardner, Popovich et al. (2000) studied 2002 Army recruits over an eight week basic training programme, reporting an overuse injury incidence 8% greater in men and 11% greater in women that had recently smoked (at least one cigarette in the month prior to training). The exact mechanism of injury is not well understood, but it is suggested that smoking hinders tissue repair, and that smokers exhibit greater risk-taking behaviour than non-smokers. Jorgensen, Kallehave, Christensen, Siana & Gottrup (1998) reported that collagen production is lower in smokers, which may have an implication in the repair cycle of tendinous tissues. Additionally, Kawakita, Sato, Makino et al. (2008) found that nicotine acts directly on growth plates to reduce bone growth. This may have implications on the repair cycle of bone, potentially

increasing the remodelling period and thus the time when bone is most vulnerable to sustaining damage from repeated loads.

Several non-biomechanical risk factors presented here may influence the quality of bone and thus increase the risk of stress fracture, and should therefore be considered or, if possible, controlled in any populations where stress fracture aetiology is being investigated. Having considered broader factors that may influence the risk of lower limb stress fracture in general, the following section discusses literature evidence regarding biomechanical characteristics specifically associated with tibial and MT3SF.

2.4.2. Anatomical characteristics associated with tibial and metatarsal stress fractures

Due to the relative proliferation of TSF affecting runners and military recruits, there is a large body of research on the topic. However, evidence remains equivocal on the mechanisms most strongly associated with the injury, and is confused by a number of confounding variables. Both passive anthropometric characteristics and gait variables have been linked with TSF risk, and will be explored in detail. MT3SF have received considerably less attention in the literature, and therefore knowledge regarding mechanisms for the injury is limited. A number of biomechanical characteristics including anthropometric, flexibility and dynamic gait variables have been identified as risk factors for these injuries, and are considered below.

2.4.2.1. Arch height

Extremes of medial longitudinal arch height, as assessed statically, have been associated with lower extremity injury risk (Cowan, Jones & Robinson, 1993; Williams, McClay, Hamill & Buchanan, 2001) with either a low arch (pes planus) or high arch (pes cavus) (Figure 2.4) suggested as a risk factor. Any discussion which relates absolute arch height to stress fracture risk is confounded by the variety of methods employed for quantifying it, as reviewed by Williams & McClay (2000). Methods vary from qualitative to quantitative, and weight bearing to non-weight bearing assessments, however the Williams & McClay (2000) review did ascertain that most quantitative measures are suitable when taken in 10% weight bearing. Usually arch height is quantified by measuring the distance of the navicular from the ground, and a greater order of validity is achieved by normalising this value to truncated foot length (without toes) (Williams & McClay, 2000).

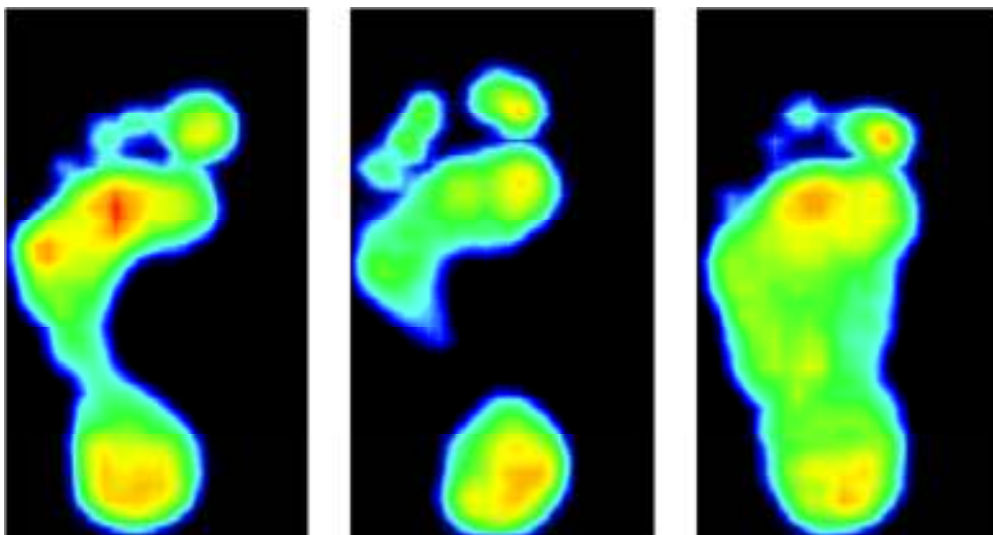


Figure 2.4. Example plantar outlines of (left to right) normal, high arched (cavus) and low arched (planus) feet.

Studies in military populations have provided conflicting results regarding arch height and injury risk. A number of studies have found no relationship between arch height and stress fracture incidence (Montgomery, Nelson, Norton & Deuster, 1989; Ekenman, Tsai-Fellander, Westblad, Turan & Rolf, 1996; Kaufman et al., 1999; Esterman & Pilotto, 2005), although methods of assessment and study designs have varied significantly amongst these examples. Where positive associations have been found, they provide an inconsistent story. Simkin, Leichter, Giladi, Stein & Milgrom (1989) used a highly accurate x-ray assessment of arch index and found that military recruits with high arches were at greater risk of femoral and TSF, with individuals with low arches at risk of metatarsal stress fractures. In contrast, Wen, Puffer & Schmalzried (1998) suggested that the pes cavus foot is protective against injuries. Research examining both high and low arch heights has observed that each is associated with different injuries. Sun, Shih, Chen et al. (2012) produced a finite element model of high, low and normal arched feet, suggesting that a high arched foot increased strain on the metatarsals, while a low arched foot produced greater strain on the rear- and mid-foot structures. Williams et al. (2001) prospectively followed up 20 high arched and 20 low arched runners, finding that high arched runners suffered more laterally located injuries which affected the ankle and bony structures, whereas low arched runners sustained knee injuries, soft tissue complaints and were generally afflicted by injuries in medial locations. This epidemiological evidence points to the effect of arch height on foot function. A high arched foot is associated with reduced pronation of the foot and subsequent loss of shock absorption, as well as greater relative internal rotation of the tibia (Nigg, Cole & Nachbauer, 1993).

A low arched foot is said to promote 'excessive' pronation, potentially leading to increased loading of foot structures, and increased demand on muscles such as the anterior and posterior tibialis muscles to provide stability (DeLacerda, 1980).

Despite some consistencies in the suggested mechanisms by which extremes of arch height may contribute to injury risk, an absence of direct correlations between static measures of arch height and dynamic foot function may explain some of the inconsistent reporting of arch height as a predictor of stress fracture. Plantar pressure analysis has been used to explore relationships between foot type and dynamic foot function (Cavanagh, Morag, Boulton et al., 1997; De Cock, Willems, Witvrouw, Vanrenterghem & De Clercq, 2006). Previous research has used plantar midfoot surface contact to investigate the relationship between foot type and lower extremity injury (Michelson, Durant and McFarland, 2002), with evidence to support a decreased midfoot contact area in pes cavus feet (Arther, Burns, Henteges et al., 2007) and increased midfoot contact area in participants with pes planus feet (Morag & Cavanagh, 1999). Cavanagh et al. (1997) reported that 35% of the variance in dynamic plantar pressure was explained by static measures, the strongest predictor being arch related measurements, suggesting a relatively weak association between static and dynamic indicators of arch height. Furthermore, Nigg et al (1993) found no significant association between arch height and maximal eversion angle, although 27% of the rearfoot-tibial coupling movement was explained by arch height. Kaufman et al. (1999) is the only study available to assess arch height statically (using the navicular drop test) and dynamically (using plantar pressure), during walking. The correlation between the two methods (static assessment versus barefoot dynamic assessment) was low ($r =$

0.22 to 0.24) but significant. Further information on the validity of static arch height measurements would be provided if they were compared with dynamic arch data obtained for running, however this represents a gap in the literature at present.

In summary, although associations between arch height and risk of stress fracture have been drawn, clear cause-effect relationships are absent from the literature. Discussion of these relationships has been clouded by the inconsistent use of techniques to quantify arch height, and a lack of association between static measurements and dynamic function.

2.4.2.2. Passive hip range of motion

Amongst numerous studies published relating to the aetiology of stress fractures in the Israeli Defence Force, Giladi et al. (1987b) detected a strong association between the passive measure of external hip rotation and the risk of stress fracture. Recruits in this study who were within the highest category of external hip rotation (>65 degrees) were twice as likely to incur TSF during training. Giladi, Milgrom, Simkin & Danon (1991) confirmed these findings, again in Israeli military recruits. In the study by Giladi et al. (1991) a mean difference of just 2.6 degrees was found between the mean external hip rotation values for TSF cases and controls, perhaps limiting the 'real world' application of this finding. Wider evidence of a link between high passive external hip rotation and TSF is lacking in both military and athletic populations (Montgomery et al., 1989; Bennell et al, 1996; Lun, Meeuwisse, Stergiou & Stefanyshyn, 2004), suggesting that these findings may be unique to the Israeli Defence Force.

Some association between passive hip range of motion and other lower limb injuries (such as exertional medial tibial pain) has been identified (Burne, Khan, Boudville et al., 2004; Moen, Schmikli, Weir et al., 2010), however the summary of reported values of external hip rotation in Table 2.1 highlights the difficulty in drawing conclusions regarding this risk factor. The link between passive hip rotation and stress fracture risk is further clouded by an absence of studies relating this variable to dynamic function. Passive measurement techniques aim to isolate hip rotators, and therefore there may be no correlation with dynamic function when numerous muscles are activated simultaneously. Variations in the method of assessment lead to disagreement in the reported values, for example there will be differences in muscular activation and femoral head position between positions utilising a flexed or extended hip. There is also subjectivity in how far the assessor pushes the hip when feeling for 'firm resistance', as is common practice in the assessment. The intra-tester reliability of hip measurement with the patient supine and the hip and knee flexed at 90 degrees has been assessed by Burne et al. (2004) and Nussbaumer, Leunig, Glatthorn et al. (2010). Burne et al. (2004) report intra-class correlation coefficients (ICCs) between 0.75 and 0.89, while Nussbaumer et al. (2010) reported ICCs of 0.91 and 0.93 depending on measurement technique. The latter study also identified possible over-estimation of hip joint range of motion using a goniometer, when compared to an electronic tracking system, due in part to difficulty placing the goniometer with good anatomical alignment.

Despite inconclusive findings regarding hip range of motion as a predictor of stress fracture, it is commonly assessed in studies examining TSF aetiology. Further investigation of this variable is required, including comparisons between

measurement techniques and the effects of range of motion on dynamic gait variables or lower limb loading.

Table 2.1. Summary of external hip rotation measurement techniques and reported values.

| Study | Population | Method | Mean values (degrees) |
|---------------------------|-----------------------------------|--|-----------------------|
| Giladi et al. (1987b) | Israeli male military recruits | Supine, hip at 90 degrees flexion | 57.0 |
| Montgomery et al. (1989) | US male military recruits | Prone, hip at 0 degrees flexion, knees at 90 degrees | 24.1 |
| Giladi et al. (1991) | Israeli male military recruits | Supine, hip at 90 degrees flexion | 55.9 (200 controls) |
| Korpalainen et al. (2001) | Finnish athletes | Not stated | 42.0 |
| Burne et al. (2004) | Australian male military recruits | Supine, hip at 90 degrees flexion | 53.0 |
| Lun et al. (2004) | Recreational runners | Sitting, hip at 90 degrees flexion | 41.1 |
| Moen et al. (2010) | Dutch male military recruits | Supine, hip at 90 degrees flexion | 56.0 |

2.4.2.3. Passive dorsiflexion

The amount of passive dorsiflexion available has been linked with MT3SF risk, although as with previously mentioned static measurements, there is very little literature exploring direct associations with dynamic gait function. Dorsiflexion is any movement which brings the dorsal surface of the foot to a position more proximal to the anterior aspect of the shank. Measurement of passive dorsiflexion is generally performed either by manipulation of the lower limb with the subject lying down, or through a weight-bearing lunge method. Hughes (1985) and Lun et al. (2004) utilised the non-weight bearing assessment

technique, however factors such as the strength of the assessor and placement of goniometric devices have led to reports of varying degrees of reliability for this method (Elveru, Rothstein & Lamb, 1988; Meyer, Werner, Wyss & Vienne, 2006). The weight-bearing lunge test has been rated to have high reliability (Bennell, Talbot, Wajswelner & Kelly, 1998), with an intra-rater intraclass correlation coefficient (ICC) value of 0.97 for dorsiflexion angle, and is therefore the preferred method of assessment.

In the case of MT3SF, highly restricted ankle flexibility, to the extent that less than 10 degrees of dorsiflexion is available (equinus deformity), has been linked with increased forefoot pathology in diabetic populations (Lin, Lee & Wapner, 1996). Hughes (1985) found that military recruits with equinus deformity were 4.6 times more likely to develop metatarsal stress fracture than those with more than 10 degrees of dorsiflexion. It was suggested that this restricted range of motion would cause an earlier heel lift and subsequent increases in the duration and magnitude of forefoot loading during locomotion. This mechanism has a logical basis, but has not been confirmed by subsequent research in healthy populations, despite a number of studies assessing this variable.

In an ageing population of patients with equinus deformity Orendurff (2006) observed only a weak association between highly restricted passive dorsiflexion and forefoot pressures during walking. When comparing diabetic sub-groups with varying levels of foot pathology and a control group of non-diabetics, Turner, Helliwell, Burton & Woodburn (2007) found restricted passive dorsiflexion in the diabetic populations did not correlate with either dynamic dorsiflexion or forefoot plantar pressure magnitudes during walking. Johansen,

Wooden, Catlin et al. (2006) found that a stretching protocol increased passive dorsiflexion but had no effect on dynamic range of motion. This evidence suggests that passive dorsiflexion is not linked with forefoot pressure, however these studies have only investigated data when walking. Research investigating the relationship between passive dorsiflexion and forefoot pressures when running would provide a greater insight into whether this static variable influences stress fracture risk.

Any investigation into the relationship between passive dorsiflexion and forefoot pressures should consider whether any adjustments in foot function occur as a result of restricted passive range of motion. For example, compensatory adjustments such as the altered timings of rearfoot movements have been demonstrated to occur in response to restricted dorsiflexion (Cornwall & McPoil, 1999). Any compensatory mechanisms that occur may in turn place stress on other musculoskeletal structures, thereby influencing the risk of overuse injury.

There is no published evidence of a link between restricted passive dorsiflexion and increased risk of TSF. Prospective studies by Bennell et al. (1996) and Yagi, Muneta & Sekiya (2012) have found no association between passive dorsiflexion range of motion and TSF risk in runners. Additionally, Korpelainen, Orava, Karpakka, Siira & Hulkko (2001) stated that passive dorsiflexion was not implicated as a risk factor for individuals sustaining multiple stress fractures. Kaufman et al. (1999) did find a positive association between restricted passive dorsiflexion and injury in naval recruits, but not specifically for stress fracture. The role of dynamic ankle dorsiflexion is considered of greater importance to TSF risk, as discussed below.

2.4.3. Dynamic gait variables

A range of dynamic variables may be quantified during gait in order to provide indications of lower limb loading. These include tri-planar rotations of joints and segments, GRF, plantar pressure variables and joint moments. Of the many variables of this nature available for analysis, those previously associated with the aetiology of tibial and MT3SF, and will be discussed below.

2.4.3.1. Ground reaction forces and plantar pressure

The magnitude and loading rate of peak impact force has been frequently examined as a factor in TSF aetiology, possibly influenced by the ease with which data can be obtained for large groups of participants. Peak impact force usually occurs within the first 0.05 s following ground contact (Nigg, Cole & Bruggemann, 1995), and therefore before appropriate active muscular response can occur to attenuate it (Wright, Neptune, van den Bogert & Nigg, 1998). Greater loading rates may indicate a force that is larger or more rapidly applied to the lower limb without the benefit of active attenuation provided by other tissue, therefore, loading rate is of interest as well as impact force. Given the 'passive' nature of this impact force, a seemingly logical relationship between the magnitude of this value and the risk of lower limb injury might be expected, however research remains equivocal on this association. Although some studies have identified greater magnitudes of vertical GRF and/or loading rate in those sustaining stress fractures when compared to controls (e.g. Grimston, Engsberg, Kloiber & Hanley, 1991; Ferber, McClay Davis, Hamill,

Pollard &McKeown, 2002), there are also a number of studies finding no difference (Bennell, Crossley, Jayarajan et al., 2004; Bischof, Abbey, Chuckpaiwong, Nunley &Queen, 2010) or significantly lower GRF variables in the stress fracture group (Grimston, Nigg, Fisher &Ajemian, 1994). In a review of thirteen relevant papers by Zadpoor &Nikooyan (2011) it was concluded that literature fails to consistently implicate the magnitude of vertical GRF in stress fracture risk, yet their meta-analysis suggested that loading rate was significantly greater in stress fracture cases compared to controls.

One proposed role of vertical impact peaks is as a sensory input device to which muscular activation levels are tuned (Nigg &Wakeling, 2001). This theory postulates that, for example, the observed changes in kinematics in response to shoe/surface hardness arise due to an innate desire to maintain peak impact force magnitudes, presumably to a level that is either comfortable or undamaging. For example, recent research has demonstrated the possibility of reducing impact peaks with real-time visual feedback whilst running on an instrumented treadmill (Crowell, Milner, Hamill &Davis, 2010), highlighting the ability of the body to achieve changes in impact force magnitudes through kinematic adaptations. If external shoe/surface hardness predisposes to high impact peaks, it may be assumed that the increased requirement to adjust or increase muscle activation in an attempt to reduce loads could lead to increased risk of overuse injury. If increased muscular demand leads to fatigue, bones may be exposed to greater or altered loading without muscular attenuation of forces, potentially accelerating the propagation of microdamage and therefore increasing stress fracture risk.

The association of vertical force loading rate with injury is perhaps even less well understood. Schaffler et al. (1989) showed that fatigue failure occurred earlier in bone samples when loads were applied at greater strain rates, suggesting a negative association with injury risk; however it is also suggested that intermittent high loading rates are necessary for bone remodelling (O'Connor, Lanyon & MacFie, 1982) and may be more influential than simply the number of loading cycles in developing mass (Rubin & Lanyon, 1984). To confirm this suggestion, Mosley & Lanyon (1998) demonstrated a profound effect of strain rate on the osteogenic response to load of rat bones. Under the same magnitude of load, bones subjected to greater strain rate experienced 54% greater bone mass increase than those subjected to moderate strains, and 67% more than the low strain rate group. Given this relationship, it might be expected that high loading rates sustained during vigorous exercise would trigger the remodelling response and subsequent training errors could lead to stress fracture development, perhaps explaining the stronger evidence for loading rate as a risk factor in TSF aetiology (Zadpoor & Nikooyan, 2011).

Plantar pressure analysis has shown that maximum vertical metatarsal loading occurs in mid stance (De Cock, De Clercq, Willems & Witvrouw, 2005), when peak active force occurs, yet there has been no research conducted to investigate whether an association between the magnitude of this force and risk of MT3SF exists. In addition to the requirement to investigate the influence of vertical active force on metatarsal stress fracture risk, the work of Arangio et al. (1998) may suggest an alternative direction for analysis, as their model of strain on the metatarsals revealed that the third metatarsal is most vulnerable to laterally applied forces. Despite this evidence, the role of lateral forces on

MT3SF has received very little attention in the literature. Only one study has reported horizontal force characteristics in relation to MT3SF risk. In subjects with and without a history of MT3SF, Dixon, Creaby & Allsopp (2006) noted that the horizontal braking force vector was applied significantly more laterally, and the magnitude of the horizontal force vector was lower during the propulsive phase of stance in injury cases. The evidence from these studies suggests that further investigation of the role of horizontal GRF and forefoot function during the braking and propulsion phases of stance is warranted.

Although risk factors for MT3SF are poorly understood overall, recent advances in plantar pressure analysis has furthered understanding of forefoot loading. Plantar pressure analysis permits the user to examine the distribution of vertical loads on the plantar surface of the foot, with the metatarsal heads providing a distinct area of interest in injury pathology research. External GRF are applied at the metatarsal heads, therefore determination of the magnitude of these forces may provide insight into metatarsal loading. The landmark paper by Arndt et al. (2002) provided evidence that plantar pressure measurements, particularly force-time integrals, give a strong indication of metatarsal deformation (strain), thus providing an appropriate tool in understanding the pathology of this structure.

The third metatarsal has been shown to be weakest under horizontal loads, however there is a lack of understanding of the pathogenesis of stress fractures in this bone and vertical loads beneath the head of the bone may also provide information on injury risk. Several studies have investigated metatarsal loading during locomotion using measurements of plantar pressure, however their

common failure to consider the third metatarsal individually has contributed to the lack of understanding of its function during gait. Consideration as part of the whole forefoot (e.g. Stewart, Gibson & Thomson, 2007), central forefoot (e.g. McPoil, Cornwall, Dupuis & Cornwell, 1999; Chuckpaiwong, Nunley, Mall & Queen, 2008; Weigerinck, Boyd, Yoder et al., 2009), medial or lateral forefoot (e.g. Che, Nigg & De Koning, 1994; Tessutti, Trombini-Souza, Ribeiro, Nunes & Sacco, 2010); grouping with the second metatarsal (e.g. Rosenbaum, Hautmann, Gold & Claes, 1994; Weist, Eils & Rosenbaum, 2004; Bisiaux & Moretti, 2008), or with the fourth and fifth metatarsals (e.g. Perry, Ulbrecht, Derr & Cavanagh, 1995; Arndt et al., 2002; Arndt et al., 2003; Nagel, Fernholz, Kibele & Rosenbaum, 2008) has hindered knowledge of the third metatarsal's individual function. Studies by Pollard, Le Quesne & Tappin (1983), Tappin & Robertson (1991), Arangio et al. (1998), Hosein & Lord (2000) and more recently, Griffin & Richmond (2005), demonstrate clearly the individual metatarsals not only experience a range of different load magnitudes during stance, but they are also vulnerable to loads from different directions, applied at different locations. Despite its relative paucity, the available research suggests some potential leads in the quest to understand metatarsal pathogenesis. For example, forefoot pressures have been shown to increase significantly following fatigue (Arendt et al., 2002; Weist et al., 2004; Nagel et al., 2008) and with increased midsole hardness (Wiegerinck et al., 2009). Despite these findings, there is limited evidence regarding the mechanisms of overuse injuries to individual metatarsals.

GRF and plantar pressure variables provide an indication of loads acting on the structures of the lower limb. In the following discussion of kinematic variables

associated with tibial and MT3SF, the effect of each upon loading of the lower limb is usually quantified using one of the above methods. Therefore, where loading of structures is inferred from force or pressure data, the limitations of such data should be considered.

2.4.3.2. Subtalar joint motion

The ankle joint consists of the tibiotalar and talocalcaneal (or subtalar) joints, which facilitate movement with six degrees of freedom. The tibiotalar joint is a hinge joint enabling dorsi/plantar flexion of the foot; the subtalar joint is an inclined hinge joint lying just inferior to this, enabling the tri-planar motions of pronation and supination (Root, Weed, Sgarlato & Bluth, 1966). Weight bearing activities inevitably lead to the creation of moments acting about the subtalar joint axis, and therefore variations in its alignment may have a profound effect on dynamic foot function. For example, medial longitudinal arch height may affect the inclination of the subtalar joint axis and subsequently the relative internal rotation of the shank to the foot (Nigg et al., 1993). Additionally, any medial or lateral deviation of the line of the subtalar joint axis will influence the moments tending to cause pronation and supination, which in turn will affect the demand on musculature and the dynamic behaviour of the foot during stance (Kirby, 1989). It is the unique position of the subtalar joint which facilitates tri-planar motion including inversion/eversion in the frontal plane, ab/adduction in the transverse plane and plantar/dorsiflexion in the sagittal plane. Pronation is a tri-planar motion, incorporating eversion, abduction and dorsiflexion of the foot relative to the shank, while supination incorporates inversion, adduction and

plantarflexion, however the movements of calcaneal inversion/eversion are often used to quantify so-called 'rearfoot movement'.

Pronation/eversion is something of a 'buzz word' in overuse injury discussions, frequently receiving negative attention in the media and commonly investigated in aetiological research. However it is a necessary function of gait along with supination/inversion, which allows humans to adapt to uneven surfaces, as well as playing a role in cushioning of GRF. There is research to support and refute claims linking rearfoot variables with TSF in particular. Variables of interest are usually the rearfoot angle at heel strike, peak eversion angle, timing of peak eversion and rate of eversion. There is considerable range in the values considered as 'normal' for peak eversion, and indeed in the range of reported typical angles in the literature, however 'excessive' eversion or 'overpronation' are often cited as risk factors for overuse injury. Injuries such as medial tibial stress syndrome, compartment syndrome and exercise-induced lower leg pain, which affect soft tissues, are proposed to lie on the earlier end of the same bone stress-response continuum as stress fractures (Anderson et al., 1997), although further evidence is required to support this. A large volume of evidence has been presented regarding the role of rearfoot motion in the development of these lower limb injuries, which may provide insight into the development of stress fractures. However, while there may be links between overuse injuries affecting the shin and the development of TSF, these are not fully understood and therefore related mechanisms should be interpreted with caution.

Retrospective analyses of the role of rearfoot motion in stress fracture aetiology have been more common than their prospective counterparts, however there

are some consistencies between reports. Greater peak eversion has been reported in runners with a history of TSF (Pohl et al., 2008; Milner, Hamill & Davis, 2010); with altered loading proposed as a suggested injury mechanism. Hetsroni, Finestone, Milgrom et al. (2008) assessed subtalar joint kinematics during treadmill walking two weeks prior to training in a cohort of 473 infantry recruits. These authors identified a longer duration of pronation as a protective factor against tibial and femoral stress fractures. The use of a walking assessment in this study is a potential limitation, with evidence suggesting that the magnitude of rearfoot eversion excursion may vary between the two modes of locomotion (Pohl, Messenger & Buckley, 2007; Morio, Lake, Gueguen, Rao & Baly, 2009). However there has been some wider evidence of the protective value of pronation against injury (Hreljac, Marshall & Hume, 2000; Stefanyshyn, Stergiou, Lun & Meeuwisse, 2001).

It is significant to note that, other than Hetsroni et al. (2008), there is no evidence of a relationship between rearfoot motion and TSF incidence in military recruits. Factors other than rearfoot motion may play a dominant role in determining tibial loading, or military footwear may restrict frontal plane movements compared to running shoes or other athletic footwear. Understanding is restricted because to date there have been no prospective studies on military populations which have considered dynamic running gait variables at the start of training and assessed their contribution to stress fracture risk. This suggests that the relationship has not been properly explored, and prospective analysis of the contribution of rearfoot movement to TSF risk is warranted in military populations.

The only evidence regarding the role of rearfoot movement in MT3SF aetiology identified the earlier occurrence of peak eversion in RM recruits with a history of this injury when compared to controls (Dixon et al., 2006). These authors proposed that this trait may cause earlier loading of the forefoot, however further research is necessary to improve understanding of this potential mechanism. For example, an area of interest may be to examine links between limited passive dorsiflexion and compensatory eversion during mid-stance.

2.4.3.3. Dynamic ankle dorsiflexion

As discussed above, it has been proposed that limited passive dorsiflexion may increase the risk of metatarsal stress fracture. Although there are no studies providing evidence of clear links between passive and dynamic function, a logical expectation that restricted passive dorsiflexion would translate to restricted dynamic function, leading to earlier heel off and greater forefoot loading has been suggested (Hughes, 1985). This association has led to this variable being assessed in the two papers to have examined the influence of ankle kinematics on metatarsal stress fracture risk. Bischof et al. (2010) found no difference in any kinematic variables, including ankle dorsiflexion, between female runners with a history of metatarsal stress fracture and matched controls. To date, the only other study to examine dynamic ankle dorsiflexion and its association with metatarsal stress fracture is the study by Dixon et al. (2006), who also found no significant difference between previously injured recruits and matched controls. In both studies, the mean peak dorsiflexion was lower for injury cases than controls, but not significantly. It must be concluded that research does not point to an association between dynamic ankle

dorsiflexion and increased risk of MT3SF, although this area has received little attention. In particular, there are no prospective studies examining this association.

With regard to TSF, the influence of ankle dorsiflexion on tibial loading is not fully understood. There is no evidence linking peak dynamic dorsiflexion values with TSF risk, however it is suggested that the ankle angle at touchdown may contribute to impact attenuation. Gerritsen, van den Bogert & Nigg (1995) demonstrated the propensity for increased plantarflexion at ground contact to reduce peak impact forces using a simulation model, while Dixon et al. (2005) observed significant reductions in dorsiflexion angle at touchdown as a strategy to reduce peak impact forces when running on harder surfaces. Although a reduction in impact magnitude may be considered a beneficial side effect, Dixon et al. (2005) built on the earlier suggestions of De Wit, De Clerq & Aerts (2000), who suggested that compensatory adjustments at ground contact were made to reduce local heel pressures. Recent studies by Lebieadowska, Wente & Dufour (2009) and Duquette & Andrews (2010) have isolated the effect of ankle dorsiflexion on impact attenuation. Lebieadowska et al. (2009) found that ankle position significantly influenced impact characteristics in hopping and walking tasks, while Duquette & Andrews (2010) investigated the role of dorsiflexion angle on impact attenuation in a fatigued state. These authors found that local fatigue of lower limb muscles had a greater role than dorsiflexion angle in impact attenuation, implying that the influence of fatigue may complicate the interpretation of the role of ankle kinematics in impact attenuation.

Ankle dorsiflexion at ground contact is an important variable to consider in TSF aetiology, as it has the potential to influence peak impact forces. It is therefore also important to control for factors which have been demonstrated to influence this variable, such as surface hardness.

2.4. Summary and thesis direction

Lower limb stress fractures are a significant problem in RM recruit basic training, leading to lengthy absence from training and high rehabilitation costs. The most frequently affected sites in this population are the third metatarsal and tibia. The frequency of MT3SF in particular is of interest and is unique to this population, indicating that certain aspects of training are linked to this. Potentially relevant unique aspects of RM training are the duration of the programme and type of footwear issued to recruits. Although longer training programmes have been associated with increased prevalence of metatarsal injuries, the influence of standard issue RM footwear is unknown and may play a significant role in forefoot loading.

MT3SF aetiology is very poorly understood. There are few studies to have considered this bone individually, despite research suggesting that it is vulnerable to different loading than its neighbouring metatarsals, particularly laterally directed forces. Arguably the greatest gains in knowledge relating to this injury have come from modelling studies, and with the availability of plantar pressure analysis and scanning techniques such as MRI, it should be possible

to assess the geometry of the bone and the location of external loads relative to the point of application on the distal plantar surface of the metatarsal. This could contribute significantly to knowledge of metatarsal loading, and given the flexibility of plantar pressure analysis devices, the estimation of metatarsal loading in various shod conditions may be possible.

There is a relative abundance of literature relating to TSF aetiology compared to that of MT3SF, and yet there is not the evidence to confidently state which factors might predispose RM recruits to increased risk of sustaining this injury. Limitations to the methods used for obtaining data on risk factors is one possible explanation for this, and the absence of prospective cohort studies in military populations is a significant omission. This is identified as a strategy to help understand which of the numerous factors associated with risk of TSF are relevant to this cohort. Given the control of a number of extrinsic risk factors in RM training (fitness at entry, nutrition, training volume and intensity, footwear, rest), differences in lower limb flexibility, geometry and function during gait are proposed to be of importance.

The overall aim of this thesis is to investigate risk factors for TSF and MT3SF in RM recruits. Given the unusually high incidence of MT3SF in this population, the influence of standard issue footwear on biomechanical risk factors for this injury is considered. The development of a model of third metatarsal bending stress is designed to provide further information on the role of bone geometry, external loading and sagittal plane kinematics on stress fracture risk. Finally, a prospective analysis of the biomechanical, anthropometric and flexibility

characteristics of recruits is performed in order to identify which baseline characteristics predispose them to injury. Based upon the review of literature, the following investigations were developed:

- Study 1: The effects of standard issue Royal Marine recruit footwear on biomechanical variables associated with third metatarsal stress fracture

Study 1 investigates kinematics, plantar pressures and ground reaction forces when running in the gym trainer and combat boot issued to RM recruits upon commencement of training. Data were obtained for seven participants and used to calculate ankle moments and subsequently joint stiffness in the two footwear conditions. It was postulated that the design of the combat boot in particular would increase forefoot loads and restrict ankle dorsiflexion, which have been associated with metatarsal stress fracture risk. Horizontal ground reaction forces were assessed in order to further investigate forefoot loading in the light of previous suggestions for injury risk.

- Study 2: A model of stress acting on the third metatarsal during gait, derived from MRI, plantar pressure and kinematic data.

The purpose of the second study is to further develop the findings of Study 1, and investigate the role of metatarsal geometry and footwear in determining third metatarsal bending strain. MRI scans of the third metatarsal of five

participants are obtained and digitised to provide the cross-sectional geometry for slices taken at 5 mm increments along the length of the bone. The initial model is developed using data obtained from barefoot running. External vertical loads are estimated using plantar pressure analysis, and the deflection angle of the metatarsal is estimated by tracking markers placed on the dorsal surface of the foot. Mechanical bending calculations are used to estimate stress at each slice of the third metatarsal, with the model then applied to data obtained from Study 1 for running in standard issue footwear.

- Study 3: A prospective investigation into variables associated with tibial and third metatarsal stress fractures in Royal Marines recruits.

Study 3 is a large-scale prospective cohort study undertaken to identify whether aspects of anthropometry, flexibility or gait are associated with TSF and MT3SF in RM recruits. Data for relevant variables are obtained for 1065 recruits at the start of training over the course of 21 months. Recruits incurring relevant stress fractures, confirmed by bone scans, are compared to a group of controls representing those recruits who passed out of training without injury, enabling the identification of specific variables associated with each injury.

CHAPTER THREE: THE EFFECTS OF STANDARD ISSUE ROYAL MARINE RECRUIT FOOTWEAR ON BIOMECHANICAL VARIABLES ASSOCIATED WITH THIRD METATARSAL STRESS FRACTURES.

3.1. Introduction

The cyclic, intensive nature of military training programmes may lead to the onset of stress fracture through repeated periods of high duration exercise without sufficient rest. It has been previously reported that 3-6% of military recruits suffer lower limb stress fractures (Almeida et al., 1999; Beck et al., 1996, 2000; Ross & Allsopp, 2002), although incidences of up to 13% have been reported in naval recruits (Kaufman et al., 1999). Such injuries cause an average of 12-20 weeks of lost training time (Ross & Allsopp, 2002), reduced likelihood of training completion and increased clinical and financial burden on military institutions. In addition to these institutional burdens, the prospect of 'back-trooping', where recruits are injured and later re-join training with another troop, is a daunting one for many recruits.

The current Royal Marine (RM) training programme is considered to be one of the longest and most physically demanding in the world, lasting 32 weeks. The reported incidence of stress fractures in 2002 was 3.8% (Ross & Allsopp, 2002), which is in line with reported rates in other military settings. However, an unusually high proportion of these are to the third metatarsal (MT3), accounting for 38% of the lower limb stress fractures recorded (Ross & Allsopp, 2002). In other military training populations, MT3SF rates have not been widely reported.

The proportion of stress fractures to the metatarsals or foot as a whole have varied from 3-8% (Milgrom et al., 1985; Khan et al., 2008), to 39.3% and 64.4% of all reported lower limb stress fractures in two populations of US Marine Corps recruits (Shafer et al., 1999). However, in such examples, the broad classifications of 'foot' and 'metatarsal' stress fractures may mask the specific incidence of MT3SF. In sporting populations, metatarsal stress fracture incidence rates have been shown to account for 8% of stress fractures in track and field athletes (Bennell et al., 1996); 16% of all stress fractures in tennis (Maquirriain & Ghisi, 2006) and 8.8% of all stress fractures in a group largely composed of runners (Matheson et al., 1987). Previous literature reports a wide range of metatarsal stress fracture injury rates, however the proportion of MT3SF in RM recruits is particularly high and worthy of investigation.

The tibia typically incurs the highest proportion of lower limb stress fractures in other military populations, suggesting that there are aspects of the RM training programme that promote the high rate of MT3SF. One such aspect could be the duration of the training programme. Finestone et al. (2011) monitored injuries to military recruits through a progressively more difficult training programme, and found that the stress fracture site swung dramatically from 78% occurring in the tibia or femur in the first two stages of the programme (total 26 weeks), to 91% occurring at the metatarsals in the final section of training (a further 26 weeks). This evidence suggests that either the increased intensity or longer duration of training was linked with the development of metatarsal, rather than tibial or femoral stress fractures. Whilst the RM training programme is both long and physically demanding, it is difficult to quantify this in comparison to other military

training programmes. Furthermore, a training review examining loading of the relevant sites during these phases of training is required in order to establish a cause-effect relationship for the development of stress fractures.

Of the homogenous factors likely to influence the development of MT3SFs in RM recruits, the affect of footwear can be biomechanically analysed with relative ease and confidence. All recruits are issued with the same standard footwear to be worn during training: a Combat Assault Boot (CAB) and a neutral, lightweight cushioned gym trainer (GT) (Figure 3.1). The gym trainer is exclusively worn in distance training exercises in the first two weeks of training, as the boot is phased in gradually in weeks 3-8. After week 8 of training, the boot is worn for all training exercises (Rice et al., 2013).



Figure 3.1. The Combat Assault Boot (left) and gym trainer (right) issued as standard to Royal Marines recruits.

Since the turn of the 20th century, it has been the expectation that military training boots are designed to be robust and protective, with the current CAB being no exception. An article in *The Lancet* from 1898 states that:

“The ideal boot for a modern soldier should be impervious to moisture, stout and yet pliable, accurately shaped, and constructed of the most durable material. It should also be easily put on and taken off, should prevent the entrance of dust, sand, or mud, and, above all, favour the free play of each pedal articulation, while at the same time affording support to the muscles of the foot and ankle collectively.”

(The Lancet, 1898, p.1204)

In many respects, this philosophy has remained intact. The modern CAB is constructed of a durable leather upper, extending above the ankle, with rigid polyurethane sole. Whilst providing protection from abrasions and foreign objects/substances, the leather upper also provides lateral stability in order to protect from ankle inversion injuries when traversing uneven terrain. The sole is similarly designed with robustness and protection in mind, however these characteristics may have significant implications when considering the development of metatarsal stress fractures.

Despite a number of studies utilising plantar pressure in the analysis of metatarsal loading during locomotion (e.g. Arndt et al., 2002; De Cock et al., 2006; Nagel et al., 2008; Hinz, Henningsen, Matthes et al., 2008), there is a common failure to focus on the individual loading of the MT3. The few studies which have examined the loading modalities of the individual metatarsals support the need for further examples of such practice (Pollard et al., 1983;

Tappin & Robertson, 1991; Arangio et al., 1998; Hosein & Lord, 2000; Griffin & Richmond, 2005). These studies show that individual metatarsals experience loads which differ in magnitude and direction, and are vulnerable to differing loads. Research regarding the mechanisms of overuse injuries to individual metatarsals remains scarce however, with the MT3 being no exception.

It is proposed that factors which increase forefoot loading may increase MT3SF risk. For example limited ankle dorsiflexion (ADF) (Lin et al., 1996), high levels of midsole hardness (Wiegerinck et al., 2009) and fatigue (Nagel et al., 2008) have been demonstrated to increase forefoot pressures, which suggest increased vertical loading at the MT3. Whether vertical forces are implicated in MT3SFs is yet to be conclusively determined however. Lateral forces are proposed to be of significance when considering MT3SF mechanisms. It has been shown previously that the angle of application of resultant horizontal ground reaction forces (GRF) was significantly more laterally applied in RM recruits with a history of MT3SF compared with a control group (Dixon et al., 2006). As with other retrospective studies, it is not clear whether this is a compensatory mechanism, but it may be of importance considering that the MT3 is suggested to be most vulnerable to laterally applied forces (Arangio et al., 1998). Despite limited research into laterally applied force variables, Hosein & Lord (2000) demonstrated with the use of shear transducers during walking, that the MT3 region experienced the highest shear force, peaking between 40-60% of stance.

The CAB is issued to RM recruits approximately two weeks before the start of training, with the expectation that they will wear them in before the programme commences. It is a stiff, heavy boot, when compared to typical running footwear, and is worn for the majority of outdoor training exercises. The standard issue GT is commercially available and designed as a neutral running shoe, therefore it is logical to expect different plantar loading patterns when wearing the two conditions. Certain aspects of the boot which have been established to protect the wearer against more overt acute injury risks may be linked with risk factors for MT3SF. For example, the high leather upper, designed to reduce the likelihood of ankle inversion injuries, extends above the ankle. Research suggests that high walking boot shafts may restrict ADF (Böhm & Hösl, 2010), with greater shaft stiffness eliciting further reductions in ADF; while walking in military boots with varying shaft stiffness showed reduction in peak ADF with a stiffer shaft (Cikajlo & Matjačić, 2007).

The raised polyurethane heel relative to the forefoot in the CAB results in the plantar surface of the foot being held at an inclination of eight degrees to the surface in stance, compared to two degrees in the GT. While this may benefit in reducing Achilles tendon strain, as found with the use of EVA heel lifts in running (Dixon & Kerwin, 1999), previous research indicates that plantar pressures at the forefoot increase with running shoe heel height (Mandato & Nester, 1999). Heel inserts cause a similar effect in walking, increasing pressure and contact time at the metatarsal heads (Ramanathan, John, Arnold, Cochrane & Abboud, 2008). When combined with the proposed effect of the leather upper, MT3 external loading would be expected to be greater in the CAB

than the GT. Analysis of the excursion of the centre of pressure (COP) may help to further reveal the effects of the raised heel in the CAB. A more rapid translation of the y-coordinate (in the anterior direction) of the COP would be expected following heel strike in the CAB, compared to the GT. If greater pressure was detected beneath the metatarsal head in the CAB, the location of the COP would assist in verifying whether this were due to foot position or midsole hardness.

The metatarsophalangeal (MTP) joint can be considered to represent the end of the lever arm for moments acting about the ankle joint, therefore increased plantarflexion moment will result in either plantarflexion of the foot or increased forefoot load. Knowledge of the applied moment and the joint stiffness therefore may provide greater insight into the causes of any differences in MT loading in the two conditions, particularly if changes in external forces are not accompanied by kinematic changes. The calculation of joint moments provides indications of loading on internal structures, and has been demonstrated as a tool for indicating knee (e.g. Stefanyshyn et al., 1999; Lilley, Dixon & Stiles, 2011), subtalar joint (e.g. McClay, 2000; van Gheluwe, Kirby & Hagman, 2005) and ankle (e.g. Winter & Robertson, 1978; Braunstein, Arampatzis, Eysel & Bruggemann, 2010) internal loads previously. Joint stiffness, which indicates the quantity of joint motion in response to an applied moment, has been shown to vary inversely in response to external cues such as surface hardness (Ferris et al., 1999), and is thought to have injury implications when present in excess, through reduction of force attenuation or increased muscular demand (Butler, Crowell & Davis, 2003). In the current study, it was anticipated that restricted

ankle rotation caused by the high ankle CAB would lead to a greater ankle joint stiffness for this condition compared with the GT.

The purpose of this study was to investigate the effect of wearing either the standard issue CAB or GT on factors proposed to be associated with MT3SF risk. Ankle joint kinematics and kinetics, plantar pressure at the MT3, and horizontal ground reaction force characteristics were investigated when running in each condition. It was hypothesised that compared to the GT, the CAB would yield: (a) lower peak ankle dorsiflexion; (b) greater peak pressure, impulse and loading rate at the MT3 head; (c) greater and more laterally applied horizontal GRF; (d) earlier heel-off; (e) greater ankle joint stiffness during the stance phase of running and (f) earlier and more rapid movement of the centre of pressure towards the MTP joint.

3.2. Methods

Seven (age 18.3 ± 0.4 years; mass 81.1 ± 8.2 kg) injury-free, physically active, naturally heel-striking male volunteers with size 11 (UK) feet were recruited from a cohort of undergraduate sports science students at the University of Exeter. Only males were selected, in order to mirror the typical RM recruit, and all volunteers were familiar with wearing and running in combat boots. Eligible foot size was restricted by the size of the available pressure insole, which was specifically chosen for its long neck, enabling data collection in an above-ankle shoe. Eligible participants were provided with information on the study and

provided written informed consent (Appendix D(i)). The study was given ethical approval by the Sport and Health Sciences Ethics Committee, University of Exeter.

Volunteers were assessed wearing the CAB and GT standard issue RM footwear. The CAB (Figure 3.1 left) was constructed of a stiff moulded polyurethane sole, heel block and stiff leather upper that extends beyond the ankle, which was laced up to one eyelet below the top in all trials. The GT (Figure 3.1 right) was constructed of a lightweight EVA neutral sole with suede/mesh upper extending to just below the line of the lateral malleolus. The GT was deemed to be representative of a typical commercially available running shoe, thus providing an indication of how the CAB compares to sample running footwear.

Mechanical test data were collected for each footwear condition (Table 3.1). The outsole stiffness of each condition was assessed. To test this, a solid plate was placed in the forefoot of the boot and clamped to allow bending at the approximate location of the metatarsophalangeal joint. An actuator, driven vertically at a velocity of approximately $0.3 \text{ m}\cdot\text{s}^{-1}$, applied a vertical force to the heel of the boot that was recorded by a load cell. The peak force required to bend the outsole 45 degrees was recorded for each of ten trials for each condition. This bespoke method of examining outsole stiffness was adapted from Oleson, Adler & Goldsmith (2005). The impact attenuation properties of the footwear were assessed by an impact-testing device (ASTM, 2001. Test method: F1976-99, Standard Test Method for Cushioning Properties of Athletic Shoes Using an Impact Test. ASTM International, West Conchohocken PA,

USA). A 45 mm diameter, 8.5 kg weight was dropped with an impact velocity of $92 \text{ cm}\cdot\text{s}^{-1}$ (as in Stiles & Dixon, 2007). Five pre-impacts preceded twenty test impacts per condition, with the average 'peak g' value presented as a measure of the cushioning properties of the material. 'Peak g' represents the peak deceleration of the missile due to the resistance of the material being struck. A greater value indicates greater resistance and thus less cushioning. The missile struck the heel section of the boot/shoe. Midsole hardness was assessed using a durometer (Durotech, model B202, Hampden Test Equipment Ltd., England) without the pressure insole present.

In-shoe plantar pressure data and synchronous kinematic and force data were collected. A pair of size 10 (UK) pressure insoles with a 30 cm neck (RSScan, Belgium), sampling at 500 Hz, was provided to accommodate the CAB's leather upper without damaging the insole or causing discomfort to the participant. The pressure insole was inserted into the footwear, before the participant put their foot in, ensuring a comfortable fit with no bending of the insole. Participants then laced their footwear to a self-selected tightness. All participants wore their own socks. The pressure insoles were connected to a wireless data logger contained in a belt worn around the waist. The cables connecting the logger to the insoles were secured using Velcro strapping in order to minimise any noise being recorded through the movement of cables. The data logger allowed approximately seven seconds of data collection once triggered. Data logging was triggered manually by the tester at the start of the run up, allowing the footstep that struck the force plate to be identified and recorded by counting the steps from the start of data logging to force plate strike. Pressure data were analysed for this footstep.

Two-dimensional kinematic data were collected using an eight camera system (Vicon Peak, 120 Hz, automatic, optoelectronic system; Peak Performance Technologies, Inc., Englewood, CO). Dynamic calibration of the kinematic capture system was performed prior to testing each day. Force plate data were collected at 960 Hz (AMTI, Watertown, MA, USA) for one right foot step from each trial. Participants performed warm-up and familiarisation trials in the laboratory in each footwear condition, practicing striking the force plate without adjusting their natural running stride. Ten successful running trials were collected per condition. Each trial required the runner to heel-strike the force plate (situated flush with the laboratory floor) with their right foot while running at a constant velocity of $3.6 \text{ m}\cdot\text{s}^{-1}$ ($\pm 5\%$). This velocity was chosen as it is representative both of training speeds employed during training, and of the average speed required to pass the treadmill run entry test (www.royalnavy.mod.uk/Careers/How-to-join/Eligibility). Running velocity was monitored using hip-height photocells placed 1 m either side of the centre of the force plate. Participants were visually observed to ensure that they struck the force plate naturally.

Mask analysis within the Footscan Insole software (version 2.39, RSScan, Belgium) was used to identify the five metatarsal head regions: M1, M2, M3, M4, M5 (see Figure 3.2). Mask placement was based on previously reported locations (Willems, De Clercq, Delbaere et al., 2006), and a separate analysis was conducted to assess marker placement reliability over three separate occasions. Intra-class correlation coefficients in excess of 0.997 were reported for peak pressure values obtained for the eight anatomical regions shown in

Figure 3.2, demonstrating excellent reliability. Peak pressure, peak loading rate of pressure and impulse were exported from the software following mask placement. Absolute and relative values at each mask location were calculated to assess any change in the distribution of pressure variables between conditions. Relative peak pressure, impulse and peak loading rate were calculated for each metatarsal region as the percentage of the sum of values across the five metatarsal head regions.

The heel mask regions were assessed to provide the time of heel-off. Instantaneous loading rate was calculated at the heel mask regions (H1, H2) using the first central difference method, and the first point at which this increased above $5 \text{ kPa}\cdot\text{ms}^{-1}$ was used to represent the start of ground contact. This instant was matched to the start of ground contact in the force plate data, which was defined as the instant when $F_z \geq 10 \text{ N}$. This method was chosen in favour of determining a minimum pressure value, as contact between the pressure insole and foot prior to ground contact may have influenced the identification of heel strike. Heel-off was identified using the pressure insoles as the time following ground contact when both heel masks reported a pressure value of 0 kPa . The time of heel off was calculated as a percentage of total stance using the ground contact time provided by force plate data, with toe-off defined as the point following heel strike when $F_z \leq 10 \text{ N}$. Time to peak pressure was calculated for each metatarsal region and converted to a percentage of total stance time in the same way. Low & Dixon (2010) performed a reliability analysis of the RSScan pressure insole system used in the present study, using similar data collection methods. ICC values of $r > 0.75$ were reported over 8 running trials, representing excellent reliability. Magnitudes of

vertical load reported by the insoles were significantly lower than those simultaneously reported by the AMTI force plate however, therefore a correction factor was applied in the present study, using the force plate as recommended by Low & Dixon (2010). The ratio between peak impact force at the heel as calculated using the pressure insole and force plate was calculated and used to scale absolute pressure values.

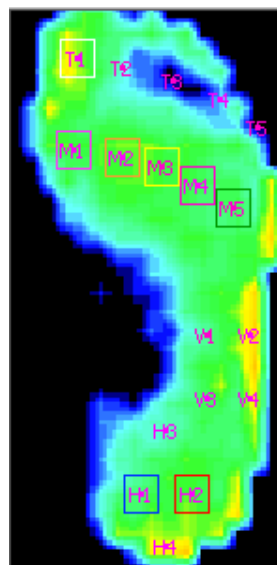


Figure 3.2. Sample mask locations used to identify the MT head regions. Plantar pressure data was obtained for running using a pressure insole worn in the gym trainer. Data is from trial participant 1, trial 1.

Reflective anatomical markers were attached to the right leg in the following locations to facilitate the calculation of 2-dimensional kinematics: two markers defining the line of the Achilles tendon, one placed on the midline of the posterior aspect of the shank, immediately below the calf muscle belly, and one placed 10 cm directly inferior to this; one placed on the superior aspect of the midline of the posterior calcaneus; one marker placed on the dorsal aspect of

the articulation of the third metatarsal and medial cuneiform. Markers were also placed at the 3rd and 5th metatarsophalangeal joints. Two-dimensional ankle kinematic data were obtained in the sagittal plane using the two Achilles tendon markers to define the shank segment, and the superior calcaneous and dorsal foot marker to define the foot segment. The angle was defined so that plantarflexion caused an increase in angle, and a neutral ankle was at zero degrees (Figure 3.3). Two-dimensional analysis of ankle kinematics has been shown to have good agreement with three-dimensional analysis in the past (Areblad, Nigg, Ekstrand, Olsson & Ekstrom, 1990), and therefore was deemed appropriate in the present study. Kinematic data were synchronised with foot strike within the Peak software, using a force plate event of >10 N. Initial ADF, peak ADF and time of peak ADF were calculated for each trial and referenced to a relaxed neutral standing position. The Peak system used in the present study has been found to have excellent accuracy during static and dynamic movements (Lilley, 2012). During dynamic assessment, a baton with markers placed 0.913 m apart was moved through each primary axis for 5 seconds on five occasions. Marker positions on the baton were reported to within an average of 3 mm and maximum of 9 mm in any axis during dynamic testing.

For the present setup, if markers defining the foot were reported with a maximum vertical error of 9 mm in opposite directions, giving an 18 mm deviation, this would lead to an error of 6.8 degrees, assuming a horizontal displacement of 150 mm. If the shank markers deviated by 18 mm in the sagittal plane, an error of 10.2 degrees would be produced if markers were 100 mm apart. For the mean deviation of 3 mm however, these errors would be reduced to 2.3 and 3.4 degrees respectively. The worst case scenario would be for the ankle angle to err by 17 degrees, however typically less than 6 degrees

variation would be expected, even if marker positions were reported erroneously in opposite directions, thus maximising the error in reporting segment position. By collecting multiple trials, the influence of random error was reduced. Although reliability data for joint kinematics were not obtained with the present setup, Ferber, McClay Davis, Williams & Laughton (2002) performed an analysis of within- and between-day reliability of lower limb joint angles calculated using a Vicon Peak system. These authors reported excellent within-day ($ICC > .88$) and between-day ($ICC > .83$) repeatability for ankle variables. Other variables were less well reproduced, with between-day peak hip rotation performing worst ($ICC = .54$). Ferber et al. (2002) used the mean of five trials and did not reference to static trials, therefore it can be assumed that the methods used in the present study ensured performance of at least a similar level.

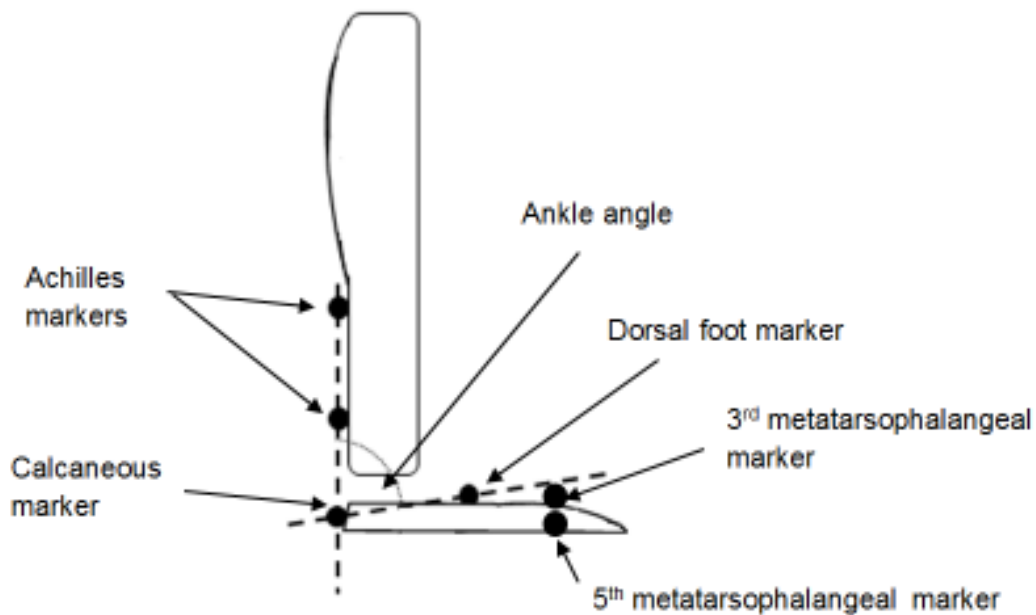


Figure 3.3. A diagram showing marker placements on the lower limb enabling the calculation of 2D ankle angle during stance. The shank and foot segments are depicted from a sagittal perspective. The intersection of the lines projected by the markers on the shank and foot represents the vertex of the ankle angle.

Moments about the ankle joint centre were calculated at 120 Hz in the sagittal plane using inverse dynamics calculations. Inertial characteristics for the foot were obtained from values presented by Dempster (1955). The magnitude and timing of maximum plantarflexion moment were calculated and normalised to bodyweight and stance duration respectively. Stiffness was then calculated using the moment and angle data (for sample time histories see Figure 3.4). Stiffness was defined as the change in ankle angle as a result of the applied ankle joint moment ($\Delta\text{Ankle moment} \div \Delta\text{ADF}$), and was calculated at two distinct phases of stance: the dorsiflexion phase (DFP, Figure 3.4, region A) and the plantarflexion phase (PFP, Figure 3.4, region B). The DFP was defined as the time between the occurrence of peak initial plantarflexion and the time of

peak dorsiflexion. The PFP was defined as the time between the occurrence of peak dorsiflexion and toe-off. Mean stiffness for each phase was calculated as the slope of the curve from the point of maximum plantarflexion moment to the point of minimum plantarflexion moment, and was normalised to the mass of the subject ($\text{N}\cdot\text{m}^{\circ}/\text{kg}$). Moment and stiffness calculations were performed using a custom Matlab code (v. 7.4, The Mathworks, USA).

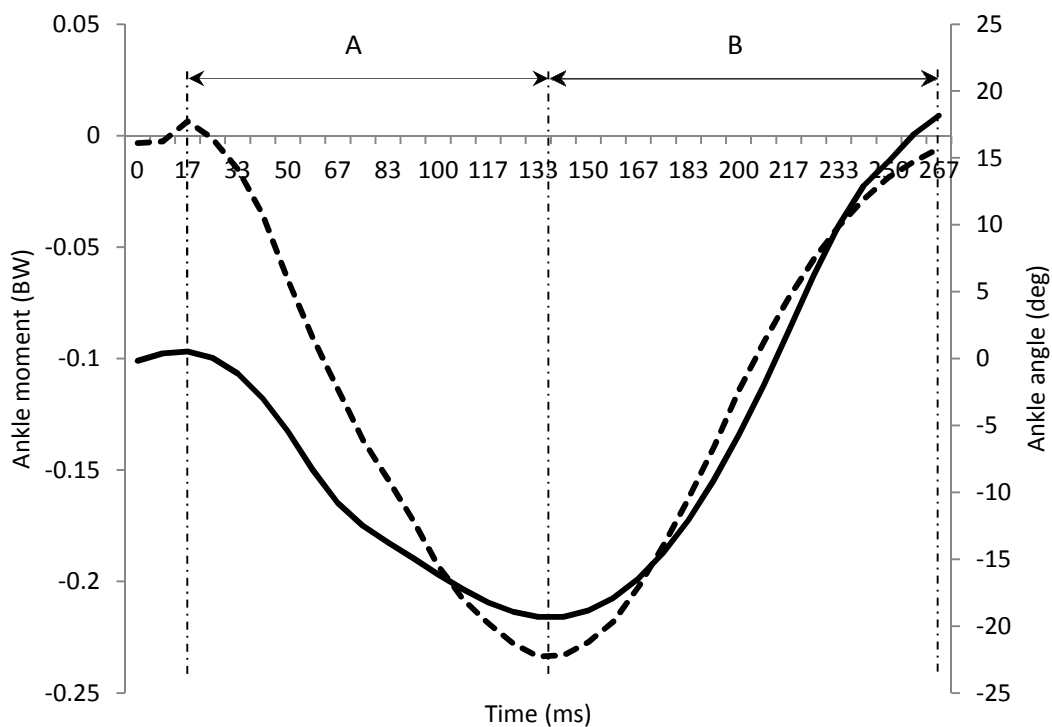


Figure 3.4. Sample ankle moment (dashed line) and ankle angle (solid line) time histories with A) dorsiflexion and B) plantarflexion phases defined for stiffness calculation during the stance phase of running. A negative ankle moment indicates a plantarflexion moment, a negative ankle angle represents dorsiflexion. Data taken from participant 3, trial 4.

For each trial, vertical and horizontal force characteristics were assessed. Resultant horizontal force (F_H) was calculated using the following formula where F_x is medio-lateral force and F_y is antero-posterior force:

$$F_H = \sqrt{(F_x^2) + (F_y^2)} \quad \text{Eq. 3.1.}$$

A sample F_H time-history is outlined in Figure 3.5, with the key phases of forefoot loading identified. Point A represents the peak F_H during the braking phase of stance, and point B represents the peak F_H during the propulsive phase. The first peak on the graph occurs within the first 50 ms of ground contact. This peak was ignored as it was deemed of less significance to metatarsal loading than the peaks occurring around the times of peak braking and propulsion. Force magnitudes were normalised to bodyweight (BW).

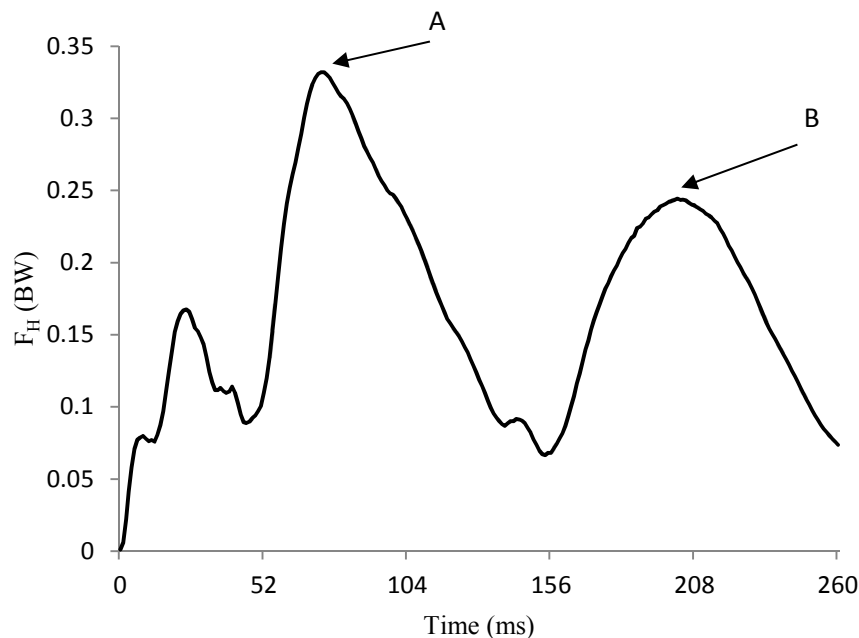


Figure 3.5. A typical time history of resultant horizontal force for a participant running at 3.6 m.s^{-1} whilst wearing a combat assault boot. Point A represents peak braking force, point B represents peak propulsive force. Data taken from participant 3, trial 4.

The angle of the application of F_H was calculated relative to the laboratory sagittal plane using the following formula:

$$F\theta_H = \tan^{-1}(F_x \div F_y) \quad \text{Eq. 3.2.}$$

A negative angle indicated a medially applied force, with a positive number indicating a lateral angle of force application. $F\theta_H$ was calculated at the instant of peak braking and peak propulsion by identifying these points on the F_H time history.

The y-coordinate of COP was selected for analysis from each trial in order to analyse COP displacement in the anterior-posterior direction (COP_Y). Markers placed at the base of the calcaneus and on the MTP joint were used for reference points for the COP_Y location. The calcaneus marker was used to represent the back of the foot, and the mean y-coordinate of the markers placed at the third (MTP3) and fifth (MTP5) metatarsophalangeal joint locations was used to represent the front of the foot. Mean COP_Y position as a percentage of foot length was calculated ($PCOP_Y$), where 0% was level with the calcaneus and 100% was level with the MTP joint. COP_Y velocity ($VCOP_Y$) was calculated using the first central difference method. The time when the COP_Y location first became level with the MTP joint was also calculated as a percentage of total ground contact time ($TCOP_Y$). Figure 3.6 represents a typical time history of the relative locations of COP_Y and the MTP joint. For comparison between force data captured at 960 Hz and kinematic data collected at 120 Hz, kinematic data were extrapolated to 960 Hz. $TCOP_Y$ was selected to the nearest 1/960s based

on visual inspection of the data. Horizontal GRF variables have been reported to be obtained with a high degree of reliability for both within- (ICC > .88) and between- (ICC > .91) day testing when assessed for running at 3.6 m.s⁻¹ (Ferber et al., 2002). The error in COP data has been reported to be lower towards the centre of the plate than the edges of the plate. Bobbert & Schamhardt (1990) reported errors of up to 20 mm at the edge of a Kistler force plate, however Gill & O'Connor (1996) found an AMTI force plate to perform favourably, with maximum errors of less than 10 mm.

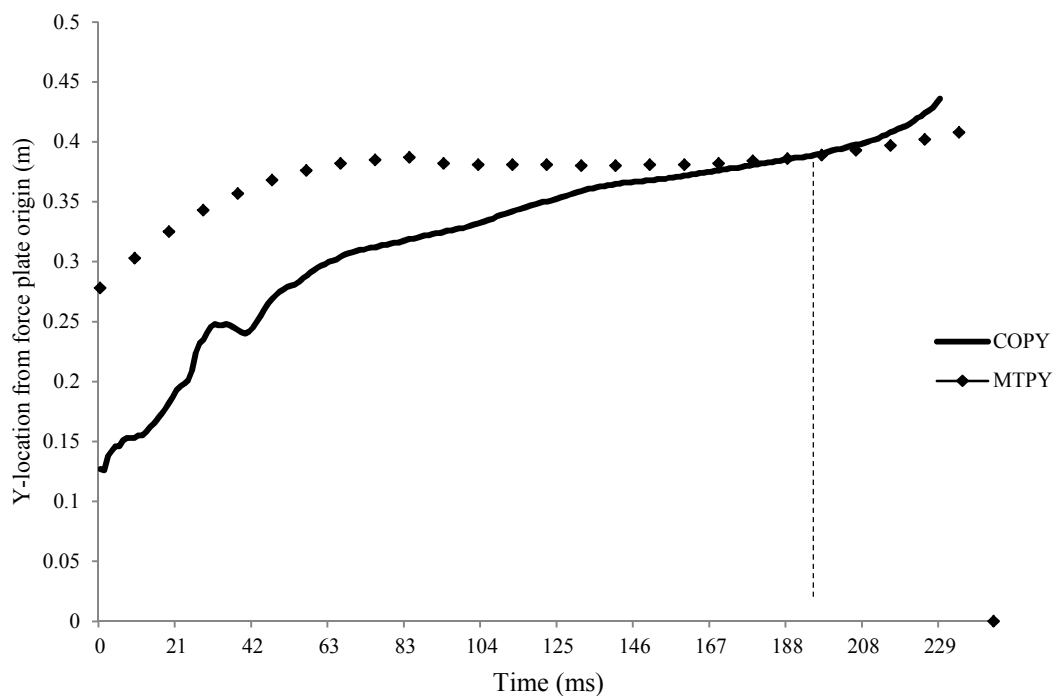


Figure 3.6. A typical time history of the relative locations of the y-coordinate of the COP and MTP joint. The COP can be seen to cross the MTP joint at approximately 190 ms (dashed vertical line).

Kinematic data were interpolated within the Peak software. Kinematic and kinetic data were processed using a quintic spline smoothing technique (Woltring, Huiskes, De Lange & Veldpaus, 1985). Values were obtained from each trial, and the mean of ten trials calculated for each participant. This

provided seven values for each variable, which were entered into SPSS (version 15). All variables were tested for normality using the Shapiro-Wilk statistic ($P < 0.05$) and found to be non-normally distributed. Means were therefore compared using Wilcoxon tests (1-tailed, $P < 0.05$), as a non-parametric alternative to paired samples t-tests. Effect sizes (Cohen's d – Cohen, 1988) were produced for all variables in order to further examine relationships between independent variables. Effect size provides information on the difference between means in light of the sample size and distribution of data, rather than relying solely on the probability-based p-value. Based on Cohen's (1988) guidelines, a medium effect size was considered to be .50, meaning that the difference between means for the two conditions is half of the standard deviation.

3.3. Results

Table 3.1 displays the results of mechanical testing of the footwear. The CAB was three times the mass of the GT, and values for peak 'g' indicate that impacts are less well attenuated by its midsole. Similar force was required to bend the outsole 45 degrees for each shoe. Table 3.2 summarises the results of kinematic and force analysis. Peak dorsiflexion angle was significantly lower and occurred significantly earlier when wearing the CAB. Heel-off occurred later in the CAB and the angle of application of the force vector at the instant of resultant horizontal peak braking force was also more laterally applied in the CAB. Peak ankle plantarflexion moment was greater and occurred earlier in the CAB, and ankle stiffness during the dorsiflexion stage was greater in the CAB.

The only significant difference for COP variables was for $VCOP_Y$, which was significantly greater in the CAB than GT. Figure 3.7 shows mean $VCOP_Y$ curves for one participant for each footwear condition. The CAB time history typically demonstrated a greater peak in the first 50 ms of stance, compared to the smoother curve apparent in the GT time history.

Table 3.3 summarises the results of plantar pressure analysis. For all pressure magnitude variables investigated, values were significantly greater when wearing the CAB ($P < 0.05$), except for peak pressure at the MT1 region, and impulse at the MT1 and MT5 regions. The time of peak pressure was only significantly different for the MT3 region, where it occurred earlier in the CAB. Figure 3.8 shows the relative distribution of peak pressures, impulse and peak loading rates at the five metatarsal head regions in the two footwear conditions. There were no significant differences in the relative pressure variables at the metatarsal heads, with the exception of peak loading rate being significantly greater in the GT than CAB at the MT1 and MT4 regions ($P < 0.05$). Notably, Table 3.3 and Figure 3.8 identify that, of the metatarsal regions, the MT3 region experienced the highest magnitudes of pressure, impulse and loading rate in the CAB. In the GT, it experienced the highest peak pressure, but second highest impulse, and the third highest peak loading rate.

Table 3.1. Mechanical test data for RM recruit standard issue footwear. The combat assault boot (CAB) and gym trainer (GT) were weighed and assessed using a drop test to determine peak deceleration (Peak 'g'), with a higher value representing more rapid deceleration; a durometer to determine midsole hardness on the Shore A scale, with a higher value being harder; and the force required to bend the outsole 45 degrees, with a higher value representing a stiffer outsole.

| Mechanical Test | CAB | GT |
|--|-----------------|-----------------|
| Mass (per shoe) [kg] | 1.2 | 0.4 |
| Peak 'g' | 17.8 (0.21) | 15.2 (0.10) |
| Midsole hardness [Shore A scale] | 75 | 45 |
| Peak force required to bend outsole 45° [N] | 108.2 (5.13) | 105.7 (9.03) |

Table 3.2. For seven participants, the mean (SD) of ten trials assessed during ground contact when running with the combat assault boot (CAB) and gym trainer (GT) are presented. Values for horizontal force, ankle angle, ankle moment and ankle stiffness variables are included. Results of Wilcoxon tests are presented ($P < 0.05$) and effect sizes presented using Cohen's d . Moments and forces are normalised to bodyweight (BW) and timings are presented as % stance. Negative values for $F\theta_H$ represent medially applied force. Statistically significant differences between conditions are identified using an *.

| Variable | CAB | GT | p | d |
|--|------------------|------------------|--------|------|
| F_H braking [BW] | 0.38 (0.09) | 0.38 (0.07) | .368 | .00 |
| F_H propulsive [BW] | 0.29 (0.09) | 0.29 (0.11) | .465 | .00 |
| $F\theta_H$ braking [deg] | 14.90 (2.90) | 10.41 (3.63) | .033* | 1.37 |
| $F\theta_H$ propulsive [deg] | 3.76 (1.12) | 5.49 (3.11) | .112 | .74 |
| Initial ADF [deg] | 4.11 (4.77) | 5.53 (4.14) | .171 | .32 |
| Peak ADF [deg] | 15.74 (2.25) | 18.13 (1.95) | .009* | 1.13 |
| Time of peak ADF [%] | 51.63 (3.71) | 55.53 (2.59) | .014* | 1.22 |
| Time of Heel Off [%] | 64.63 (10.96) | 58.10 (10.88) | .009* | .60 |
| Peak APF moment [BW] | -0.27 (0.02) | -0.19 (0.04) | .009* | 2.53 |
| Time of peak APF moment [%] | 54.92 (4.25) | 57.23 (3.65) | .014* | .58 |
| Ankle stiffness (DFP) [N.m/deg/kg] | 13.15 (2.86) | 9.64 (2.41) | .022* | 1.33 |
| Ankle stiffness (PFP) [N.m/deg/kg] | 11.46 (2.64) | 9.88 (1.98) | .300 | .75 |
| PCOP _Y [%] | 90.85 (4.10) | 92.22 (1.91) | .370 | .43 |
| VCOP _Y [m.s ⁻¹] | 8.55 (2.14) | 4.99 (1.43) | <.001* | 1.96 |
| TCOP _Y [%] | 67.27 (10.19) | 60.89 (6.95) | .080 | .73 |

F_H = horizontal force vector; $F\theta_H$ = angle of horizontal force vector; ADF = ankle dorsiflexion; APF = ankle plantarflexion; DFP = dorsiflexion phase of stance; PFP = plantarflexion phase of stance; PCOP_Y = mean position of the y-coordinate of the centre of pressure, as a percentage of foot length; VCOP_Y = peak velocity of the y-coordinate of centre of pressure; TCOP_Y = time point at which the y-coordinate of centre of pressure crosses the metatarsophalangeal joint.

Table 3.3. For seven participants, the mean (SD) of ten trials assessed during ground contact when running with the combat assault boot (CAB) and gym trainer (GT) are presented. Values for peak pressure, impulse, peak loading rate and the timing of peak pressure included for each metatarsal head region (MT1-5). Results of Wilcoxon tests are presented ($P < 0.05$) with effect sizes presented directly below using Cohen's d . Statistically significant differences between conditions are identified using an *. Peak pressures are presented in kilopascals (kPa) and timings are presented as % stance.

| Region | Peak Pressure [kPa] | | | Impulse [Ns] | | | Peak Loading Rate [kPa.ms ⁻¹] | | | Timing of Peak Pressure [% stance] | | |
|--------|---------------------|-------------------|---------------|------------------|------------------|---------------|---|----------------|---------------|------------------------------------|-----------------|--------------|
| | CAB | GT | P d | CAB | GT | P d | CAB | GT | P d | CAB | GT | P d |
| MT1 | 403.11 (106.43) | 356.66 (79.85) | .059 .49 | 56.19 (13.27) | 47.01 (7.70) | .038* .86 | 7.62 (1.72) | 5.55 (1.72) | .009* 1.20 | 63.27 (4.01) | 63.34 (4.44) | .433 .02 |
| MT2 | 412.46 (59.32) | 341.53 (33.26) | .009* 1.47 | 54.39 (12.05) | 42.04 (11.54) | .009* 1.05 | 8.22 (2.05) | 5.20 (1.77) | .009* 1.58 | 61.18 (4.41) | 63.13 (4.15) | .119 .46 |
| MT3 | 456.34 (25.16) | 369.94 (25.53) | .009* 3.41 | 61.84 (7.36) | 44.57 (6.37) | .009* 2.51 | 9.61 (1.33) | 5.01 (1.56) | .009* 3.17 | 58.61 (5.55) | 61.46 (5.59) | .009* .51 |
| MT4 | 400.84 (76.00) | 315.17 (55.77) | .009* 1.29 | 58.03 (8.69) | 42.47 (7.29) | .009* 1.94 | 8.65 (3.79) | 4.08 (1.60) | .009* 1.57 | 53.76 (3.28) | 55.69 (4.95) | .045* .46 |
| MT5 | 320.87 (45.95) | 253.46 (40.16) | .023* 1.56 | 43.59 (11.88) | 35.36 (5.45) | .069 .89 | 9.31 (6.21) | 4.43 (1.95) | .014* 1.06 | 69.55 (6.24) | 69.01 (5.91) | .176 .89 |

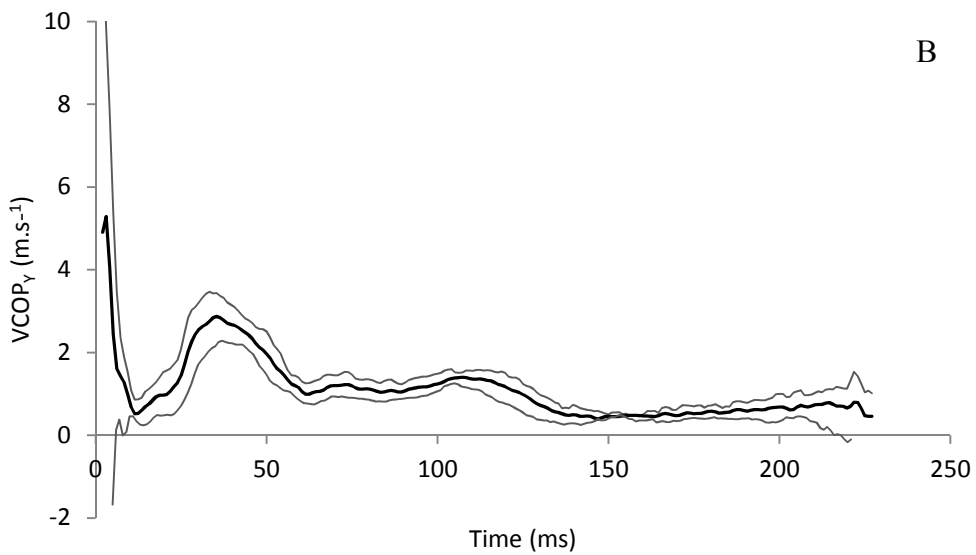
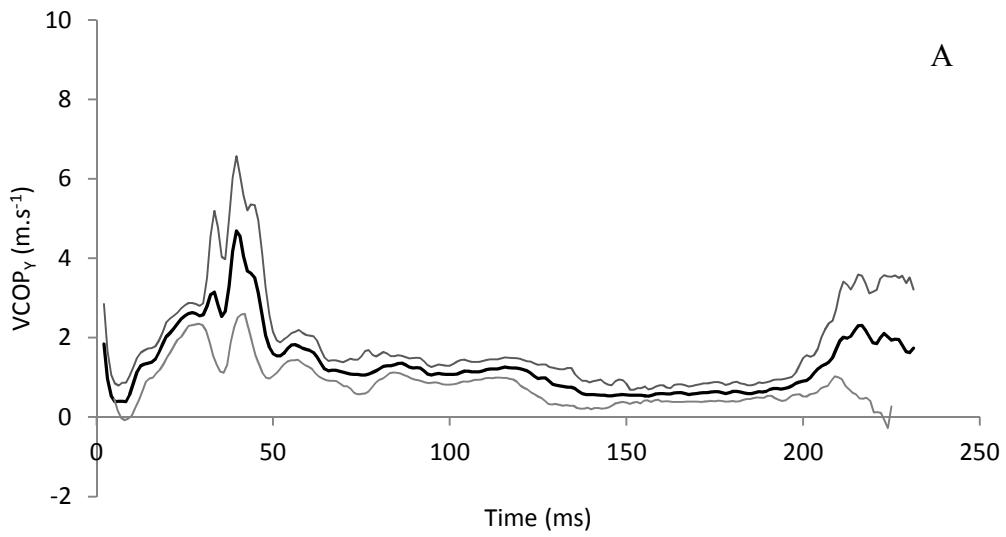


Figure 3.7. Time histories for participant 5 showing the mean anterior-posterior velocity of the centre of pressure during ground contact from ten running trials whilst wearing the combat assault boot (A) and gym trainer (B). Standard deviations are represented by the grey lines.

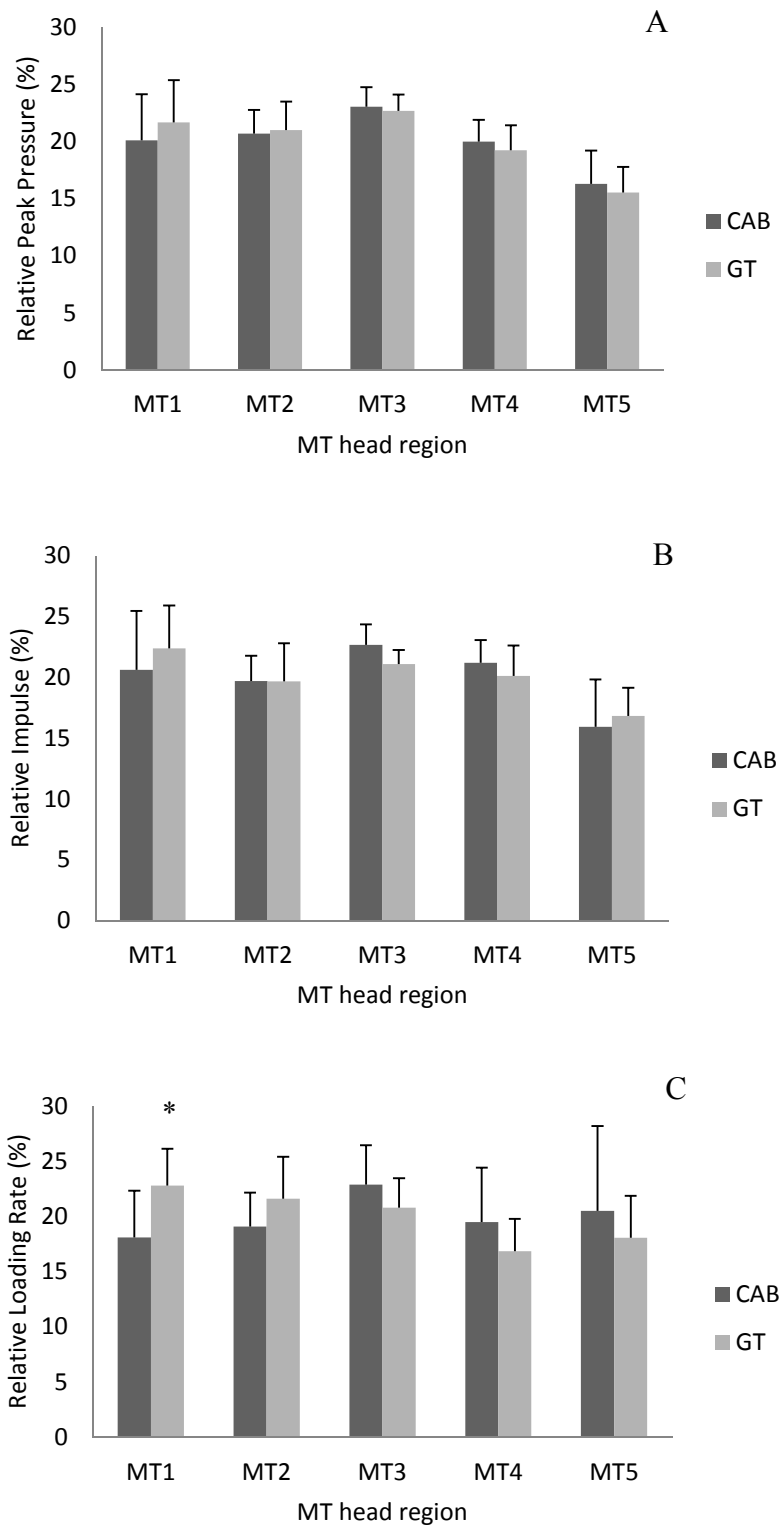


Figure 3.8. (A) Mean relative peak pressure, (B) impulse and (C) peak rate of loading at each of the metatarsal head regions for running in the combat assault boot (CAB) and gym trainer (GT), with error bars representing standard deviations. The significant difference between conditions is marked by an * ($P < 0.05$).

3.4. Discussion

The present study was conducted to investigate whether running in the standard issue RM CAB and GT was associated with changes in variables associated with increased risk of MT3SF. The assessment of seven healthy males revealed differences in ankle kinematics and kinetics, horizontal ground reaction forces and in-shoe plantar pressure data between conditions which suggest overall that wearing the CAB may increase the risk of sustaining a MT3SF. The outcomes relating to each individual hypothesis are now discussed.

The finding that peak ADF was reduced and occurred earlier in the CAB supports hypothesis (a). This observation is in agreement with previous research showing that increased boot shaft stiffness reduced ADF range of motion in hiking boots (Cikajlo & Matjacic, 2007; Böhm & Hösl, 2010). The peak ADF values presented here are comparable to a previous study using the same footwear (Dixon et al., 2006), although differences in boot characteristics such as shaft stiffness may account for the slightly lower values presented here than some examples (e.g. Stackhouse, Davis & Hamill, 2004; Hardin, van den Bogert & Hamill, 2004). It should be noted that the stiffness of the shaft of the CAB is likely directly related to how high it is laced, as the leather material itself is of low stiffness. Although lacing was consistent in all trials (one eyelet below the top), future research should investigate whether lacing up to a lower eyelet can improve ADF range of motion without compromising frontal plane stability. To date, this has not been investigated in above-ankle boots.

Previous research links highly restricted passive ADF range of motion ($<10^{\circ}$) with metatarsal injury risk (Hughes, 1985), and increased forefoot ulceration in diabetic patients (Lin et al., 1996), although prospective studies of MT3SF risk factors are absent from the literature. Hughes (1985) suggested that without 10° of ADF, compensatory pronation and flexion of the first ray during stance must occur. As the first ray rotates, this causes increased load to be accepted by the central MT heads. Although the results of the present study show that more than 10° of ADF was available (dynamically) in both conditions, the lower range of motion in the CAB was expected to cause an earlier heel-off and a subsequent increase in pressure at the MT3 head.

Despite a greater peak pressure at the MT3 head region in the CAB, supporting hypothesis (b), the expected earlier occurrence of heel-off for this condition (hypothesis d) was not supported by the results of this study. Contrary to expectations, heel-off was found to occur later for the CAB than for the GT condition. In addition, peak pressure at the MT3 head region occurred before heel-off in the CAB, whereas in the GT, peak pressure occurred after heel-off. Therefore the suggestion that reduced ADF in the CAB would lead to early heel-off and subsequently increased MT3 head pressure, is rejected. An alternative suggestion is therefore put forward in light of the ankle stiffness findings.

Whilst there was no difference in the joint stiffness during the plantarflexion phase of stance, the greater resultant joint moment in the CAB during the

dorsiflexion phase was accompanied by a lower ankle angle displacement, resulting in a stiffer ankle joint in this condition, supporting hypothesis (e). With the foot in a fixed position prior to heel-off, the rotation of the lower leg and consequent ankle flexion occurring during this phase, is controlled by a combination of the boot upper stiffness and eccentric muscular control of the triceps-surae muscle group. It is suggested that the greater plantar-flexor moment during this dorsiflexion (eccentric) phase, with the heel maintained in contact with the ground, contributes to the greater forefoot pressures observed for this CAB condition. This offers a potential explanation for the greater pressure at the MT heads in the CAB, and further implications for MT stress fracture susceptibility. As well as the greater plantarflexion moment without foot movement resulting in greater forefoot pressures, the increased eccentric muscular activity, as implied by these results, can be expected to lead to earlier fatigue of the calf muscles. Studies such as Arndt et al. (2002) and Weist et al. (2004) demonstrate that fatigue of the plantar-flexor muscles increases the load on the metatarsals, which may be significant in the high rate of MT stress fractures in RM recruits wearing the CAB. Direct estimation of muscle activity and fatigue in future studies using similar footwear conditions would allow this suggestion to be investigated.

Previous work has identified that laterally applied forces are potentially the most damaging to the MT3 (Arangio et al., 1998), highlighting the importance of considering horizontal forces acting on the foot. Resultant horizontal force magnitude was not significantly different between conditions, however a significantly more laterally applied force vector at the instant of peak braking

force in the CAB was observed, partially rejecting hypothesis (c). Dixon et al. (2006) found that RM recruits with a history of MT3SF demonstrated a more laterally applied force when running in military boots, compared to their matched controls. It was suggested by those authors that individuals who had experienced MT3SF may demonstrate altered forefoot function during braking compared to controls. In the present study, the CAB caused horizontal braking forces to be applied more laterally than in the GT. Given the suggestions regarding increased risk of MT stress fracture in general when wearing the CAB, this may be significant in explaining the particularly high rate of MT3SFs in the population training in this boot. However, prospective studies of the individual characteristics that predispose certain RM recruits to MT3SF are required to provide stronger evidence regarding the possible role of lateral force application in the development of this injury.

Plantar pressure analysis allowed the examination of localised normal forces beneath the MT heads and indicated in the present study that pressure, impulse and loading rate at the interaction of shoe and foot at the MT3 region were significantly greater in the CAB, supporting hypothesis (b). Peak pressures obtained for the GT were comparable with other studies of running at similar velocities in cushioned shoes (e.g. Weigerinck et al., 2009), whilst the values obtained for the CAB were similar to previously reported values in the same model of CAB (House, Waterworth, Allsopp & Dixon, 2002). The relative load experienced by the MT3 compared to the other metatarsal heads did not change between footwear conditions, with the MT3 head region experiencing the highest peak pressure, impulse and loading rate of all the MT head regions

in the CAB. In the GT, it experienced the highest peak pressure, but the second highest impulse and the third highest loading rate.

Hinze et al. (2008) also reported the highest impulse at the MT3 head region when walking in military boots, although these authors found that the MT2 head region experienced the highest peak pressure. This may be due to differences in the boots tested (German army boot in Hinze et al., 2008), or differences in pressure distribution between walking and running. Arndt et al. (2003) demonstrated that impulse in the MT2-MT5 region during a fatiguing walk was lower in a more flexible-soled Swedish military boot compared to a stiffer one, although this was not evaluated statistically, while Chackpaiwong et al. (2008) demonstrated in barefoot locomotion that the order of magnitudes of pressure and impulse changed between running and walking. The greater MT3 head region loading in the CAB suggests greater vertical loads are transmitted to this bone for this boot condition compared with the GT. With the proximal fixation of the MT3, these loads may lead to bending strain (simultaneous compression and tension) acting on the metatarsal shaft, but this suggestion requires further examination.

In addition to changes in net muscle moment and foot position, the harder midsole material likely contributed to the greater pressures experienced beneath all MT heads when wearing the CAB. Durometer and drop test results (Table 3.1) showed that the CAB midsole was harder than the GT midsole, and previous research indicates that increased midsole hardness results in greater forefoot plantar pressures (De Wit, De Clercq & Aerts, 2000; Wiegerinck et al.,

2009). As peak pressure at the forefoot has been shown to reduce with the use of even degraded cushioning insoles by between 3.3 – 23% (House et al., 2002), the results of this study suggest that cushioning insoles may be beneficial to reduce forefoot pressures with the CAB. Off-loading orthotic devices (e.g. Ashry, Lavery, Murdoch, Frolich & Lavery, 1997) or a rocker-sole design (e.g. Praet & Louwerens, 2003) have been shown previously to reduce forefoot loads, however orthotic devices must be perfectly fitted to avoid discomfort and blistering, and the CAB already has a rocker design incorporated (see Figure 3.1).

The greater loading rate of forefoot pressures in the CAB may also have been linked to the results of COP analysis. Results showed that there was no difference in the mean position of the COP ($PCOP_Y$) or the time that the COP crossed the MTP joint ($TCOP_Y$), however the COP did move significantly more rapidly towards the forefoot in the CAB, therefore hypothesis (f) was partially accepted. Figure 3.6 highlights that the time of peak $VCOP_Y$ occurred within the first 50 ms of ground contact. It also suggests why $TCOP_Y$ was not affected, as the greater peak in the CAB was followed by a trough of lower velocity, while the GT remained relatively constant throughout this period. Both sample $VCOP_Y$ time histories in Figure 3.6 are similar to that reported by De Cock et al. (2008), the main difference being a greater magnitude of the peak in the first 50 ms evident in the CAB. This phase typically coincided with the forefoot contact phase (FFCP) in De Cock et al. (2008), and although the confident acquisition of the timing of phases such as FFCP is not possible when analysing pressure insole data, this suggests a link to forefoot loading. In particular, a more rapid

transition to the FFCP may be directly linked to the greater loading rates of pressure experienced at the MT heads in the CAB, especially when combined with the presence of a harder midsole material.

The greater $VCOP_Y$ in the CAB implies a greater need for eccentric control of initial ankle motion in this condition, which was reflected by the stiffer ankle during the DFP. Although the ankle joint was stiffer, the $VCOP_Y$ was still greater in the CAB, suggesting either that the $VCOP_Y$ would have been much greater without muscular control, or that a less stiff knee joint, for example, allowed overall leg stiffness to be maintained and bodyweight to rock forward with the CAB. Previous research focusing on leg stiffness indicates that overall 'leg-surface stiffness' is maintained when running on surfaces of different stiffness or shoes with different hardness. Ferris et al. (1998, 1999) found that runners decreased their leg stiffness to accommodate a stiffer surface, even with the first step on that surface, while Smith & Watanatada (2002) showed greater leg stiffness in subjects when wearing soft shoes compared to hard shoes. Furthermore, when modelling the relative contributions of the knee, hip and ankle joints to overall leg stiffness in hopping, Farley & Morgenroth (1999) found that ankle joint stiffness made the primary contribution to overall leg stiffness. Although research specifically considering ankle joint stiffness in response to changes in footwear or surface is lacking, results of the current study may suggest that overall leg stiffness was maintained, despite increased ankle stiffness. Knee stiffness data were not calculated in this study, and should be incorporated into future research.

It is evident from this study that differences in the design aspects of the two footwear conditions account for variation in loading and movement at the forefoot and ankle. The effect of standard issue military footwear on risk factors for lower limb injury is an important research area, given the potential of footwear to affect injury risk, and the cost associated with injured military recruits. Future research should consider a systematic approach to investigating which military boot design aspects could potentially contribute to injury risk, thus allowing potential customisation of these aspects towards a more effective boot.

3.4.1. Limitations

While the present study provides information on the potential footwear-related risk factors associated with the development of MT3SFs in RM recruits, there are limitations to the approach. One weakness of the current approach is the lack of a systematic investigation of each design feature in turn (e.g. raised heel, leather upper), thus assumptions are made that the combination of factors are responsible for altered loading. In addition, the relatively small sample size is a limitation, due to the increased risk of type 2 error. However, plausible significant results were obtained which, in conjunction with previous research, lead to the expectation that the addition of data sets may only strengthen the observed relationships. The observation of large (Cohen's $d > .80$) effect sizes accompanying all statistically significant results adds further confidence in the data.

The GT is representative of commercially available footwear. The peak pressure values obtained in this condition compare well with other trainers (Weigerinck et al., 2009), no above-ankle material was incorporated into the design and the force required to bend the outsole would be classified as mid-range amongst those reported by Oleson et al. (2005) when testing four types of running shoe using a similar method. The choice of a non-issue trainer as a third condition may have provided more insight into how the GT performs compared to other footwear, however with the variety of modern designs available on the footwear market, this selection would be difficult to justify.

In attempting to apply the current findings to RM training, consideration must be given to various additional factors that may influence recruits' gait when wearing the CAB or GT. For example, many exercises are performed over uneven, muddy, hilly or frozen ground, and when carrying heavy loads. Additionally, the CAB in particular may have a different influence on lower limb biomechanics after a significant period of wear.

Although they provide a useful tool for the monitoring of in-shoe pressure during gait, pressure insoles have some limitations. The foot-insole-shoe midsole interaction influences results, where the pressure insole provides a layer of cushioning, however this is likely to be minimal and is systematic across conditions. 'Synchronisation' between the insoles and the force plate cannot be guaranteed to the same degree of accuracy as that between the force plate and kinematic data. Difficulty in defining footsteps using the *footscan* software may have introduced error due to the pre-loading of the insole in the shoe. This may

have varied due to differences in how well the CAB or GT fit the individual participants. Further research should aim to investigate the direction of the horizontal force vector in relation to the foot rather than the laboratory. The use of three-dimensional kinematic data would allow this, and would allow greater understanding of MT3SF injury mechanisms. Knowledge of overall leg stiffness, and the angle of force application relative to the longitudinal axis of the MT3 would be of particular interest, given the previous work of Arangio et al. (1998).

3.5. Conclusion

The results of this study suggest that RM recruits may be at greater risk of MT3SF when wearing the standard issue Royal Marine Combat Assault Boot, compared to a standard issue neutral running shoe. The nature of MT3 loading was investigated with regard to potential mechanisms for MT3SF. While it was expected that the high-cut leather upper would restrict ankle dorsiflexion in the CAB, the effect of the raised heel was deemed influential in both the lower ankle joint range of motion and greater forefoot plantar pressures experienced when wearing the CAB. Altered ankle joint kinetics and kinematics were suggested to contribute to the greater peak plantar pressure, impulse and peak loading rate at the MT3 head region in the CAB. The forefoot was also subjected to a more laterally applied horizontal force vector, relative to the direction of travel, when wearing the CAB. COP analysis suggested that the CAB encouraged more rapid loading of the forefoot, with potential implications for leg stiffness. Further research should be conducted to inform and improve CAB design. Altered boot lacing strategies may be implemented to reduce the restriction of sagittal plane

ankle motion, whilst retaining frontal plane support to reduce inversion injury risk, and cushioning or off-loading orthotic devices may be implemented to reduce the forces acting on the MT heads. More permanent solutions in terms of revised boot design should be possible with sufficient research support.

CHAPTER FOUR: A MODEL OF STRESS ACTING ON THE THIRD METATARSAL WHEN RUNNING BAREFOOT IN THE COMBAT ASSAULT BOOT AND IN TRAINERS.

4.1. Introduction

Stress fractures occur as a result of the inability of bone to recover from damage sustained during repeated submaximal loading. Such damage may occur due to either the characteristics of the applied load or the bone's inability to withstand it. Ideally, direct *in vivo* measurement techniques, such as the implantation of strain gauges in cortical bone, could be used to indicate the strain acting on bone for a given activity, and thus provide direct cause-effect information on how certain loads effect bone. However, as discussed in section 2.3.1 this is associated with numerous limitations and restrictions, meaning that these studies are rare. An alternative approach allowing the estimation of bone loading is mathematical modelling. Mathematical models of bone loading require information about the properties of the bone and the nature of the applied load. Previous examples have used laboratory-based biomechanical techniques allowing the estimation of the magnitude and direction of loads (e.g. Gross & Bunch, 1989), while the ability of the bone to withstand load has been estimated by mechanical tests of strength or scanning techniques providing information on bone geometry and quality (e.g. Milgrom et al., 1989; Arangio et al., 1998). A limitation to previous models of metatarsal loading is that these approaches have not been combined, with arbitrary values for loading (Arangio et al., 1998) or bone geometry (Gross & Bunch, 1989) incorporated into the

model. Mathematical models often have a number of limitations, are often complex and, depending on the data required for calculations, may be restricted for use with a low number of participants.

Despite their weaknesses, mathematical models may provide greater insight into injury mechanisms than traditional biomechanical methods. Third metatarsal stress fractures are an under-researched area, and knowledge of injury mechanisms is therefore limited. Where *in vivo* methods are unavailable or inappropriate, biomechanists are forced to make inferences and deductions about the causes of MT3 stress fractures by, for example, looking at patterns amongst externally measured variables that can be assumed to have a mechanical link to the development of high loads on the bone. However, one of the significant advancements in the understanding of MT3 stress fracture aetiology arose due to the mathematical modelling of the MT3 by Arangio et al. in 1998. Using computerised tomography (CT) scanning, these authors obtained information on the cross-sectional area of metatarsals two to five at 5 mm intervals. For each cross-section, they obtained coordinates for discrete locations on the inner and outer surfaces of the bones, and simulated the shear and normal stresses experienced when loads were applied at the metatarsal head at 15 degree intervals between the horizontal and the vertical directions. Their model indicated that stress in the MT3 was highest 3.5 cm from the proximal end of the bone, under a laterally-applied (90 degree) load. This finding links well with data from Study 1, and that provided by Dixon et al. (2006), which indicate that a more laterally applied horizontal ground reaction force vector may be a risk factor for MT3 stress fracture. However, were it not

for the Arangio et al. (1998) model, this mode of loading may not have been considered as a risk factor.

The model presented by Arangio et al. (1998) built upon earlier work focusing specifically on the fifth metatarsal (Arangio et al., 1997), in which beam theory was used to calculate stress acting on slices of a plastic mould of the bone taken at 1.5 mm intervals. Points along the inner and outer surfaces of each slice were obtained and used to determine a local coordinate system for each cross-section, the centroid, and subsequently the moments acting about the three axes of the bone for a given load. Gross and Bunch (1989) had previously used inverse dynamics and beam theory to estimate saggital plane bending moments and subsequent bending strain acting on the metatarsals during gait. These authors used reference data for bone characteristics, represented the metatarsals as simple uniform ellipses, and only estimated strain at the midpoint of the metatarsal. Despite the interesting findings of work by Gross & Bunch (1989) and Arangio et al. (1998), there has yet to be a study which has combined accurate geometric, bone orientation and external load data to estimate third metatarsal bone stress in individuals during locomotion. The development of such a model of bone loading, while challenging, may provide significant advances in the understanding of MT3SF aetiology.

The aims of this study were twofold. The first aim was to develop a model to estimate bending stress acting on the third metatarsal during running when barefoot (BF). This model would improve on previous attempts in the literature through the use of participant-specific bone geometry, plantar loading and

kinematic data for running. The second aim was to apply the model to running when shod in the Royal Marines standard issue combat assault boot (CAB) and gym trainer (GT) in order to compare estimated MT3 bending strain in each condition. In light of the evidence provided in Study 1, it was hypothesised that peak MT3 bending stresses would be greater in the CAB than the GT.

4.2. Methods

4.2.1. Participants

The seven participants detailed in Study 1 were recruited from a cohort of undergraduate sports science students at the University of Exeter. Five of these (age: 18.8 ± 0.83 yrs; mass: 79.8 ± 2.28 kg) were involved in this study, following examination of MRI data (see below). In addition to data obtained and analysed in Study 1, the same participants agreed to participate in data collection that would inform the development of a model indicating strain on the MT3. Participants were aware that data would be used for two different studies (see Appendix E(i) – Study 1 and 2 information sheet). Due to the requirements of Study 1, only males were selected, in order to mirror the typical RM recruit, and all volunteers were heel-toe runners with size 11 (UK) feet, familiar with wearing and running in military boots. All volunteers completed an MRI safety questionnaire prior to testing (Appendix E(ii)). The study was given ethical approval by the Sport and Health Sciences Ethics Committee, University of Exeter.

4.2.2. MRI data

Bone geometry was obtained from MRI scans. A pilot study was conducted to ensure that appropriate scans could be attained with the minimum time demand on participants. One male volunteered to be a participant in the pilot, allowing appropriate foot position and scan settings to be determined. As a result, the following methods were developed.

With the participant lying supine on a bed, their right foot was first prepared with cod-liver oil markers. As previously shown (Dixon, 1996) the capsules are safe for use in MRI scanning, and are clearly visible on the images obtained. Cod-liver oil markers were secured using masking tape in the following locations: dorsal aspect of the proximal end of the MT3 (articulation with the lateral cuneiform); dorsal aspect of the distal end of the MT3 (third metatarsophalangeal joint); medial aspect of the articulation of the first metatarsophalangeal joint, lateral aspect of the articulation of the fifth metatarsophalangeal joint. Marker placement locations were estimated by palpation of the foot, and verified by subsequent observation of the MRI scans obtained. The cod-liver oil markers served a dual purpose: to provide reference points for identifying the MT3 during scanning, and to relate the location of externally located joint markers to the underlying bone during gait analysis. After preparation with markers, the participant's right foot was fixed to a rigid plastic block with Velcro ties so that the ankle was kept at approximately a 90 degree angle (neutral), and the plantar surface of the foot was in contact with a surface. This was an attempt to replicate the position of the foot during stance.

Skin markers provided a reference to ensure that the MT3 was captured, and the scanned section was defined with respect to the axes of the foot. Scans were obtained at 5 mm intervals in three planes relative to the foot: sagittal, transverse and frontal. Sample images are presented in Figure 4.1.

With reference to the images presented in Figure 4.1, the following MRI scan sequences were used to obtain scans in each plane. For the frontal plane scan: 0.5 x 0.7 mm in-plane resolution was used to obtain 16 slices 3 mm thick. The sequence was a turbo spin echo sequence, with an echo time of 20 ms, a repetition time of 500 ms, and averaged over 4 separate acquisitions. For the transverse plane image: the in-plane resolution was 1.0 x 1.1 mm, obtaining 50 slices of 2.5 mm thickness. The sequence was a T1 weighted gradient echo sequence, with an echo time of 25 ms, a repetition time of 20 ms, and averaged over 2 separate acquisitions. For the sagittal image: 0.5 x 0.6 mm in-plane resolution, 3 mm thick slice, 16 slices. The sequence was a turbo spin echo sequence, with an echo time of 20 ms, a repetition time of 500 ms, averaged over 3 separate acquisitions. The MRI sequence parameters were determined during pilot testing.

The frontal plane scans obtained at 5 mm intervals were converted from raw DICOM images to AVI clips using ImageJ software (v. 1.46r, National Institutes of Health, USA). All relevant scans of the MT3 were concatenated in sequence (proximal to distal) and converted to an AVI clip before being digitised in AVI Digitiser (RF Spectrum Modelling, UK). AVI Digitiser allows sub-pixel digitising using image interpolation, providing high levels of precision. Ninety-six points in

total were digitised, 32 of the inner surface of the bone, and 64 of the outer surface. In order to promote even spacing of digitised locations, points were digitised in sequence, as demonstrated by the schematic in Figure 4.2(A). After digitising each slice of the MT3, x, y coordinates were exported for each of the 96 points and further analysed to provide a description of the geometry of the bone. Rather than simplifying the geometry of each cross section of the metatarsal to a hollow ellipse (as in Gross & Bunch, 1989), the 96 reference points were used to define 96 irregular triangles (Figure 4.2(B)) from which the area and moment of inertia of the slice were determined. The area of each triangle was calculated using Heron's formula (Equation 4.1), and the cross-sectional area of each slice (CA) represented by the sum of the areas of these 96 triangles.

$$\text{Area of triangle} = \sqrt{(s(s-a)(s-b)(s-c))} \quad \text{Eq. 4.1}$$

Where s is half the perimeter of a triangle with sides a, b, c.

The centroid of each triangle was calculated from the mean of the vertices, and the length of the vector between this point and the centroid of the slice determined to represent its distance from the slice centroid. This distance was used to calculate the moment of inertia about the horizontal (I_{xx}) and vertical (I_{yy}) axes for each slice. The area of each triangle was multiplied by the square of the distance of its centroid from the slice centroid in the relevant direction. The slice inertia was derived from the sum of 96 triangle areas.

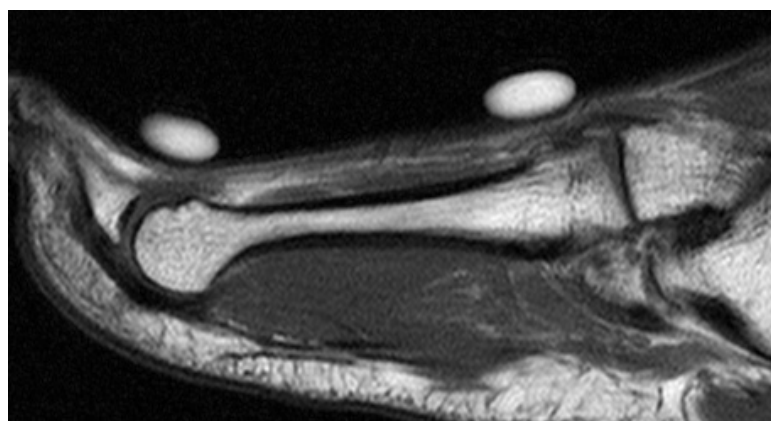
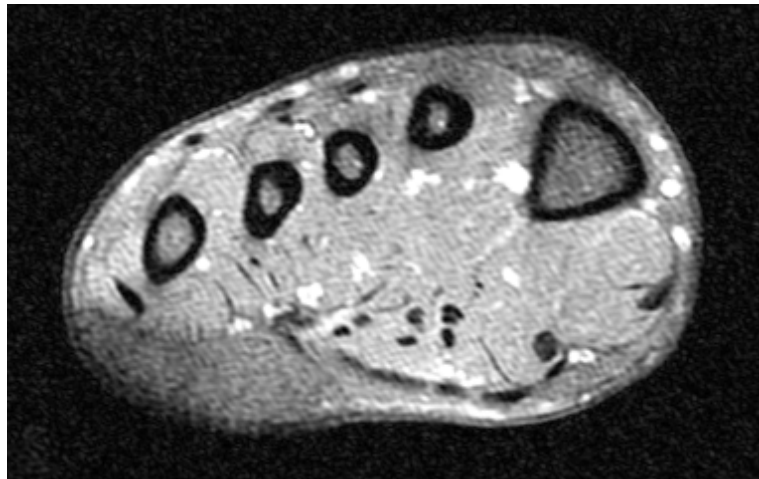
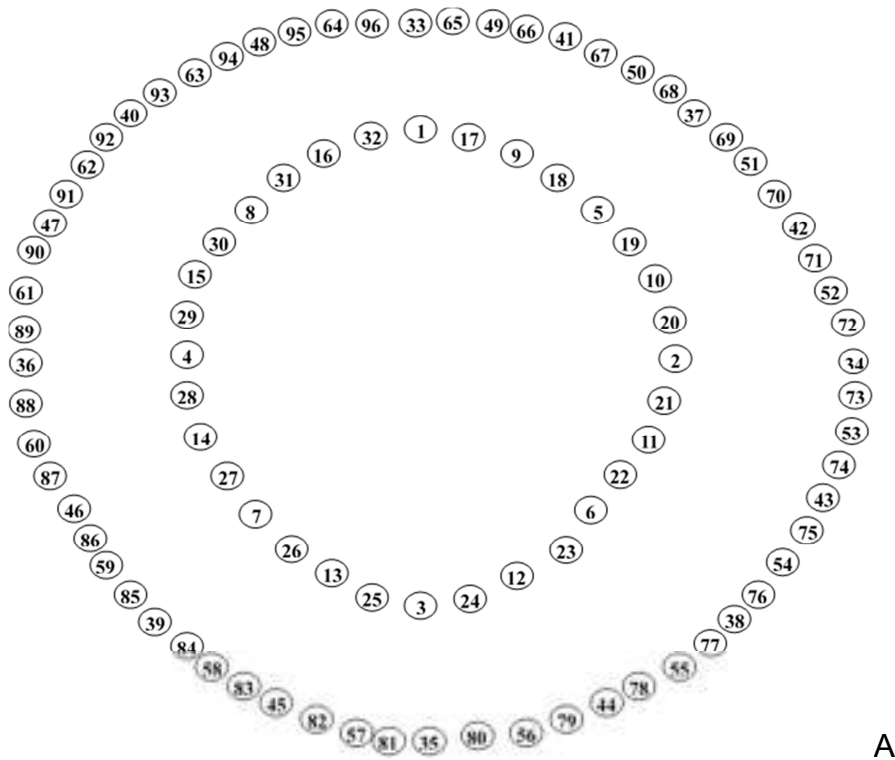
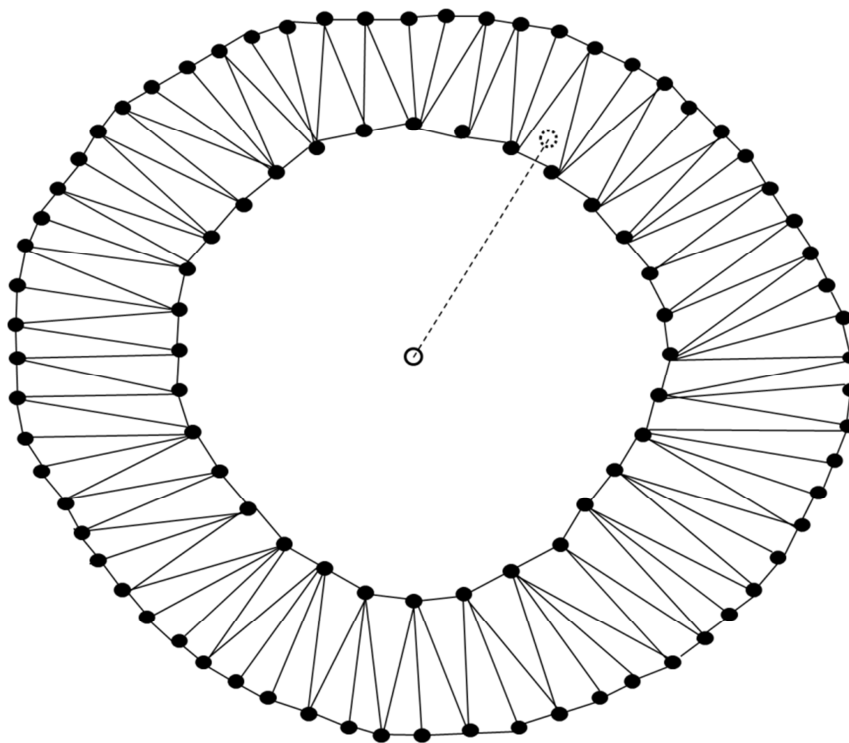


Figure 4.1. Sample images of the third metatarsal of the right foot of one male participant. Top: frontal plane view of mid-section; middle: transverse (plantar) view; sagittal plane view, showing cod-liver oil skin markers.



A



B

Figure 4.2. A) Schematic of digitised points. Numbers represent the order in the sequence in which the point was digitised. B) The 96 triangles used in the calculation of cross-sectional area and moment of inertia. For one example triangle, the distance from the slice centroid (circle) to the triangle centroid (dashed circle) is highlighted, as used in the calculation of moment of inertia.

4.2.3. *Dynamic gait data*

Simultaneous kinematic, ground reaction force and plantar pressure data were obtained for running in the CAB, GT and barefoot (BF). For a description of the methods used to obtain force and pressure data in the shod conditions, refer to Study 1 (p. 103). The collection of barefoot data differed only in that a pressure plate sampling at 500 Hz (0.5 m long, 4096 sensors, RSScan, Belgium) was used instead of the in-shoe pressure insole device previously described. The pressure plate was placed directly on top of the force plate, flush within a 9 mm thick EVA runway with a Shore A rating of 40, measured with a durometer (Durotech, model B202, Hampden Test Equipment Ltd., England). The runway was utilised in conjunction with several habituation trials to protect runners from the potential discomfort of running on the hard laboratory floor and subsequently altering their footstrike pattern, as suggested by Nunns et al. (2013). The pressure plate was dynamically calibrated prior to data acquisition using prescribed methods which involved entering the mass of the tester, who then performed a walking trial over the plate.

As described in Study 1, kinematic data were collected at 120 Hz for the right foot while running at a constant velocity of $3.6 \text{ m}\cdot\text{s}^{-1}$ ($\pm 5\%$). Kinematic data were interpolated to 500 Hz to match force data. In all conditions, markers were placed at the following locations in order to determine the three-dimensional orientation of the foot: superior posterior calcaneous; inferior posterior calcaneous; inferior lateral calcaneous; lateral malleolus; dorsal aspect of the proximal end of the MT3 (articulation with the lateral cuneiform); dorsal aspect

of the distal end of the MT3 (third metatarsophalangeal joint); medial aspect of the articulation of the first metatarsophalangeal joint, lateral aspect of the articulation of the fifth metatarsophalangeal joint. The dorsal MT3 markers identified each end of the bone, and thus were used to calculate the inclination of the metatarsal during gait. The angle of MT3 inclination indicated by skin markers was verified against the MRI scan for each individual. The sagittal plane angles provided by the cod-liver oil capsules, the midline of the MT3 shaft and the sole of the foot were measured and compared. The results of this analysis are presented in Table 4.1, and indicate that markers represent the angle of inclination of the MT3 to within an average of 0.04 ± 3.2 degrees (range 8.2 degrees). For the analysis of shod running, it was judged that the shoe upper prevented accurate placement of markers, which would also fail to provide accurate MT3 inclination data. Therefore, using the data from Table 4.1, the foot angle, determined by markers placed in line with the inferior calcaneus and fifth metatarsophalangeal joint, was corrected using the relevant bone angle for each participant and used to represent MT3 inclination angle.

Table 4.1. Inclination of the MT3 as determined by cod-liver oil capsules and the shaft of the bone. Individual angles are the mean of three measurements, are presented in degrees and represent the relative angle between the MT3 and the sole of the foot. The differential represents the extent to which the marker angle overestimates MT3 bone inclination. Metatarsal length (mm) is also included for reference. The group mean (SD) is presented for all variables.

| Participant | Marker angle | Bone angle | Differential | Metatarsal length |
|------------------|-------------------|-------------------|--------------------|-------------------|
| 1 | 21.5 | 23.8 | -2.3 | 69 |
| 2 | 21.6 | 20.4 | 1.2 | 73 |
| 3 | 24.0 | 23.3 | 0.7 | 63 |
| 4 | 19.3 | 21.3 | -4.0 | 76 |
| 5 | 27.9 | 23.7 | 4.2 | 71 |
| Mean (SD) | 22.9 (3.3) | 22.5 (1.6) | -0.04 (3.2) | 70.3 (4.9) |

The external force acting on the MT3 was assumed to be applied at the plantar surface of the MT3 head. Estimation of vertical load at this location was performed using plantar pressure analysis. For each participant the transverse plane MRI scan best showing the plantar surface of the metatarsal heads was digitised. The location of the centre of the MT3 head was determined as follows:

- Location of the x-coordinate of the MT3 head: a percentage distance along a line from the most medial location of the most distal third of the truncated foot (without the phalanges) and the most lateral location of the most distal third of the truncated foot
- Location of the y-coordinate of the MT3 head: a percentage distance along a line from the most posterior aspect of the heel and the most anterior aspect of the most distal third of the truncated foot

Table 4.2 shows the calculated location for each participant. For each running trial, the relative locations of the x, and y coordinates of the MT3 head were measured on-screen in the *footscan* software (v7. RSScan, Belgium), and the central point of the mask for the third metatarsal head was placed at this location. A 'size 3' mask (3.4 cm²) was used as this was deemed to best represent the surface area of the metatarsal head. Vertical force data for this mask location was exported and scaled to vertical force data obtained simultaneously from the force plate, using peak impact force as a reference.

Table 4.2. Coordinates of MT3 head location for each participant, relative to foot dimensions.

| Participant | X-coordinate of MT3 head (% distance from medial forefoot aspect to lateral forefoot aspect) | Y-coordinate of MT3 head (% distance from posterior calcaneous to anterior truncated foot) |
|------------------|--|--|
| 1 | 53.7 | 86.9 |
| 2 | 55.6 | 87.1 |
| 3 | 53.4 | 84.7 |
| 4 | 56.9 | 87.2 |
| 5 | 57.9 | 83.9 |
| Mean (SD) | 55.5 (1.96) | 86.0 (1.55) |

4.2.4. Model development

The third metatarsal was modelled as a fixed cantilever, and as such when it is singly loaded at the distal end (when vertical ground reaction forces are applied to the base of the MT3 head) it experiences axial loading; bending resulting in

compression and tension; and torsional loading. The resistance of the bone to bending is governed by its 'flexural stiffness' – the product of Young's modulus and the area moment of inertia; while its resistance to torsion or twisting is governed by its torsional stiffness – the product of the shear modulus and the polar moment of inertia. The Young's modulus and shear modulus were assumed to be homogenous properties of bone and therefore uniform along the length of the metatarsal, therefore the resistance to bending and twisting was determined by the cross sectional area and the distribution of material away from the neutral plane of the bone (represented by the centroid). The area moment of inertia and polar moment of inertia will therefore increase not only with greater quantity or density of bone material, but distribution of this material further from the centroid; hence a hollow cylinder is more resistant to bending and twisting than a regular cylinder of the same mass and density. Bone material distributed further from the neutral axis will provide the greatest resistance to bending and simultaneously experience the greatest stress; therefore in long bones where mid-shaft stress fractures occur in cortical bone, it is desirable to calculate the bending and torsional stresses acting at these locations.

The model described below considered the MT3 as a cantilever fixed at the proximal end, with the distal end being free and point-loaded. For each cross-section of the bone obtained through MRI scanning, three axes originating from its centroid were defined. The x-axis was horizontal, the y-axis was vertical and the z-axis was parallel to the longitudinal axis of the bone. In a similar approach to that of Milgrom et al. (1989) when modelling the tibia, bending stresses at the

cross-section of the MT3 were calculated relative to the x-axis and the y-axis. Torsional stress was calculated about the z-axis. Although plantar pressure analysis only provided data for superior/inferior bending (about the x-axis) calculations, the same load was used in a hypothetical simulation of horizontal bending (about the y-axis). This analysis was included to consider the relative resistance of the metatarsal under loading in both the vertical and horizontal directions. In addition to the assumptions made when modelling the MT3 as a fixed cantilever, the following further assumptions were made:

- no consideration was paid to the influence of surrounding structures, such as muscle or ligamentous attachment, or bone contact forces;
- the geometry of the metatarsal was well represented by 5 mm intersections;
- only vertically applied ground reaction forces were considered, and plantar pressure analysis gave adequate representation of these;
- bone acts in accordance with Hooke's law within normal physiologic loading

When a non-axial force is applied at the end of the bone, in this case assumed to be the metatarsal head, a bending moment is produced and normal and shear forces are transmitted along the length of the bone. At any cross section of the metatarsal, the bending moment can be determined by equation 4.2:

$$M_{be} = F(L - x) \quad \text{Eq. 4.2}$$

Where F is the applied force, derived from the vertical force time history acquired by plantar pressure analysis; L is the length of the metatarsal and x is the perpendicular distance from the section to the point of load application. The perpendicular distance x was calculated with trigonometry using the angle of the metatarsal relative to the ground. When a bending moment is applied, tensile stress is developed on the surface on the outside of the curve, compressive stress on the inside of the curve, and axial stress through the neutral axis. In the case of metatarsal loading, the application of vertical ground reaction force causes compression on the dorsal surface of the bone and tension on the plantar surface. The standard equations for these three stress parameters are displayed below:

$$\sigma_{ax} = \frac{F}{A} \quad \text{Eq. 4.3}$$

$$\sigma_c = \sigma_{ax} + \sigma_{be} \quad \text{Eq. 4.4}$$

$$\sigma_t = \sigma_{ax} - \sigma_{be} \quad \text{Eq. 4.5}$$

where σ_{ax} is axial stress; σ_c is compressive stress; σ_{be} is bending stress; σ_t is tensile stress; F is the force applied and A is the cross-sectional area of the bone. In order to determine maximal σ_{be} the following general equations were used:

$$\sigma_{be}(\text{about } x\text{axis}) = \frac{(M_{be} \cdot y)}{I_x} \quad \text{Eq. 4.6}$$

$$\sigma_{be}(\text{about } y\text{axis}) = \frac{(M_{be} \cdot y)}{I_y} \quad \text{Eq. 4.7}$$

where y is the maximal distance of the cross section from the neutral axis in the relevant direction and I_x or I_y is the area moment of inertia about the neutral axis in the relevant direction. Maximal torsional stress was calculated using the following equation:

$$\sigma_{tor} = \frac{(M_{be} \cdot R)}{I_z} \quad \text{Eq. 4.8}$$

where I_z is the polar moment of inertia about the neutral axis, and is the sum of I_x and I_y and R is the radius of the outer surface of the bone. R was determined by calculating the mean of the radial distances of the outer 64 points from the centroid.

For each cross-section of the MT3 of each participant, the use of the equations above provided data for the three stress categories (σ_{ax} , σ_t , σ_c) as a result of the vertical force considered to be acting at the distal end of the MT3. Figure 4.3

displays a simplified diagram of the components of the model. For each stress category, the magnitude and location (distance from distal end of MT3) of peak stress was identified. This process was repeated for horizontal compressive and tensile stress, with the assumption that the same load was applied (in lieu of horizontal force data). Repeated analysis of one participant's data revealed very low variation in calculated values, giving confidence in the reliability of the process (Appendix A). All calculations involved in the derivation of these values were performed using custom Matlab scripts (v.2008b, The Mathworks Inc, USA). For the comparison of footwear conditions, peak vertical compressive, vertical tensile and torsional stresses were compared for each participant. An approach to statistical analyses similar to Study 1 was adopted. The ankle angle and pressure variables were shown not to be normally distributed in Study 1, therefore one-tailed Wilcoxon tests were again performed in SPSS (v.21, IBM, USA) using an alpha level of 0.05. Effect sizes (Cohen's d – Cohen, 1988) were produced for all variables in order to further aid interpretation. Based on Cohen's (1988) guidelines, a medium effect size was considered to be .50.

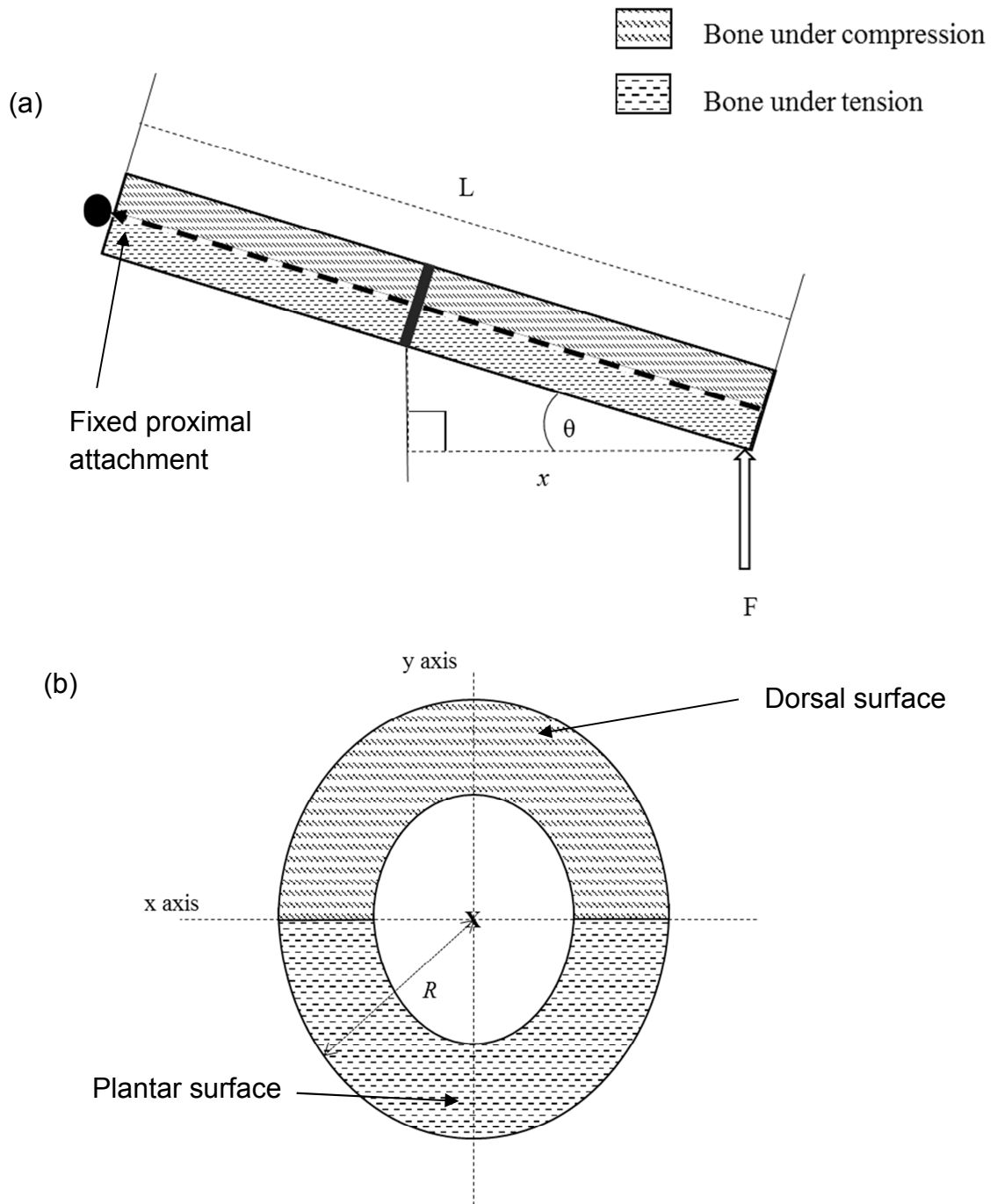


Figure 4.3. Schematic showing the inputs and measurements considered in the model. (a) Displays the free-body representation of the third metatarsal, in which the axially-directed dashed line (← - -) represents the line of axial stress; θ = angle of inclination of metatarsal to the ground; F = vertical force obtained from pressure data; L = length of metatarsal; x = perpendicular distance of slice from point of force application; the thick black intersection represents a sample slice taken from the mid-shaft. This slice is depicted from a frontal perspective in (b), where X = Centroid of slice; R = radius of outer surface of metatarsal. The axes about which bending stress moment arms (y) and inertial properties are calculated are also identified.

4.3. Results

4.3.1. Stage 1 results – barefoot running

Table 4.3 summarises the maximum and minimum cross-sectional areas (CA) for each metatarsal. Due to the calculation of axial stress, the location of minimum area was also the location of maximum axial stress. In four cases the minimum area was in the slice adjacent to the distal slice, with participant 4 also having a low CA at a distal location. Mid-point and minimum CA were similar between participants, however large variation was seen in the maximum CA.

Table 4.3. The cross-sectional area (CA) at the minimum, maximum and mid-point locations of each individual's third metatarsal, and the magnitude and location of peak axial stress are displayed. Mean (SD) values for the group are also presented. Locations are the distance from the distal tuberosity.

| Participant | Min CA (mm ²) | Location (mm) | Max CA (mm ²) | Location (mm) | Mid CA (mm ²) | σ_{ax} (MPa) | Location (mm) |
|------------------|---------------------------|---------------|---------------------------|---------------|---------------------------|---------------------|---------------|
| 1 | 22.66 | 5 | 83.77 | 70 | 38.69 | 5.41 (1.94) | 5 |
| 2 | 28.73 | 5 | 176.26 | 70 | 39.65 | 4.71 (1.33) | 5 |
| 3 | 30.96 | 5 | 65.65 | 0 | 38.65 | 4.89 (1.76) | 5 |
| 4 | 32.46 | 15 | 94.73 | 65 | 46.19 | 4.09 (1.14) | 15 |
| 5 | 27.05 | 5 | 66.06 | 50 | 47.70 | 6.57 (2.39) | 5 |
| Mean (SD) | 28.37 (3.81) | - | 97.29 (45.83) | - | 45.54 (4.31) | 5.13 (0.93) | - |

σ_{ax} = axial stress.

The mean magnitude and timing of peak plantar force are included in Table 4.4, in addition to the times of heel off and peak stress. Heel-off always preceded peak force, with peak stress and peak force occurring at similar times. In four participants, peak stress preceded peak force by approximately 2% of stance, however in participant 2 this order was reversed, with peak stress occurring 1.2% after peak force. The maximum range of peak forces was around 45 N, with participant 5 the highest and 1 the lowest. The peak stress magnitudes do not reflect this rank order of magnitudes however, with compressive, tensile and torsional stresses highest in participant 1 and lowest in participant 4 (Table 4.5). Horizontal bending stresses were highest for participant 3 and lowest in participant 4 (Table 4.6). Horizontal stresses were greater in magnitude than vertical stresses and, with the exception of participant 2, occurred at a different location to that of peak vertical stress. Additionally, peak torsional stresses did not always occur at the same location as either vertical or horizontal bending stresses.

Table 4.4. The mean (SD) of ten trials of the peak plantar force, time of peak force, time of heel off and time of peak stress is shown for each participant during the stance phase of barefoot running. The mean (SD) values for the group are also included.

| Participant | Peak force (N) | Time of peak force (%) | Time of heel off (%) | Time of peak stress (%) |
|------------------|-----------------------|------------------------|----------------------|-------------------------|
| 1 | 128.99 (42.21) | 58.76 (2.01) | 50.11 (1.85) | 56.67 (2.12) |
| 2 | 135.57 (31.09) | 57.09 (1.06) | 48.55 (1.57) | 58.21 (3.89) |
| 3 | 151.54 (54.58) | 58.37 (2.83) | 51.63 (2.26) | 56.58 (3.67) |
| 4 | 145.80 (53.97) | 57.79 (1.93) | 43.40 (1.54) | 54.89 (2.31) |
| 5 | 173.65 (68.86) | 63.39 (3.55) | 55.17 (1.80) | 61.87 (7.84) |
| Mean (SD) | 147.11 (17.23) | 59.08 (2.49) | 49.77 (4.32) | 57.64 (2.64) |

Table 4.5. The mean (SD) of ten trials of the magnitude and location (mm from the distal end of the metatarsal) of peak vertical compressive and tensile stresses and peak torsional stress for the right third metatarsal of each participant during barefoot running. Mean (SD) peak stress values for the group are also presented. Tensile stresses are negative.

| Participant | σ_c (MPa) | σ_t (MPa) | Location (mm) | σ_{tor} (MPa) | Location (mm) |
|------------------|-----------------------|------------------------|---------------|----------------------|---------------|
| 1 | 138.73 (35.36) | -132.88 (33.82) | 45 | 68.83 (17.53) | 35 |
| 2 | 106.41 (31.45) | -100.43 (29.78) | 50 | 48.12 (14.24) | 50 |
| 3 | 115.18 (42.63) | -108.70 (40.27) | 25 | 57.06 (21.13) | 35 |
| 4 | 77.35 (22.16) | -72.23 (20.78) | 30 | 34.76 (9.98) | 30 |
| 5 | 134.49 (63.01) | -125.41 (58.85) | 35 | 62.05 (29.09) | 35 |
| Mean (SD) | 114.43 (24.66) | -107.43 (23.77) | - | 54.16 (13.22) | - |

σ_c = compressive stress; σ_t = tensile stress; σ_{tor} = torsional stress.

Table 4.6. The mean (SD) of ten trials of the magnitude and location (mm from the distal end of the metatarsal) of peak horizontal compressive and tensile stresses for the right third metatarsal each participant whilst running barefoot. Mean (SD) peak stress values for the group are also presented. Tensile stresses are negative.

| Participant | σ_c (MPa) | σ_t (MPa) | Location (mm) |
|----------------------|---------------------------|----------------------------|------------------|
| 1 | 171.39 (43.68) | -164.61 (41.90) | 35 |
| 2 | 133.37 (39.43) | -127.40 (37.76) | 50 |
| 3 | 182.80 (67.67) | -176.08 (65.23) | 35 |
| 4 | 89.82 (25.75) | -86.07 (24.74) | 45 |
| 5 | 181.34 (69.00) | -137.36 (64.23) | 25 |
| Mean (SD) | 151.74 (40.00) | -138.30 (35.24) | - |

σ_c = compressive stress; σ_t = tensile stress; σ_{tor} = torsional stress.

Table 4.7. The mean (SD) of ten trials of the magnitude of peak axial, vertical (V), horizontal (H) and torsional stresses at the metatarsal mid-point for each participant whilst running barefoot. Mean (SD) values for the group are also presented. Tensile stresses are negative.

| Participant | σ_{ax} (MPa) | V σ_c (MPa) | H σ_c (MPa) | V σ_t (MPa) | H σ_t (MPa) | σ_{tor} (MPa) |
|----------------------|------------------------|---------------------------|---------------------------|---------------------------|----------------------------|--------------------------|
| 1 | 3.17 (1.14) | 138.15 (35.21) | 171.39 (43.68) | -129.38 (31.84) | -164.61 (41.90) | 68.83 (17.53) |
| 2 | 3.42 (1.08) | 88.58 (24.73) | 115.60 (32.10) | -86.90 (25.78) | -122.62 (36.35) | 42.20 (11.69) |
| 3 | 3.65 (1.32) | 110.96 (41.07) | 160.38 (59.37) | -103.73 (38.44) | -153.48 (56.74) | 53.89 (19.96) |
| 4 | 2.00 (0.71) | 71.03 (20.28) | 85.19 (24.33) | -65.52 (18.79) | -79.58 (22.82) | 33.74 (9.65) |
| 5 | 3.64 (1.44) | 117.70 (50.44) | 138.98 (65.10) | -111.26 (52.22) | -124.43 (43.91) | 52.86 (24.78) |
| Mean (SD) | 3.18 (0.69) | 105.28 (26.07) | 134.31 (34.78) | -99.36 (24.30) | -128.94 (33.06) | 50.30 (13.25) |

σ_{ax} = axial stress; σ_c = compressive stress; σ_t = tensile stress; σ_{tor} = torsional stress.

4.3.1. Stage 2 results – shod running

The results for running in the CAB and GT are summarised below. Focus has been maintained on compressive and tensile stresses under vertical loading, and torsional stress. The locations of peak stress are determined by metatarsal geometry and therefore are the same as for barefoot running. Table 4.8 summarises the peak plantar force beneath the MT3 in each condition, which was greater in the CAB for each recruit, the mean difference between CAB and GT being 27.65 N. The mean times of peak force and peak stress are similar between conditions, however individual results show variation (Table 4.8). For example, participant 1 experienced a difference in the timing of peak force of ~7.5% stance, however peak stress occurred at almost identical times. All peak stresses were significantly greater in the CAB than the GT, with a medium effect size reported (Table 4.9). This was a consistent finding at individual level and overall.

Table 4.8. The mean (SD) of ten trials of the peak plantar force, time of peak force, time of heel off and time of peak stress is shown for the right third metatarsal of each participant whilst running in the combat assault boot (CAB) and gym trainer (GT). The group mean (SD) is also included.

| Participant | Peak force (N) | | Time of peak force (%) | | Time of heel off (%) | | Time of peak stress (%) | |
|------------------|---------------------------|--------------------------|-------------------------|-------------------------|-------------------------|-------------------------|-------------------------|-------------------------|
| | CAB | GT | CAB | GT | CAB | GT | CAB | GT |
| 1 | 161.11 (14.76) | 128.56 (10.70) | 51.78 (2.32) | 58.20 (4.62) | 56.00 (4.18) | 47.60 (3.41) | 60.66 (3.56) | 60.69 (3.48) |
| 2 | 144.13 (13.34) | 125.58 (18.48) | 55.20 (2.91) | 57.51 (1.04) | 62.10 (2.41) | 60.17 (1.23) | 66.64 (2.16) | 64.67 (5.24) |
| 3 | 165.34 (11.97) | 133.11 (16.95) | 63.07 (4.12) | 66.57 (3.32) | 68.43 (2.67) | 52.98 (1.99) | 67.84 (2.35) | 67.78 (4.07) |
| 4 | 150.23 (16.55) | 115.48 (1.46) | 55.66 (1.48) | 57.97 (2.84) | 56.29 (3.17) | 55.37 (3.01) | 56.60 (2.07) | 60.93 (9.32) |
| 5 | 133.78 (13.50) | 113.64 (11.63) | 70.75 (4.12) | 68.07 (5.96) | 70.49 (3.35) | 69.30 (6.80) | 59.06 (3.83) | 66.96 (4.92) |
| Mean (SD) | 150.92 (12.77) | 123.27 (8.42) | 59.29 (7.61) | 61.66 (5.20) | 62.66 (6.71) | 57.08 (8.19) | 62.16 (4.88) | 64.20 (3.30) |

Table 4.9. The mean (SD) of ten trials of the magnitude of peak vertical compressive and tensile stresses and peak torsional stress for each participant whilst wearing the combat assault boot (CAB) and gym trainer (GT). Mean (SD) peak stress values for the group are included. The results of a Wilcoxon paired tests are presented in the bottom row, with effect size (Cohen's *d*) included.

| Participant | σ_c (MPa) | | σ_t (MPa) | | σ_{tor} (MPa) | |
|-----------------------|---------------------------|--------------------------|----------------------------|---------------------------|--------------------------|--------------------------|
| | CAB | GT | CAB | GT | CAB | GT |
| 1 | 181.20 (12.22) | 141.00 (14.90) | -173.62 (12.05) | -135.21 (14.36) | 89.44 (6.56) | 70.00 (7.42) |
| 2 | 120.10 (12.23) | 109.30 (16.44) | -113.81 (11.61) | -103.65 (16.58) | 54.42 (5.55) | 49.54 (7.47) |
| 3 | 117.00 (10.37) | 92.54 (19.23) | -109.90 (9.87) | -88.04 (18.06) | 57.81 (5.16) | 46.27 (9.50) |
| 4 | 79.10 (5.34) | 60.59 (4.37) | -73.71 (5.00) | -55.81 (4.70) | 37.84 (2.56) | 28.99 (2.09) |
| 5 | 78.14 (9.52) | 61.72 (8.13) | -71.79 (9.12) | -53.75 (7.62) | 36.08 (4.40) | 28.42 (3.76) |
| Mean (SD) | 115.11 (42.01) | 93.03 (33.91) | -108.67 (41.32) | -87.29 (34.21) | 55.11 (21.49) | 44.64 (17.16) |
| <i>P</i> (<i>d</i>) | .022* (.58) | | .022* (.56) | | .022* (.54) | |

σ_c = compressive stress; σ_t = tensile stress; σ_{tor} = torsional stress.

4.4. Discussion

Axial, bending and torsional stresses were estimated along the MT3 of the right foot of five males with size 11 feet during running, using a simple model based on beam theory. MRI scans were used to estimate geometric properties, and dynamic loading and kinematic data acquired for running initially whilst barefoot, and then when shod in the RM recruit standard issue CAB and GT. Variation in the geometric properties of the MT3, even within a sample of individuals with the same foot size, lead to large variations in the estimated peak bending stresses. Additionally, in support of the hypothesis, it was shown that wearing

the CAB resulted in significantly greater estimated stresses than wearing the GT.

This model provides a simplified estimation of third metatarsal loading using beam mechanics. The use of individual bone geometry and gait data is particularly valuable in highlighting the variation in stresses experienced by individuals during gait. Previous models of third metatarsal loading have provided useful information which has informed theory on third metatarsal stress fracture, albeit with certain limitations. For example, Arangio et al. (1998) analysed just one set of metatarsals and did not account for the dynamic nature of gait in their calculations, while Gross & Bunch (1989) used dynamic plantar pressure data but relied on reference data for the geometric properties of their participants' metatarsals using static estimations of metatarsal inclination. The present study is the first to incorporate both individual geometry and individual kinematic and kinetic data into a model of MT3 loading, therefore providing the opportunity to investigate the influence of variations in these factors on metatarsal bending stresses. Furthermore, the utilisation of in-shoe plantar pressure data during running to inform the model is unique, and provides some insight into the differences in MT3 loading when shod in standard issue RM footwear.

Stage 1 of the analysis highlights several interesting points. First, examination of the locations of maximum stress and minimum CA highlight the importance of the geometry of bone, in particular its arrangement about the centroid and its position relative to the point of load application in determining its resistance to

bending. Although the lowest CA of each metatarsal was at the distal end, peak stresses occurred at much more proximal locations, within the middle five slices of the metatarsal. Although the bending moment systematically increases with each more proximal slice of the metatarsal, the occurrence of peak stresses in the mid-shaft for all participants is indicative of the role of bone geometry in determining peak stresses. The differing locations of peak stress for the various stress types further highlights the non-homogeneous nature of individual MT3 geometry. Although the vertical load was used arbitrarily in horizontal stress calculations, peak horizontal stresses were greater in magnitude and occurred at different locations to the vertical stresses, corroborating the analysis performed by Arangio et al. (1998). The greater horizontal stresses reflect the anisotropic nature of bone – horizontal loading occurs to a lesser extent than vertical loading, therefore the metatarsal has remodelled and strengthened in these frequently loaded sections of bone, leading to greater inertial properties in the vertical direction.

Examination of the peak plantar forces (Table 4.4) and mid-section CA (Table 4.3) also demonstrate the importance of factors other than MT3 head loading in determining bone bending stresses. Peak force was highest in participant 5 and lowest in participant 1, with participant 4 having the middle value. Mid-section peak stresses were highest in participant 1, with participant 5 producing the middle value. Participant 4 however experienced around 50% of the stress magnitude of participant 1. This suggests that the combination of geometry, angle of inclination and force is important in determining mid-shaft loading. This

is particularly important when considering the utilisation of peak pressure values alone to assess metatarsal stress fracture risk (as in Study 2).

The variation in the generated peak stresses may be considered large, given that the five participants had the same shoe size and their metatarsal lengths were within 13 mm. If participants 2 and 4 are considered, the difference in their MT3 lengths was only 3 mm, and peak force 10 N, yet their peak vertical tensile stress differed by 19 MPa, equivalent to approximately 1100 microstrain. This variation between individuals within a relatively homogeneous group demonstrates that without knowledge of individual bone geometry, estimations of internal loading may be erroneous. In particular, the arrangement of bone mass about the centroid is crucial information, as this determines the inertial properties of the section. Reference data could be obtained for a given individual of known height, mass, age and foot dimension (as in Gross & Bunch, 1989), however the present results suggest that variation in individual metatarsal geometry may be a notable source of random error. It should be noted that the values estimated here are peak stresses, the inertial component of which is largely determined by the maximum distance of any digitised point from the centroid of the slice of MT3 under consideration. The validation of the model (Appendix A) reports very low variation between three repeat analyses of the same participant, suggesting that digitisation reliability is high. Further examination of data could provide mean stresses, or stresses at set azimuthal intervals (angles relative to the horizontal axis), which may be less sensitive to variations in geometry. However if referring to maximal stresses, the current

results suggest that the use of reference geometry data in order to compare between individuals is not appropriate.

The data for shod running indicate that peak bending and torsional stresses were greater when participants ran in the CAB compared to the GT ($P < 0.05$). The majority of this effect is likely explained by differences in peak plantar force beneath the MT3 in the two conditions, therefore in within-group comparisons such as this, external loading is an important determinant of the magnitude of metatarsal bending stress. Arndt et al. (2003) were able to relate external plantar loading to metatarsal deformation with the use of bone-mounted strain gauges during shod walking. This valuable study noted that surface bone strains on the second metatarsal were increased by fatigue when wearing two models of military boot, with plantar loading also increasing. Although individual responses to footwear were quite variable in the Arndt (2003) paper, and impulse rather than peak pressure was reported, this is further evidence of the correlation between plantar loading and bone strain. The model validation performed in the current work showed that maximal compressive stress increased by around 0.6 MPa per Newton of added force (Appendix A). This, combined with the results for shod running, suggests that increased plantar loading will increase MT3 bending stresses and subsequently stress fracture risk. Whether or not this is harmful will also be determined by the inclination of the MT3 at the time of peak loading, and perhaps more importantly by the geometry of the bone. Strategies to reduce MT3 stress fracture risk should look to reduce plantar loading as this variable can be influenced by footwear interventions. One such example of the influence of cushioning insoles in the

RM recruit population was demonstrated by Windle et al. (1999), who observed that they reduced plantar pressures at the heel and forefoot. The efficacy of cushioning insoles in reducing metatarsal stress fracture risk in particular is not guaranteed however, as this type of injury was not significantly reduced in the RM recruit population when such insoles were prescribed (House et al., 2013). Custom orthotics which provide forefoot cushioning in areas of high pressure identified through plantar pressure analysis may be a more effective strategy in reducing this type of injury risk, with further research required in this area.

Barefoot data reveal that peak force and peak stress occurred at similar times during stance, with both events occurring after heel off, however the sequence was less clear for the shod conditions presented in Stage 2. When wearing the GT, peak force and peak stress occurred at similar times for each participant (within 2-7% of stance), but with little relationship to the time of heel off. In the CAB, there was no pattern in evidence between the time of heel off, time of peak force or time of peak stress. The small sample size adds to the difficulty in interpreting any relationship between the timing of events, however it is of interest to note that peak stress occurred at near identical stance times in both footwear conditions for three participants, the other two varying by 4-8%. In participants 2, 4 and 5, the time of peak force was similar between conditions (CAB 2% earlier for 2 and 4, and 2% later for 5), but the time of peak stress varied by differing amounts (2% later for 1, 4% earlier for 4, 8% earlier for 5), suggesting that another variable, such as the inclination of the MT3 within the footwear, may play an important role in determining the time of peak stress. The inherent lack of validity in estimating foot position using shoe-mounted markers

may also have influenced these results. Further evidence is required to understand whether footwear can influence the timings of peak stresses at the MT3, particularly if these are associated with increased stress fracture risk. For example, customised footwear allowing access to the surface of the foot for accurate marker placement (i.e. through removal of the last), would allow the influence of manipulations in sole properties to be assessed using the current model.

The model presented here did not take into account surrounding structures or muscular attachment. The model of Gross & Bunch (1989) could be considered superior in this regard because it accounted for the influence of toe forces at the metatarsophalangeal joint, as well as the influence of the plantar attachment of the tibialis posterior. The muscular attachment of the tibialis posterior was given a standard location and the magnitude of force applied by it derived from an equation including other estimated properties (Gross & Bunch, 1989), therefore the value of including this in the model could be questioned. There is evidence that the tibialis posterior plays an important role in reducing bending stress acting on the metatarsals. Donahue & Sharkey (1999) observed an increase in metatarsal strains measured with bone-mounted gauges when plantar attachments were removed, building on earlier work by Sharkey (1995) demonstrating the capacity of plantarflexor muscle contractions to reduce metatarsal strains. Therefore the role of plantar musculature should be considered in future, but in order to achieve meaningful data, it should be accurately modelled if possible, perhaps with the use of EMG.

The mean peak compressive stress value reported for barefoot running was 114.43 Pa which, if a Young's modulus of 17 GPa is assumed (as in Gross & Bunch, 1989), yields a peak strain of 6749 $\mu\epsilon$. Gross & Bunch (1989) reported mean peak strain at the MT3 to be 5160 $\mu\epsilon$, despite reporting greater mean peak plantar force than the present study (200 N compared with 147 N). *In vivo* second metatarsal strain data have been reported during walking to reach around 2000 – 2500 $\mu\epsilon$ (Sharkey & Ferris, 1995; Donahue & Sharkey, 1999; Donahue et al., 2000; Milgrom et al., 2002; Arndt et al., 2003). There is no published data for *in vivo* MT3 strains during walking or running to which comparison could be made. However, second metatarsal strains have been modelled to be greater than the MT3 (Gross & Bunch, 1989), and the transition from walking to jogging has been shown to double the observed MT2 strains (Milgrom et al., 2002). Therefore it could feasibly be expected that strains in the range of 3500 to 5000 $\mu\epsilon$ could be expected on the shaft of the MT3 depending on running speed, in line with the estimations of Gross & Bunch (1989) but below the predictions of the current model. This needs to be verified with future work, and it is accepted that the simplifications of the current model will lead to over-estimation of peak MT3 bending loads. It is likely that the additional consideration of muscular attachment contributed to the differences in magnitudes observed in the two studies, although the use of accurate geometric data in the present model may also have been important. Without suitable detailed information on muscle forces and attachments (for example), integrating further estimations into the model would be inappropriate.

In addition to the limitations already discussed, the design of the model incorporated a number of assumptions which should be considered when evaluating the accuracy of predicted bone stresses. Applying beam mechanics to the problem, the MT3 was considered to be a fixed cantilever, with a single load point at the distal end (Figure 4.3). In reality, the MT3 is subject to bone contact forces at the proximal end and laterally at the articulation with the adjacent metatarsal bones distally. There is also an influence of plantar musculature and ligamentous attachment. The decision not to include these factors in the model was taken because of the difficulty in estimating values for these forces. As such, the model produced is simple but clear, and provides a basis for development of a more complete assessment in future iterations.

The loading and motion of the MT3 were considered only in the sagittal plane, while stresses were calculated in the sagittal, frontal and transverse axes, relative to the laboratory reference frame. The frontal cross-section of each MT3 slice was divided into sections based on the coordinates of the centroid, with these remaining fixed throughout. Subsequently, the MT3 was assumed to retain the orientation it was in at the point of MRI scanning throughout stance, and any rotation of the MT3 relative to the assumed vertical force vector was ignored. This assumption will lead to errors when estimating the vertical and horizontal distances of the cortical wall from the centroid during stance (variable 'y' in the stress calculations), and therefore in peak calculated stresses. Given the elliptical nature of the MT3, rotation out of the assumed position may lead to changes in y of a few mm. For a hypothetical mid-shaft slice with a bending moment applied of 4.5 N.m (150 N force, 0.03 m moment arm) and an inertial

value about the x-axis of 150 mm^4 , a change in y from 3 mm to 4 mm would result in a change in bending stress of 30 MPa. It is not acceptable to fail to account for this level of variation in the model, therefore estimations of peak bending stresses are most applicable to a foot position identical to that in which the MRI scans were taken. To improve this aspect of the model, the axial rotation of the MT3 should be tracked during gait, and the relevant rotation of cross-sectional bone coordinates performed.

A further issue with the two-dimensional nature of the model is that torsional stresses are likely to be under estimated. Rotation of the forefoot relative to the rearfoot in the frontal plane has not been considered, however Pohl, Messenger & Buckley (2006) reported poor coupling between forefoot and rearfoot, with around 10 degrees of variation in the amount of eversion during stance when running. A difference between the position of the proximal attachment point of the MT3 (medial cuneiform) and the distal metatarsophalangeal joint may impart significant torsional stress on the MT3, depending on whether the MT3 head is assumed to be fixed within the metatarsophalangeal joint. The tough ligamentous attachments with the adjoining metatarsal heads mean that some torsional force will be imparted on the MT3 if the forefoot rotates relative to the rearfoot, however the quantity of this torque would be very difficult to estimate, and rely on yet more assumptions. The angle of twist (difference between inversion/eversion of the midfoot and forefoot) could be estimated with kinematic data, and entered into a standard equation to estimate torsional stress due to this differential. The consideration of each slice of the MT3 as an

object more complex than a hollow ellipse leads to difficulty in accurately modelling stresses, however this data could strengthen the present model.

Horizontal forces were not integrated into the model, although it was established in Study 1 that horizontal forces may be of importance in the aetiology of MT3SF, building on the work of Arangio et al. (1998) and Dixon et al. (2006). Integration of horizontal force data into the present model, rather than arbitrarily using the vertical loading magnitudes, may strengthen the predicted stress values. Effective and accurate integration of horizontal loads would require knowledge of the location, magnitude and direction of the resultant horizontal force vector relative to the MT3; estimation of the damping of this force before it reaches the MT3 head; and the relative ab-adduction of the forefoot with regard to the rearfoot. The use of pressure insoles in the shod conditions allowed the identification of load directly beneath the MT3, however horizontal forces would have to be obtained using a force plate. This would mean that the damping properties of the footwear would have to be modelled to consider the nature of the force before it reaches the foot. Although considerations such as these are challenging, they could be implemented with future development of the model.

The application of beam theory has been demonstrated in previous biomechanical models of long bone bending stresses (e.g. Gross & Bunch, 1989; Milgrom et al., 1989), however while this approach is computationally straightforward, it may not be as accurate as a finite element (FE) modelling approach. Beam theory requires the bone to assume a rectangular beam

shape, and while the model presented here negated this to some extent by considering the dimensions of several intersections (slices) of the beam, a FE approach could implement a mesh analysis technique, such as the one used in Brassey et al. (2013) to more accurately determine stress across the entire surface of the bone. Indeed, Brassey et al. (2013) attempted to quantify the difference in estimations of bone stress using beam mechanics and FE modelling, finding that the simpler approach underestimated compressive stresses by up to 142%, with differences between methods directly correlated ($r^2 = .56$) to the curvature of the shaft. Aside from the influence of shaft curvature and asymmetrical geometry, the methods agreed to within 12-14%, implying that the failure to consider the curvature of the MT3 may lead to large error in bending stress predictions. Given that the MT3 has significant curvature, bone stresses predicted by a FE modelling approach may be much higher than those reported here. Referring again to the already high peak stress values, the consideration of the plantar attachments, which serve to reduce bending stress on the MT3, may be even more important in future. It should be considered that, while potentially more accurate, FE modelling is much more computationally demanding than the application of beam mechanics to the problem.

4.5. Conclusions

The MT3s of five participants were successfully modelled using beam theory to estimate axial, bending and torsional stresses acting upon them during running when barefoot and in standard issue military footwear. This was the first model

utilising the individual geometry data of several feet as well as individual dynamic gait data to provide such information, although the model itself was very simplified. Geometric bone properties were within acceptable ranges indicated by previous research, while the magnitude of predicted peak stresses was higher than previous values, likely due to the absence of plantar musculature in the model. Data indicate the importance of plantar loading, inclination of the MT3 and particularly individual bone geometry in determining the magnitude of peak stresses.

While increased plantar loading and decreased MT3 inclination will cause bone stress to increase, the variation in geometry between even a homogenous group of individuals may be the dominant factor in determining whether the levels of stress experienced are potentially damaging. When comparing within individuals or groups, footwear adaptations which permit a reduction in plantar loading and/or increased angle of inclination of the foot may be effective in reducing damaging loads. When comparing data between individuals however, inferences about internal metatarsal strain made from external load data may be highly inaccurate unless knowledge of the properties (bone geometry and quality) is obtained.

CHAPTER FIVE: A PROSPECTIVE INVESTIGATION INTO PREDICTORS OF TIBIAL AND THIRD METATARSAL STRESS FRACTURES IN ROYAL MARINES RECRUITS

5.1. Introduction

It is well documented that individuals involved in regular running are at risk of overuse injury, including lower limb stress fracture. The 32-week training programme for Royal Marines (RM) infantry recruits is continuous in nature, and involves various activities such as running, marching and jumping, over a range of exercise intensities, load carriages and terrain. A high incidence (30%) of lower limb injury has been reported in RM recruits, with lower limb stress fractures accounting for 4-8% of these (Ross & Allsopp, 2002). The third metatarsal (38%) and tibia (30%) were the most common sites of stress fracture in this report. Recent figures indicate that the incidence and distribution of injuries, including stress fractures, has remained relatively constant or increased, with around 6% of 1416 recruits sustaining a stress fracture and the third metatarsal and tibia again the two most commonly affected site (House et al., 2013). A recruit who sustains a stress fracture will be removed from training and given time for rest and rehabilitation. The Ministry of Defence (MoD) has a duty of care to reduce the risk of injuries during training, and with full recovery and reintegration into training following stress fracture requiring 12-20 weeks (Ross & Allsopp, 2002), the associated costs of rehabilitation are a further motive to reduce injury risk. It is reported (personal communication) that each week in rehabilitation costs the MoD around £1500, therefore if 60 stress fractures occur in a year, each requiring 20 weeks for recovery, £1.8 million

would be expended on rehabilitation from stress fracture. In order to facilitate the reduction of lower limb stress fractures an increased understanding of factors contributing to this type of injury is required. If individuals susceptible to stress fracture can be identified before participating in periods of physical activity, appropriate interventions may be applied to reduce the likelihood of stress fracture development.

Previous research has been undertaken in military and civilian populations in order to understand and therefore reduce the risk of stress fractures. Whilst studies analysing stress fracture risk factors have been performed in a number of military institutions, including US Marine Corp recruits (Beck et al., 1996; 2000), the Israeli Defence Force (Milgrom et al., 1994), the US Navy (Kaufman et al., 1999) and other military recruits (Almeida et al., 1999; Mahieu et al., 2006), Chapter 2 highlights that confounding variables such as terrain, weather conditions, the availability of rest periods and load carriage make it difficult to apply any interpretations regarding injury risk to those undertaking the RM infantry training programme. It is therefore necessary to study the population of interest directly, where external factors are relatively controlled. In addition, while biomechanical aspects of gait are likely to have an influence on injury risk in populations with relatively controlled extrinsic conditions, there have been no prospective studies to assess gait factors in military cohorts, possibly due to the time required to undertake such an investigation. By collecting baseline biomechanical data at the beginning of a controlled training programme, it may be possible to identify factors that predispose individuals to injury. This

approach is preferred to retrospective studies that attempt to draw conclusions about gait characteristics that may have changed since the occurrence of injury.

A full consideration of risk factors for stress fracture, including those specific to each injury, is found in section 2.4 (p. 62). The review of literature demonstrates that, in general, stress fracture development is influenced by either the ability of bone to withstand load, or the characteristics of the applied load (magnitude, direction etc). Despite both being long bones, the tibia and third metatarsal are loaded differently during gait and therefore the specific aetiology of tibial stress fractures (TSF) and third metatarsal stress fractures (MT3SF) is different, although factors such as younger age (Milgrom et al., 1994) and lower lean muscle mass (Milgrom, 1989; Bennell et al., 1996) as indicated by lower BMI may increase the risk of lower limb stress fractures in general. Due to their greater prevalence in most populations, a number of aspects of lower limb structure and function associated with TSF have been identified, including low tibial cross-sectional area (Milgrom et al., 1989); low calf circumference (Milgrom, 1989); retroverted hips (Giladi et al., 1987b); excessively high (Milner et al., 2010) or low (Hetsroni et al., 2008) rearfoot movement; pes cavus (Korpelainen et al., 2001) or pes planus (Sullivan et al., 1984); as well as impact characteristics (Zifchock, Davis & Hamill, 1996). Few established risk factors for MT3SF have been identified in the literature but there are proposed relationships between restricted passive dorsiflexion and increased forefoot loading (Hughes, 1985); and both the earlier occurrence of peak eversion and more laterally applied ground reaction forces have been identified as important characteristics in a retrospective study (Dixon et al., 2006). Increased forefoot

pressures and restricted dynamic dorsiflexion have also been linked with increased third metatarsal loading (Nunns et al., 2012), while pes cavus has been linked with metatarsal stress fractures in general (Sun et al., 2012). The role of both passive and dynamic ankle function has been cited in MT3SF aetiology, however relationships are poorly understood and should be further examined. A particular omission in research into MT3SF aetiology is the frequent failure to consider this bone as a separate from the other metatarsals, despite evidence implying that the metatarsals are all uniquely susceptible to certain load characteristics (Arangio et al., 1998).

Although general risk factors exist for stress fracture, there are certain individual characteristics that have been previously identified as unique to TSF and MT3SF aetiology. These commonly focus on the characteristics of the lower limb assessed both passively and during gait, and should be investigated in the study of specific injury causes amongst RM recruits. In the case of MT3SF, where previous literature is lacking, loading of the metatarsal head is likely to be important, and kinematic variables which influence this should be identified. Consideration of mechanisms for stress fractures at specific sites, rather than grouping different injury types together, should facilitate greater understanding of the nature of injury at each site. It will be particularly beneficial to consider factors unique to MT3SF development, given the uniquely high incidence of this injury amongst RM recruits.

The aim of this study was to prospectively identify risk factors that predispose RM recruits to the occurrence of tibial and third metatarsal stress fractures. In

addition, it was desirable to explore previously suggested relationships between ankle flexibility (both passive and dynamic) and forefoot loading.

It was hypothesised that, for TSF cases compared to controls: age would be younger; BMI would be lower; bimalleolar distance would be lower; calf circumference would be lower; passive external hip rotation (EHR) would be greater; passive internal hip rotation (IHR) would be lower tibial rotation relative to the rearfoot would be lower; there would be a difference in dynamic ankle angle at touchdown and peak dorsiflexion; there would be a difference in rearfoot kinematics; there would be a difference in peak pressures at the heel; there would be a difference in dynamic arch height.

It was hypothesised that, for MT3SF cases compared to controls: age would be younger; BMI would be lower; peak plantar pressure characteristics at the MT3 would be greater; peak forefoot pressures would occur earlier; passive dorsiflexion would be lower; dynamic dorsiflexion would be lower; peak eversion would occur earlier; dynamic arch height would be lower; there would be a difference in ankle kinematics between MT3SF and controls.

It was also hypothesised that there would be correlations between several kinematic variables and pressure variables at the forefoot. In addition, it was expected that the amount of passive dorsiflexion available would correlate with the: magnitude of dynamic peak dorsiflexion; timing of dynamic peak

dorsiflexion; time of heel-off; magnitude of peak pressure at the forefoot; timing of peak forefoot pressures.

5.2. Methods

5.2.1. Participants

Given previous occurrence rates of TSF and MT3SF in RM training (Ross & Allsopp, 2002), approximately twelve TSF cases and fifteen MT3SF cases of each injury would be expected per 1000 recruits. An *a priori* power analysis indicated that this would yield an observed power of 0.7-0.8. Following the provision of ethical approval by the Ministry of Defence Research Ethics Committee, data collection occurred between September 2010 and June 2012 at Commando Training Centre, Royal Marines Lympstone (Devon, UK) (CTCRM). Each of the 28 troops of recruits enrolling in training during that period was visited the week prior to testing and provided with information about the study (Appendix D(iii)). Recruits willing to participate signed a consent form (Appendix D(iv)) and attended testing on day 8 or 9 of their 32 week basic training programme. On data collection days, investigators aimed to test all consenting recruits, but were restricted by time or recruits opting out of the study. In total, 1065 recruits of the 1504 that initially enrolled in training were assessed (70.8%).

As with all injuries sustained during training, recruits with lower limb pain were examined by the medical staff at CTCRM. Presence of stress fracture was confirmed by positive MRI scan. Medical records were consulted following the completion of the final troop involved in the study, and all recruits sustaining a TSF or MT3SF were recorded. Of the 1065 recruits participating in this study, 10 TSF (0.9%) and 15 MT3SF (1.4%) cases were reported. One recruit sustained both types of injury, but was placed in the TSF group as medical records indicated that this injury occurred first. Therefore the injured leg from 10 TSF and 14 MT3SF cases were analysed. The injury distribution by week of training for each injury is displayed in Figure 5.1 (note, data unavailable for one TSF case).

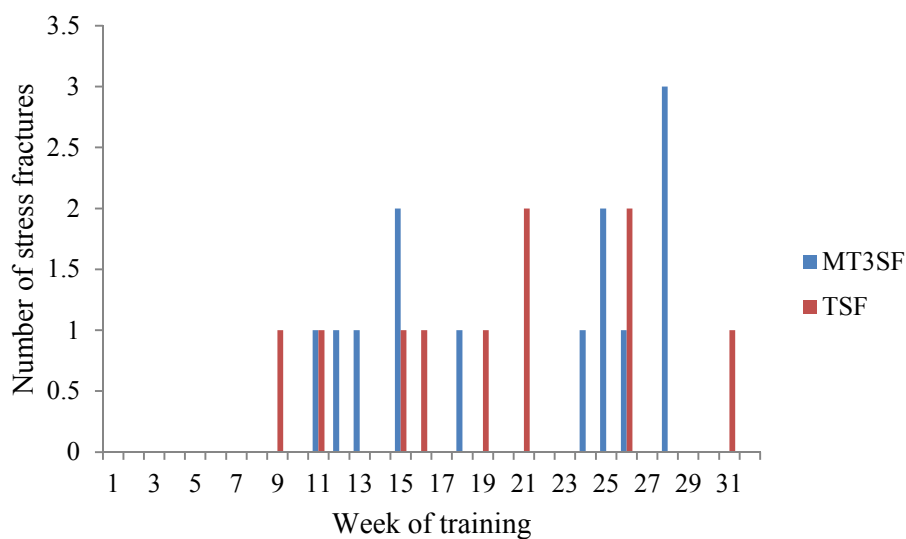


Figure 5.1. Distribution of injuries by week of occurrence.

Controls were selected from the recruits who passed out of training with their original troop, and therefore assumed not to have incurred injuries causing them

to withdraw from training. This type of completion is the desired route through training. Of the original 1065 recruits participating in this study, 419 passed out with their original troop (39.3%). There is considerable variety in the size of control groups used in previous studies. For example, Bennell et al. (1996) compared data for 10 stress fractures affecting men against 39 controls; Valimaki et al. (2005) compared data for 15 stress fractures with 164 controls; while Giladi et al. (1991) compared data for 60 tibial stress fractures with that for 229 without. In each of these examples, the study utilised a prospective cohort design, and controls were comprised of all participants not incurring the injury of interest. In the present study, this approach would represent an unethically time-consuming process. In order to determine a suitable control group size, a stability analysis was performed (Appendix B), revealing that stable, representative data would be obtained from a group containing 120 controls. Each injury group was therefore compared with the control group of 120.

5.2.2. Procedure

As detailed in Chapter 1 (pXX), running features heavily in the training programme, particularly in the first 10 weeks. As the programme progresses, load carriage (20.5 kg) is introduced (in week 5) and then gradually increased in magnitude, duration or frequency as training progresses. After week 15, operational simulations and assessment runs become more frequent, and unspecified distances are covered through walking and marching. Given the proliferation of running-based activities in the RM recruit training programme, a focus on running biomechanics was maintained in the current study.

Testing took place at an on-site facility at CTCRM. Upon arrival at the testing centre, anthropometric, flexibility and dynamic gait data were acquired for each recruit. Participants were required to run barefoot at $3.6 \text{ m}\cdot\text{s}^{-1}$ over a 2 m pressure plate (RSScan International, Belgium, 2 m x 0.4 m x 0.02 m, with 16,384 resistive sensors, 200 Hz, 10 sensors/4 cm²) set within an EVA runway (0.02 m thick, hardness rating of 65 Shore A) covered by a thin rubber mat with a total carpeted length of 15 m with approximately 3 m before and 2 m after the pressure plate and EVA runway. Synchronised bilateral 3D kinematic data were obtained at 200 Hz using two aligned Coda mpx30 units (Charnwood Dynamics Ltd., Leicestershire, UK). Figure 5.2 shows the layout of the testing environment. The accuracy and reliability of the Coda mpx30 system was deemed to be acceptable (Appendix C).

Synchronisation of devices was achieved through a digital signal output from the RSScan control box, which was received by the Coda system. The pressure plate was set to capture, activating the signal. Once the participant began running, the Coda system was set to record, capturing the digital input as well as marker data. The signal switched off automatically upon reaching the system buffer capacity. Each sensor of the plate requires 1 byte of data per frame, and the plate has a total of 16384 sensors. Operating at 200 Hz, with 128 lines, this required 1638400 bytes per second of data to be captured, and with a total buffer of 4096 kilobytes, this allowed 2.5 s, or 500 frames to be captured. The signal was identified in Coda, the point it switched to 'off' was identified, and the instant 500 frames prior to this identified as the frame of ground contact.

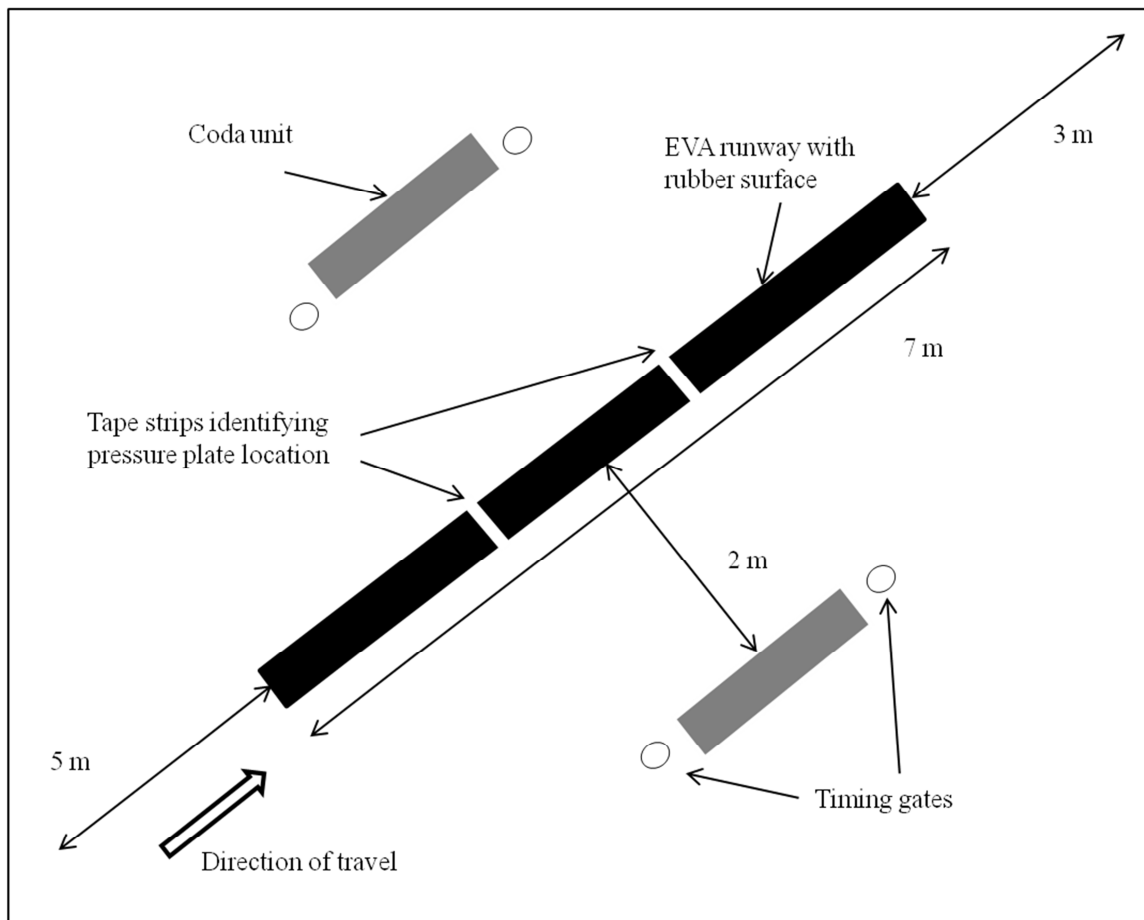


Figure 5.2. Diagram of setup indicating orientation of runway, and locations of pressure plate, Coda units and timing gates. Lengths before and after runway indicate carpeted area.

Barefoot running was preferred over shod running in order to remove the influence of footwear on gait characteristics. However, barefoot running was not a familiar activity for most recruits, initially inducing a midfoot or forefoot strike in many recruits. In order to encourage recruits' 'usual' running style, sufficient habituation trials were afforded prior to data collection. This involved several practice trials, where the assessor observed running style and speed. Participants were encouraged to try to run in a relaxed fashion. With 77% of recruits adopting a heel strike pattern in the study (Nunns et al., 2013), a figure

similar to the distribution previously observed in shod runners (Kerr, Beauchamp, Fisher & Neil, 1983; Hasegawa, Yamauchi & Kraemer, 2007), habituation was deemed successful. Following habituation, five acceptable running trials and one static trial were recorded per recruit. Trials were deemed acceptable if all markers were tracked, both feet landed within the 2 m pressure plate without unnatural stride adjustment, and the target running velocity was met. Static trials were recorded with the recruit standing relaxed with feet shoulder-width apart.

5.2.3. Anthropometric Measurements

Body mass, height, bimalleolar distance and calf circumference were measured by an investigator trained to undertake these procedures. Appropriate steps were taken to ensure that the environment in which these measures were taken, and the measurement approach, conformed to best practice as detailed by *The International Society for the Advancement of Kinanthropometry* (ISAK). Each recruit's mass was obtained in shorts and t-shirt to the nearest 0.1 kg (Sartorius AG, Goettingen, Germany). Height was measured to the nearest 0.01 m using a stadiometer (Seca 202, Seca, Hamburg, Germany) with feet together and feet, buttocks and scapulae in contact with the back of the stadiometer. Body Mass Index (BMI) was calculated by dividing body mass by the square of height ($\text{kg}\cdot\text{m}^{-2}$). Calf circumference was measured at the site of maximum circumference using a Silverflex anatomical metal tape measure (Rabone Chesterman, England). Calf circumference was measured to the nearest mm, using the techniques described in the *Anthropometric Standardization*

Reference Manual (Lohman, Roche & Martorell, 1988). This girth measure was corrected for fat mass by determining the calf skinfold thickness with Harpenden callipers (BodyCare UK), according to the method of ISAK (Figure 5.3). Measurements were made to the nearest mm, and the average of three measurements was taken from each site. Corrected calf circumference (CC) was calculated by subtracting the mean skinfold value from the mean calf circumference for each recruit.



Figure 5.3. Assessment of calf skinfold.

The amount of passive ankle dorsiflexion (PADF) available to each recruit was assessed using a weight-bearing static lunge test based on that described by Bennell et al. (1999). Recruits were instructed to place their foot onto a 15° wooden wedge with the longitudinal axis of their foot parallel to the edge of the block (Figure 5.4). The block was set perpendicular to a wall, and recruits were

instructed to lunge towards the wall with their knee, whilst keeping their foot flat on the wedge. Recruits were instructed to lunge as far as possible without their heel lifting, and were allowed to use the wall for stability. A clinical fluid goniometer (MIE Medical Research Ltd., UK) was placed against the midline of the Achilles tendon and the inclination of the shank recorded to the nearest degree ($^{\circ}$). The procedure was repeated until three trials within a 5° range were recorded for each leg; the mean of the three measurements was reported to represent the PADF for each recruit.



Figure 5.4. Assessment of Passive Ankle Dorsiflexion.

Internal and external hip range of motion was assessed in a seated position (Figure 5.5). Whilst sitting with the hip flexed at 90 degrees, the recruit adopted a position with the knee bent and the lower leg hanging over the end of the bench. This allowed rotation of the hip by medial-lateral movement of the lower

leg with the knee fixed in position. Internal and external hip rotation was measured by the use of a fluid goniometer attached to a stick aligned with the tibia. Values were reported to the nearest degree until three values within 5° had been obtained for internal and external rotation of each leg. The mean internal hip rotation (IHR) and external hip rotation (EHR) was reported for each subject. Additionally, the relative internal-external range of motion (HROM) was calculated for each leg (EHR – IHR). A positive value for HROM indicated a retroverted hip. All anthropometric measurements were obtained by experienced staff involved in frequent training and skill assessment.



Figure 5.5. Assessment of hip rotation (internal rotation shown).

In addition to following the recommended Prior to the start of data collection, a pilot study was performed to assess the reliability of anthropometric data obtained using the study setup. Fourteen (mean \pm SD age 22.8 ± 2.27 years;

mass 76.16 ± 7.43 kg; height 1.78 ± 0.05 m) RM recruits were assessed using the techniques described above to obtain values for calf circumference, calf skinfold, hip range of motion and ankle passive flexibility. Mean values were obtained on three separate occasions. Sessions one and two were separated by four days, with a third session taking place seven days after session two. ICC values were calculated each variable and for each leg for the group across the three sessions. For both left and right legs, ICC scores were deemed excellent (>0.8) for calf girth and bimalleolar distance; strong (0.7-0.8) for calf skinfold and ankle dorsiflexion; and good (0.6-0.7) for hip data.

5.2.4. Kinematic analysis

For the collection of kinematic data, active markers were placed at the following locations on the lower limbs: greater trochanter; medial epicondyle of the knee; lateral epicondyle of the knee; midline of the Achilles tendon, just inferior to the muscle belly of the gastrocnemius (posterior lower leg); midline of the anterior aspect of the tibia, directly opposite the Achilles tendon marker (anterior lower leg); superior posterior aspect of the calcaneus; inferior posterior aspect of the calcaneus; lateral malleolus; lateral aspect of the fifth metatarsophalangeal joint; superior aspect of the proximal end of the third metatarsal; superior aspect of the distal end of the third metatarsal; medial aspect of the first metatarsophalangeal joint (Figure 5.6).

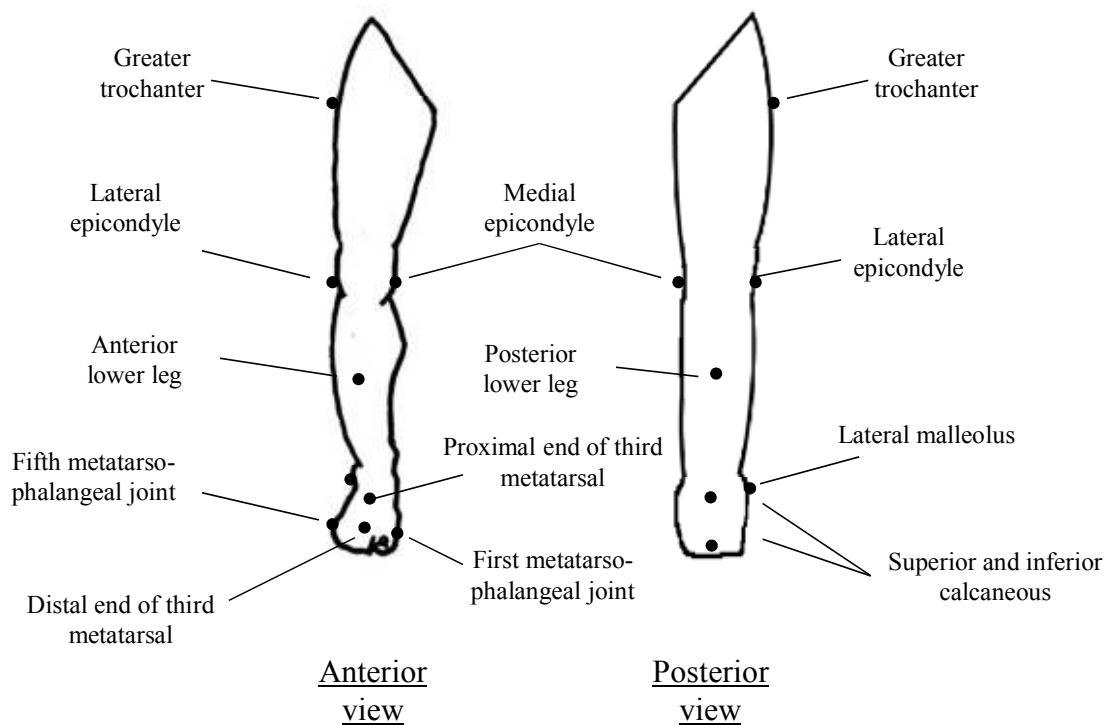


Figure 5.6. Marker Placements on the Anterior and Posterior Aspects of the Lower Limb.

The joint coordinate system was based on the principles established by Grood & Suntay (1986) and Soutas-Little et al. (1987). The thigh segment z-axis was defined by a unit vector created between markers placed on the greater trochanter and lateral epicondyle of the knee, the x-axis by markers on the medial and lateral epicondyle of the knee, and the y-axis by the cross product of these unit vectors. The shank z-axis was defined from the virtual points representing the mid-knee and mid-ankle virtual locations. The mid-knee marker was defined as the mid-point between the medial and lateral knee markers. The mid-ankle marker coordinates were determined as follows:

$\text{Mid-ankle}_x = \text{Mid-knee}_x + (\text{Lateral malleolus}_x - \text{Lateral knee}_x);$

$\text{Mid-ankle}_y = \text{Mid-knee}_y + (\text{Lateral malleolus}_y - \text{Lateral knee}_y);$

$\text{Mid-ankle}_z = \text{Lateral malleolus}_z.$

Having defined the z-axis of the shank, the x-axis was defined as the unit vector running from the mid-ankle virtual marker to the lateral malleolus, with the y-axis being the cross product of the x and z axes. The foot segment z-axis was defined with a line from the inferior calcaneous to the superior calcaneous; the y-axis of the foot defined between the superior calcaneous and the distal third metatarsal marker; the x-axis being the cross product of these vectors. With three axes defined for each segment, 3D angles were calculated using the descriptions provided by Soutas-Little et al. (1987).

The following variables were calculated in order to assess ankle kinematics: ankle angle at touchdown; peak dorsiflexion angle, time of ankle dorsiflexion, normalised to percentage of ground contact time; ankle range of motion from initial plantarflexion to peak dorsiflexion. Rearfoot kinematics were assessed using the following variables: rearfoot angle at touchdown; peak eversion angle, time of peak eversion expressed as a percentage of ground contact time; range of motion from touchdown to peak; average rate of eversion, calculated as the range of motion divided by the time of occurrence of peak eversion in seconds. The z-axis rotation of the shank relative to the foot was calculated to represent tibial rotation. Due to the variability of angles at touchdown for this variable, only the range of motion was reported. This was calculated as the range of motion from the angle at touchdown and the angle of peak internal rotation of the shank relative to the foot. Foot angle relative to the ground was calculated at the

instant of touchdown. Raw coordinate data were exported from the Coda software and filtered using a 12 Hz recursive fourth-order low-pass Butterworth filter. This filter design was based on previous examples used with kinematic data (e.g. Nigg, Stergiou, Cole et al., 2003; Hardin et al., 2004) and preferred to 6, 8, 10 or 14 Hz filters based upon observation of angle time histories produced using these levels. Filtering and angle calculations were performed in Matlab (v2008a, The Mathworks, US).

5.2.5. *Plantar pressure analysis*

For the analysis of plantar pressure data, each foot was initially divided into regions of interest as depicted by Figure 5.7. These were the five metatarsals (M1-5), the medial and lateral heel (HM, HL) and the midfoot (MF). The hallux (T1) and the four lesser toes (T2-5) were allocated but not reported. Zones were automatically assigned by the *footscan* software (version 7, RSScan, Belgium) and manually adjusted using the original foot outline as guidance (Rice, Nunns, House et al., 2013). The intra- and inter-rater reliability of mask and zone analysis of the foot was deemed to be excellent following a reliability study where three repeats of fifty pressure trials were analysed by two observers and assessed. Intra-observer reliability was high (ICC = 0.998) for the investigator, as was inter-observer reliability with another investigator (ICC = 0.992). Following adjustment of zones, a number of measurements were obtained from the *footscan* software.

In order to provide an indication of dynamic arch function, midfoot surface contact area and midfoot impulse were recorded for each trial. Midfoot surface contact area represented the percentage of the plantar foot surface occurring within the midfoot zone. Midfoot impulse represented the percentage of the whole-foot impulse passing through the foot that occurred through the midfoot zone. Foot axis angle was reported by the *footscan* software as the angle between the longitudinal axis of the pressure plate and a line drawn from the centre of the heel through to the M2 head. A negative foot axis angle indicated that the foot was adducted relative to the longitudinal axis of the plate.

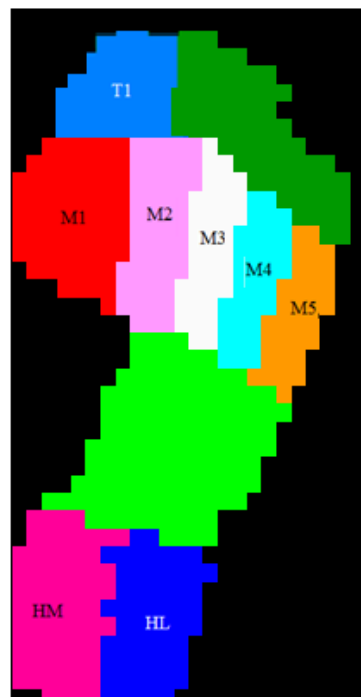


Figure 5.7. Definitions of foot zones identified for analysis.

In addition to midfoot variables, peak pressure, peak force, time of peak pressure (and force), and the force-time integral (impulse) were recorded for the

M1-5, HM and HL regions. All pressure and force values were normalised to body weight. Time of peak pressure/force was calculated as a percentage of ground contact time. As part of the pilot study conducted to obtain anthropometric data prior to the start of main data collection, pressure data were acquired for the 14 participants using the procedures described above. Mean peak pressures at the M1-5, HM and HL regions were shown to have acceptable agreement between sessions, with ICC scores ranging from .363 (M1 region) to .785 (M3 region) across the three sessions. Agreement between the first two sessions was poorer than between sessions two and three. In session one, footstrike type was not consistent, suggesting participants were not allowed enough familiarisation with the protocol in the first session. This finding led to increased familiarisation time being integrated into final data collection, and consistent footstrike type being required. Research indicates that five trials are sufficient for stable plantar pressure data (Hafer, Lenhoff, Song, Jordan, Hannah & Hillstrom, 2013), and that the test-retest reliability of plantar pressure data is acceptable (ICC values > .70) (Wearing, Urry, Smeathers & Battistutta, 1999; De Cock et al., 2006).

5.2.6. Data analysis

For each case included in the analysis, mean values for each variable were calculated. The 10 TSF legs and 14 MT3SF cases were separately compared with the 120 legs included in the control group. All variables were tested for normality using the Shapiro-Wilk statistic ($P < 0.05$). For normally distributed variables, independent *t*-tests were used to identify risk factors for each injury,

while non-normally distributed variables were analysed using Mann-Whitney *U*-tests. Directional hypotheses were assessed with a one-tailed approach, non-directional hypotheses were assessed with a two-tailed approach, all using an alpha level of 0.05. Effect sizes (Cohen's *d* – Cohen, 1988) were produced for all variables. Given the potential limitations of comparing small injury groups with a larger population, significant variables identified by univariate analysis were explored further to determine the consistency of findings. Exploratory plots comparing individual group members with the control mean were produced for variables producing a *P*-value of <0.10. Correlations were explored between ankle and rearfoot kinematics, timing of heel off, passive dorsiflexion and pressure variables at the M3 region using Pearson's tests, with R^2 values reported (two-tailed, $P < 0.05$). All controls and injured cases were included in the correlation analysis, however the number of cases with data for all variables varied (minimum $n=109$) and were reported for significant correlations. All statistical analyses were performed using SPSS (v. 21, IBM, USA).

5.3. Results

5.3.1. TSF v Controls

Of the anthropometric (Table 5.1), kinematic (Table 5.2) and plantar pressure (Table 5.3) variables assessed between groups, analysis revealed that there were five which were significantly different between TSF cases and controls. Bimalleolar width, BMI and corrected calf girth were smaller in the TSF group

($P < 0.05$), with large effect sizes reported for these differences. Tibial rotation range of motion was lower, and peak pressure at the medial heel greater in the TSF group compared to controls ($P < 0.05$), both being associated with medium effect sizes. The results for age were also explored, as this variable was very close to statistical significance, with a medium effect size.

Table 5.1. Mean (SD) for anthropometric variables investigated between the tibial stress fracture (TSF) and control groups. The results of univariate analysis and effect sizes are presented for each variable. Statistically significant differences between groups are in bold.

| Anthropometric variable | TSF (n=10) | Controls (n=120) | <i>P</i> | <i>d</i> |
|----------------------------------|-----------------------|-----------------------|--------------|------------|
| Age (years) | 19.70 (2.00) | 21.36 (3.12) | .054 | .63 |
| Bimalleolar width (mm) | 68.40 (3.50) | 72.16 (4.96) | .006* | .88 |
| BMI (kg.m²) | 22.77 (1.68) | 24.31 (1.82) | .006* | .88 |
| Corrected calf girth (mm) | 352.81 (15.34) | 367.52 (19.53) | .011* | .84 |
| External hip rotation (°) | 26.60 (8.54) | 25.41 (1.82) | .331 | .19 |
| Internal hip rotation (°) | 24.80 (6.70) | 25.92 (7.72) | .493 | .15 |
| Hip range of motion (°) | 51.40 (13.35) | 51.32 (12.88) | .251 | .01 |
| Height (m) | 1.79 (0.08) | 1.78 (0.05) | .373 | .15 |
| Mass (kg) | 74.27 (6.82) | 76.67 (6.62) | .139 | .36 |
| Peak ankle dorsiflexion (°) | 29.30 (7.09) | 31.12 (5.38) | .319 | .29 |

P = statistical significance from either independent samples *t*-test or Mann-Whitney *U*-test; *d* = effect size, indicated by Cohen's *d*.

Table 5.2. The mean (SD) of ten barefoot running trials for kinematic variables investigated between the tibial stress fracture (TSF) and control groups. The results of univariate analysis and effect sizes are presented for each variable. Statistically significant differences between groups are in bold.

| Kinematic variable | TSF (n=10) | Controls (n=120) | <i>P</i> | <i>d</i> |
|---|--------------------|---------------------|--------------|------------|
| Ankle angle at touchdown (°) | -0.26 (1.36) | -0.77 (1.97) | .411 | .30 |
| Peak dorsiflexion angle (°) | -11.85 (3.53) | -11.85 (3.15) | .998 | <.01 |
| Time of peak dorsiflexion (%) | 51.31 (4.16) | 51.59 (4.63) | .868 | .06 |
| Rearfoot angle at touchdown (°) | 1.95 (1.74) | 2.14 (1.84) | .766 | .11 |
| Peak eversion angle (°) | -4.78 (2.53) | -5.62 (2.60) | .265 | .33 |
| Rearfoot range of motion (°) | 6.77 (3.47) | 7.85 (3.65) | .269 | .30 |
| Time of peak eversion (%) | 45.99 (8.87) | 41.60 (7.61) | .132 | .53 |
| Rate of eversion (°/sec ⁻¹) | 61.65 (33.88) | 77.19 (35.39) | .131 | .49 |
| Tibial range of motion (°) | 6.41 (4.30) | 9.72 (5.06) | .029* | .70 |

P = statistical significance from either independent samples *t*-test or Mann-Whitney *U*-test; *d* = effect size, indicated by Cohen's *d*.

Table 5.3. The mean (SD) of ten barefoot running trials for plantar pressure variables investigated between the tibial stress fracture (TSF) and Control groups. The results of univariate analysis and effect sizes are presented for each variable. Statistically significant differences between groups are in bold. Peak pressure and impulse were expressed as a percentage of body weight per square centimetre. Timing of peak pressure was expressed as a percentage of total stance time. Midfoot surface contact and impulse were expressed as a percentage of whole foot contact and impulse respectively.

| Plantar pressure variable | TSF (n=10) | Controls (n=120) | <i>P</i> | <i>d</i> |
|---|--------------------|---------------------|--------------|------------|
| HM peak pressure (%BW.cm⁻²) | 2.92 (1.01) | 2.34 (0.62) | .038* | .69 |
| HL peak pressure (%BW.cm ⁻²) | 3.09 (1.50) | 2.39 (0.75) | .120 | .59 |
| HM impulse (%BW.s) | 3.81 (2.06) | 3.56 (1.43) | .701 | .14 |
| HL impulse (%BW.s) | 2.71 (1.47) | 2.32 (1.03) | .373 | .31 |
| HM time of peak pressure (%) | 8.58 (3.63) | 10.16 (5.84) | .691 | .32 |
| HL time of peak pressure (%) | 6.91 (1.63) | 6.93 (3.09) | .512 | .01 |
| Midfoot surface contact (%) | 23.52 (6.60) | 21.68 (3.34) | .875 | .35 |
| Midfoot impulse (%) | 6.02 (2.06) | 5.53 (2.56) | .256 | .21 |

P = statistical significance from either independent samples *t*-test or Mann-Whitney *U*-test; *d* = effect size, indicated by Cohen's *d*. HM = medial heel; HL = lateral heel.

For each variable identified by univariate analysis as differing between TSF cases and controls with an alpha level below 0.10, plots were produced highlighting the difference between the value for each TSF case and the control group mean (Figure 5.8). Visual inspection of Figure 5.8 gives confidence that the values obtained for TSF cases are different from the control group means. For each of the six variables, individual case responses were outside the 95% confidence intervals (95%CI) for the control group mean at least 80% of the time. The results for peak pressure at the medial heel appear the least consistent, with 40% of TSF cases not falling above the upper 95% CI limit.

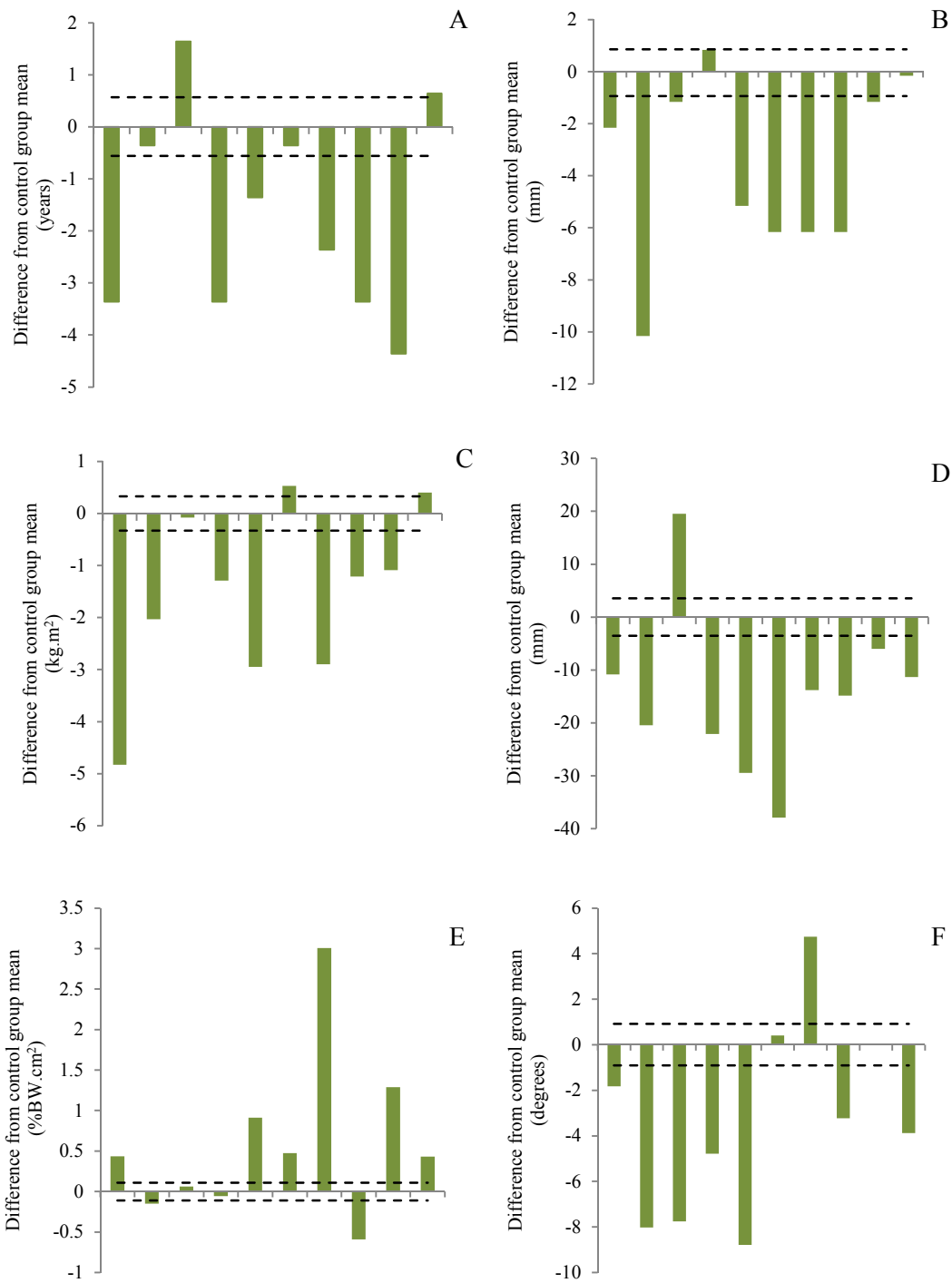


Figure 5.8. Exploration of risk factors for TSF. The difference between the individual case mean and the control group mean is displayed for each variable. A=age; B=bimalleolar width; C=BMI; D=corrected calf girth; E=peak pressure at the medial heel; F=tibial rotation range of motion. Dashed lines represent the upper and lower 95% confidence intervals for the control group mean.

5.3.2. MT3SF v Controls

Of the anthropometric (Table 5.4), kinematic (Table 5.5) and plantar pressure (Table 5.6) variables assessed between groups, results indicated that thirteen variables were different between the MT3SF and control groups ($P < 0.05$). Compared with controls, recruits incurring a MT3SF were younger; had a lower BMI; struck the ground with a plantarflexed, rather than dorsiflexed ankle; had greater pressures at the M3-M5 regions; had greater impulse at the M4 and M5 regions; had a later occurrence of peak pressure at the M1-M3 regions; and had a more abducted foot during ground contact ($P < 0.05$). In addition to these variables, exploratory plots were produced for four variables which produced a P of < 0.10 : passive ankle dorsiflexion; time of peak dorsiflexion; time of peak pressure at the M4 and M5 regions (Figure 5.9).

Table 5.4. Mean (SD) for anthropometric variables investigated between the third metatarsal stress fracture (MT3SF) and control groups. The results of univariate analysis and effect sizes are presented for each variable. Statistically significant differences between groups are in bold.

| Anthropometric variable | MT3SF (n=14) | Controls (n=120) | P | d |
|-------------------------------|---------------------------|---------------------|--------------|------------|
| Age (years) | 19.86 (2.60) | 21.36 (3.12) | .040* | .52 |
| Bimalleolar width (mm) | 69.79 (5.46) | 72.16 (4.96) | .131 | .45 |
| BMI (kg.m²) | 23.40 (1.47) | 24.31 (1.82) | .018* | .55 |
| Corrected calf girth (mm) | 365.77 (22.43) | 367.52 (19.53) | .471 | .08 |
| External hip rotation (°) | 28.14 (6.77) | 25.41 (1.82) | .139 | .53 |
| Internal hip rotation (°) | 26.52 (7.14) | 25.92 (7.72) | .775 | .08 |
| Hip range of motion (°) | 49.52 (10.97) | 51.32 (12.88) | .618 | .15 |
| Height (m) | 1.78 (0.05) | 1.78 (0.05) | .507 | <.01 |
| Mass (kg) | 74.73 (6.98) | 76.67 (6.62) | .153 | .29 |
| Peak ankle dorsiflexion (°) | 28.53 (6.43) ⁺ | 31.12 (5.38) | .061 | .44 |

P = statistical significance from either independent samples t -test or Mann-Whitney U -test; d = effect size, indicated by Cohen's d ; ⁺ n=12.

Table 5.5. The mean (SD) of ten barefoot running trials for kinematic variables investigated between the third metatarsal stress fracture (MT3SF) and control groups. The results of univariate analysis and effect sizes are presented for each variable. Statistically significant differences between groups are in bold.

| Kinematic variable | MT3SF (n=12) | Controls (n=120) | <i>P</i> | <i>d</i> |
|---|--------------------|---------------------|--------------|------------|
| Foot angle at touchdown (°) | 6.96 (3.06) | 7.79 (3.12) | .395 | .27 |
| Ankle angle at touchdown (°) | 0.75 (2.03) | -0.77 (1.97) | .024* | .79 |
| Peak dorsiflexion angle (°) | -11.46 (3.42) | -11.85 (3.15) | .346 | .12 |
| Ankle angle range of motion (°) | 12.21 (4.41) | 12.15 (3.19) | .307 | .01 |
| Time of peak dorsiflexion (%) | 54.40 (7.34) | 51.59 (4.63) | .077 | .46 |
| Knee angle at touchdown (°) | -12.74 (4.86) | -11.73 (5.01) | .641 | .20 |
| Peak knee flexion angle (°) | -28.76 (5.00) | -27.24 (5.59) | .440 | .29 |
| Knee angle range of motion (°) | 15.70 (5.06) | 15.52 (3.54) | .909 | .04 |
| Time of peak knee flexion (%) | 40.13 (11.30) | 38.93 (5.54) | .767 | .13 |
| Rate of knee flexion (°/sec ⁻¹) | 159.88 (49.89) | 163.28 (36.02) | .999 | .08 |
| Rearfoot angle at touchdown (°) | 1.69 (1.57) | 2.14 (1.84) | .509 | .26 |
| Peak eversion angle (°) | -6.24 (5.80) | -5.62 (2.60) | .408 | .14 |
| Rearfoot range of motion (°) | 8.30 (6.40) | 7.85 (3.65) | .703 | .09 |
| Time of peak eversion (%) | 38.56 (14.95) | 41.60 (7.61) | .260 | .26 |
| Rate of eversion (°/sec ⁻¹) | 64.79 (37.18) | 77.19 (35.39) | .359 | .34 |
| Tibial range of motion (°) | 10.97 (4.43) | 9.72 (5.06) | .278 | .26 |

P = statistical significance from either independent samples *t*-test or Mann-Whitney *U*-test; *d* = effect size, indicated by Cohen's *d*.

Table 5.6. The mean (SD) of ten barefoot running trials for plantar pressure variables investigated between the third metatarsal stress fracture (MT3SF) and Control groups. The results of univariate analysis and effect sizes are presented for each variable. Statistically significant differences between groups are in bold. Peak pressure and impulse were expressed as a percentage of body weight per square centimetre. Timings were expressed as a percentage of total stance time. Midfoot surface contact and impulse were expressed as a percentage of whole foot contact and impulse respectively.

| Plantar pressure variable | MT3SF (n=14) | Controls (n=120) | <i>P</i> | <i>d</i> |
|---|---------------------|---------------------|--------------|------------|
| M1 peak pressure (%BW.cm ⁻²) | 1.65 (0.61) | 1.64 (0.60) | .440 | .02 |
| M2 peak pressure (%BW.cm ⁻²) | 2.83 (0.99) | 2.60 (0.82) | .311 | .25 |
| M3 peak pressure (%BW.cm⁻²) | 3.36 (1.06) | 2.82 (0.84) | .027* | .56 |
| M4 peak pressure (%BW.cm⁻²) | 3.09 (0.98) | 2.48 (0.76) | .009* | .70 |
| M5 peak pressure (%BW.cm⁻²) | 1.76 (0.46) | 1.50 (0.68) | .036* | .45 |
| HM peak pressure (%BW.cm ⁻²) | 2.37 (0.70) | 2.34 (0.62) | .362 | .05 |
| HL peak pressure (%BW.cm ⁻²) | 2.41 (0.92) | 2.39 (0.75) | .405 | .02 |
| M1 impulse (%BW.s) | 3.44 (1.91) | 3.17 (1.37) | .364 | .16 |
| M2 impulse (%BW.s) | 5.02 (2.60) | 4.92 (1.67) | .236 | .05 |
| M3 impulse (%BW.s) | 4.95 (1.77) | 4.62 (1.39) | .369 | .21 |
| M4 impulse (%BW.s) | 4.38 (1.38) | 3.73 (1.23) | .043* | .50 |
| M5 impulse (%BW.s) | 2.65 (0.91) | 2.21 (1.18) | .035* | .42 |
| HM impulse (%BW.s) | 4.10 (2.47) | 3.56 (1.43) | .451 | .27 |
| HL impulse (%BW.s) | 2.73 (1.68) | 2.32 (1.03) | .208 | .29 |
| M1 time of peak pressure (%) | 58.37 (8.44) | 53.90 (5.94) | .037* | .61 |
| M2 time of peak pressure (%) | 60.40 (5.75) | 56.65 (4.79) | .004* | .71 |
| M3 time of peak pressure (%) | 57.75 (3.88) | 54.70 (5.01) | .015* | .68 |
| M4 time of peak pressure (%) | 52.02 (4.16) | 49.88 (5.50) | .081 | .44 |
| M5 time of peak pressure (%) | 48.68 (5.77) | 45.98 (7.53) | .075 | .50 |
| HM time of peak pressure (%) | 10.42 (5.02) | 10.16 (5.84) | .711 | .05 |
| HL time of peak pressure (%) | 7.23 (2.05) | 6.93 (3.09) | .344 | .11 |
| M1 time of first contact (%) | 11.87 (6.34) | 12.55 (4.80) | .313 | .12 |
| M2 time of first contact (%) | 9.32 (5.13) | 10.36 (4.33) | .199 | .22 |
| M3 time of first contact (%) | 6.85 (4.06) | 7.59 (3.88) | .221 | .19 |
| M4 time of first contact (%) | 5.33 (3.87) | 5.58 (3.12) | .216 | .07 |
| M5 time of first contact (%) | 5.53 (4.18) | 5.84 (3.33) | .205 | .08 |
| Foot axis angle (°) | 13.09 (7.50) | 9.23 (6.21) | .034* | .56 |
| Midfoot surface contact (%) | 21.84 (3.40) | 21.68 (3.34) | .320 | .05 |
| Midfoot impulse (%) | 5.00 (1.33) | 5.53 (2.56) | .304 | .26 |
| Time of heel off (%) | 53.63 (7.67) | 49.45 (5.52) | .034* | .63 |

P = statistical significance from either independent samples *t*-test or Mann-Whitney *U*-test; *d* = effect size, indicated by Cohen's *d*.

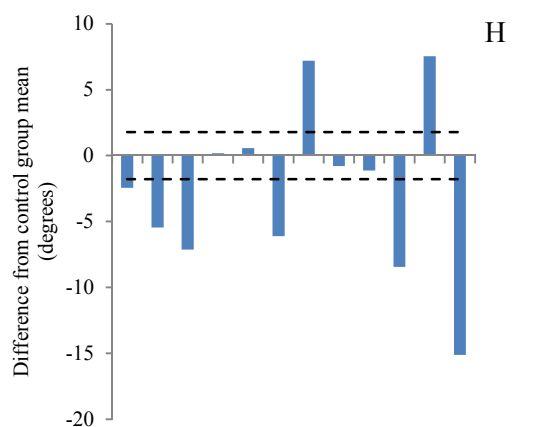
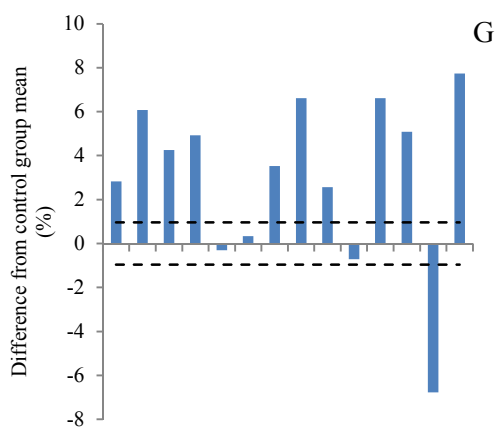
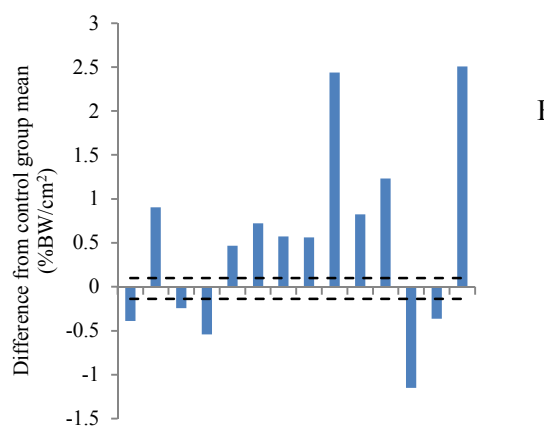
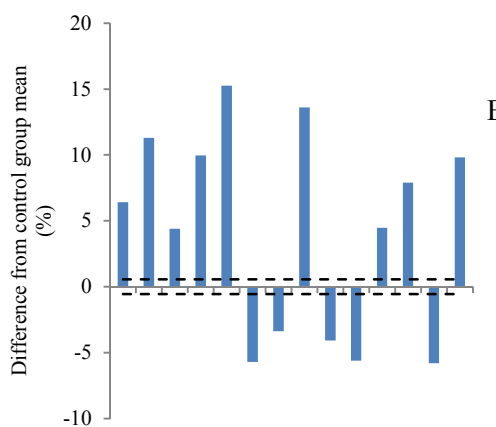
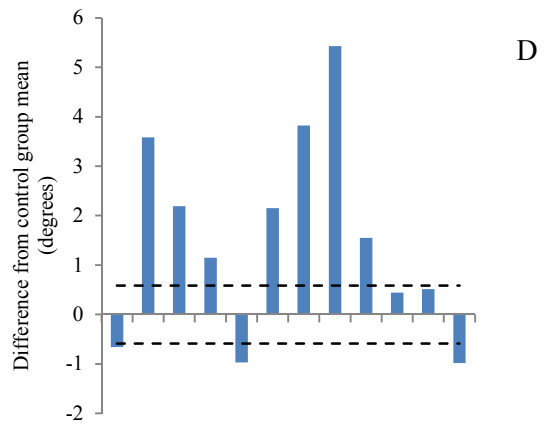
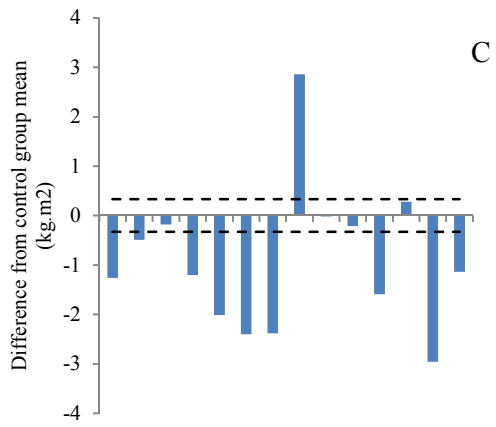
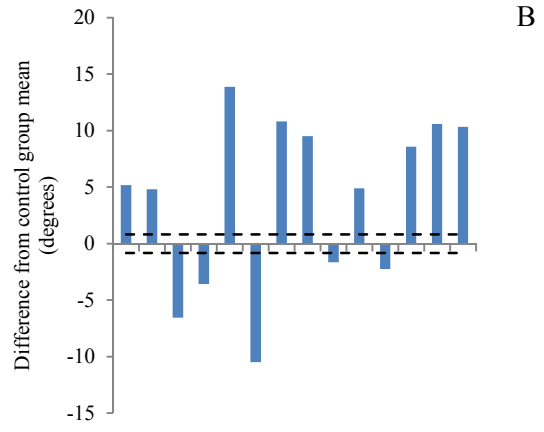
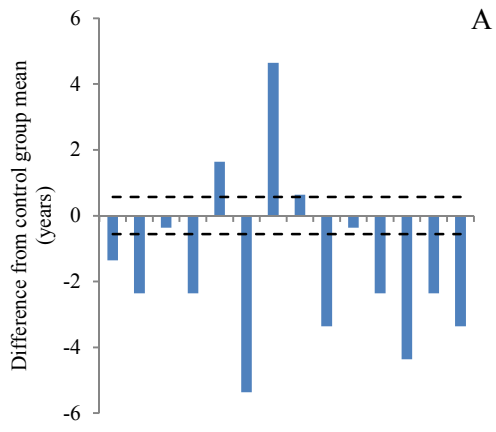


Figure 5.9. Exploration of risk factors for MT3SF. The difference between the individual case mean and the control group mean is displayed for each variable. A = Age; B = Ankle angle at touchdown (n=12); C= BMI; D = Foot axis angle; E = Heel off time; F= Peak pressure; G = Time of peak pressure; H = Passive dorsiflexion (n=12). Dashed lines represent the upper and lower 95% confidence intervals for the control group mean.

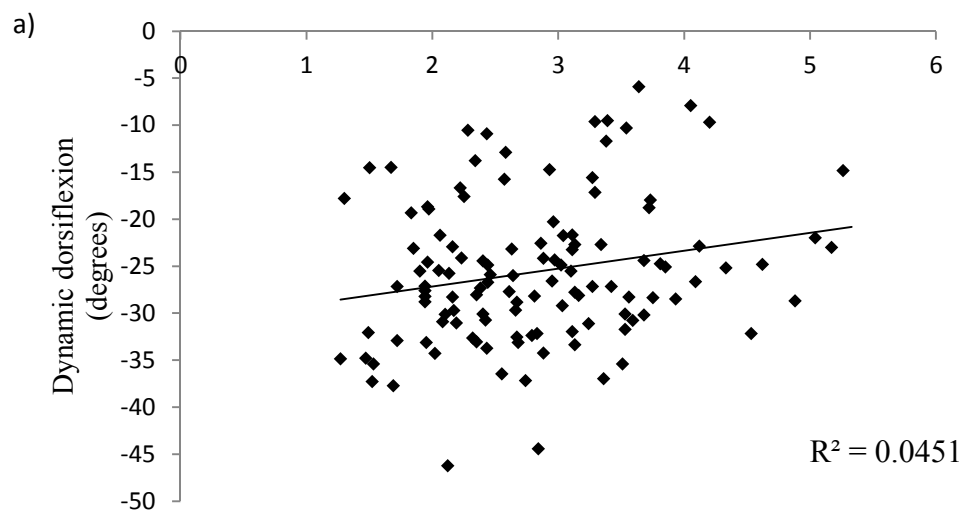
The results of correlation analysis are summarised in Table 5.7, with significant and hypothesis-related correlations presented. Passive ankle dorsiflexion did not correlate with dynamic ankle variables, time of heel off or pressure variables at the M3 region ($P>0.05$). Peak pressure at the M3 region was weakly correlated with the amount of dynamic dorsiflexion (Figure 5.10a) and the timing of peak eversion (Figure 5.10b). Peak pressure and impulse were highly correlated, however neither were correlated with the timing of peak pressure ($P>.05$). Ankle angle at touchdown (Figure 5.11a), and timing of peak dorsiflexion (Figure 5.11b) were weakly but significantly correlated with the timing of peak pressure while the timing of heel off was more strongly linked with this event (Figure 5.11c). Significant associations were also seen between ankle angle at touchdown and dynamic dorsiflexion (Figure 5.12a) time of heel off and time of peak eversion (Figure 5.12b) and rearfoot range of motion and passive dorsiflexion (Figure 5.12c).

Table 5.7. Summary of analysis of correlations between variables associated with MT3SF risk. The correlation strength is shown by the square of Pearson's r (R^2), and the statistical significance of the correlation included in the final column (P).

| | Variables | R^2 | P |
|----------------------|---------------------------|-------|------|
| Passive dorsiflexion | Dynamic dorsiflexion | -.008 | .370 |
| | Time of peak dorsiflexion | .000 | .999 |
| | M3 peak pressure | -.017 | .160 |
| | M3 time of peak pressure | -.000 | .978 |

| | | | |
|--------------------------|----------------------------------|---------------|-----------------|
| | M3 impulse | -.007 | 370 |
| | Time of heel off | -.017 | .152 |
| | Rearfoot range of motion | .049* | .024 |
| Dynamic dorsiflexion | Ankle angle at touchdown | .059* | .007 |
| | M3 peak pressure | .045* | .020 |
| | M3 time of peak pressure | .002 | .599 |
| | M3 impulse | .005 | .439 |
| | Time of heel off | .001 | .705 |
| Ankle angle at touchdown | M3 time of peak pressure | -.040* | .028 |
| M3 peak pressure | M3 time of peak pressure | .010 | .253 |
| | M3 impulse | .762* | <.001 |
| | Time of heel off | -.000 | .850 |
| | Time of peak eversion | .041* | .027 |
| Time of heel off | M3 time of peak pressure | .393* | <.001 |
| | Time of peak eversion | .052* | .012 |
| M3 time of peak pressure | Time of peak dorsiflexion | .058* | .009 |

M3 = Third metatarsal head region



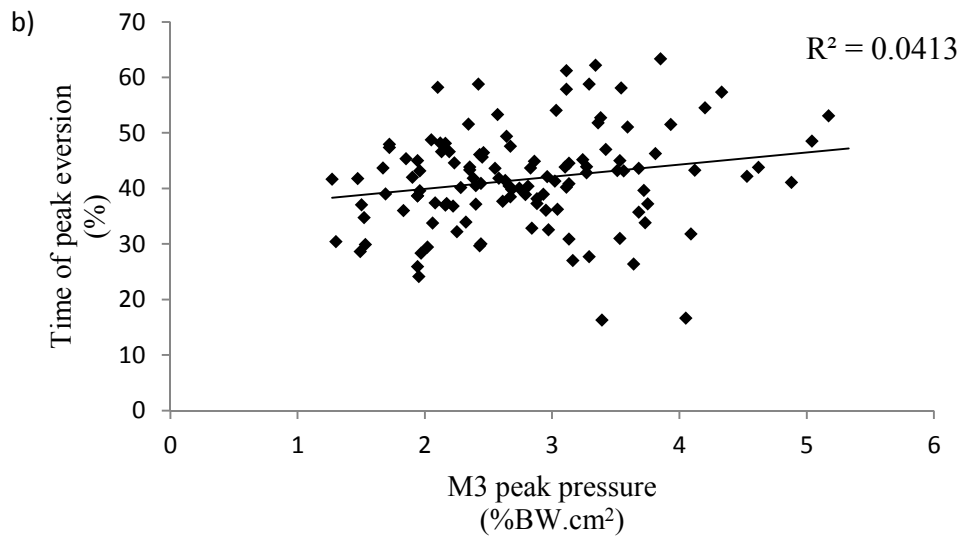
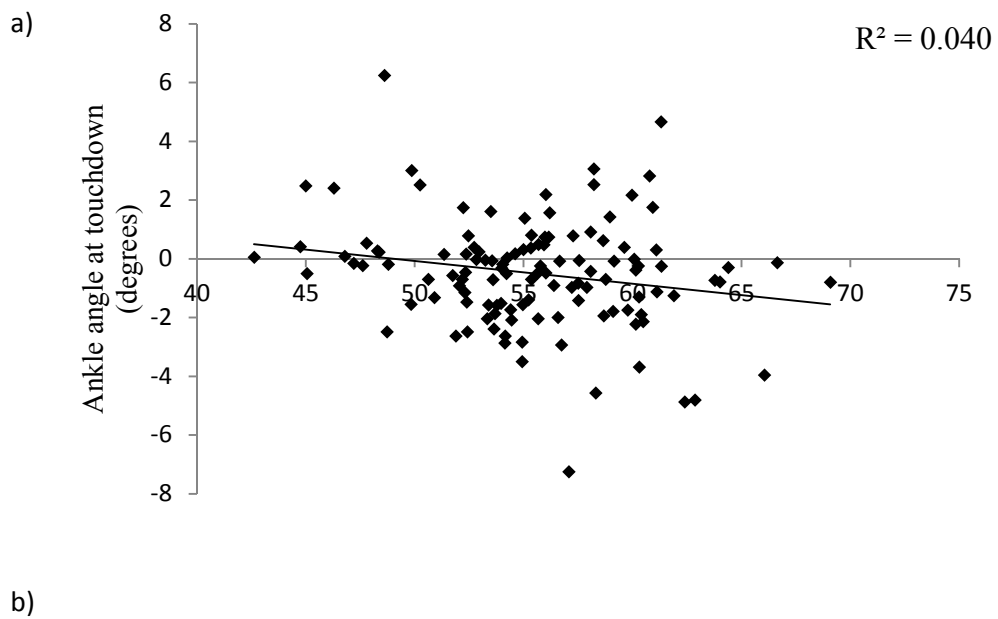


Figure 5.10. Correlations between peak pressure at the M3 region and: a) dynamic dorsiflexion; b) time of peak eversion.



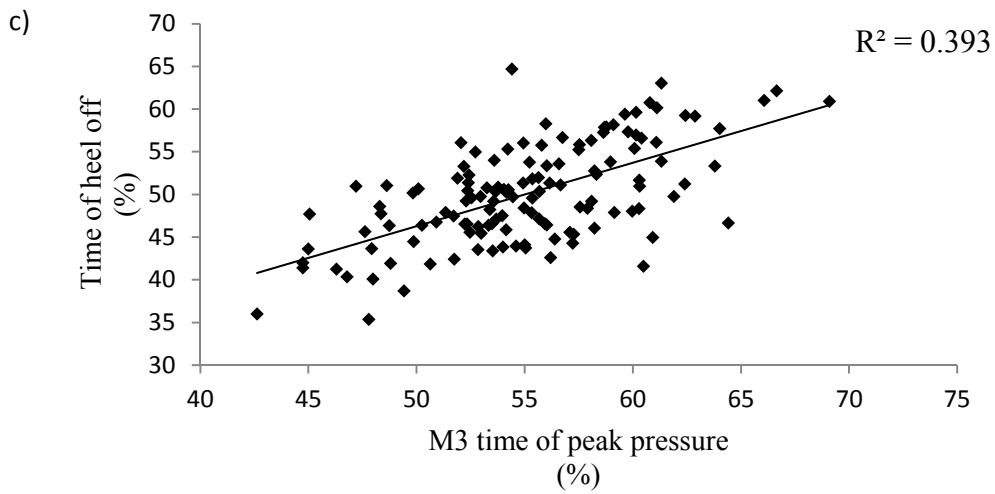
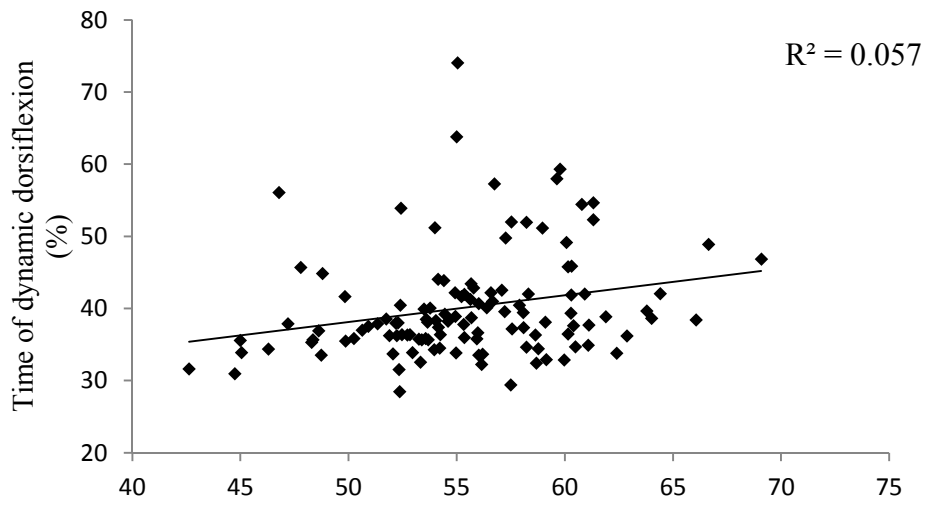
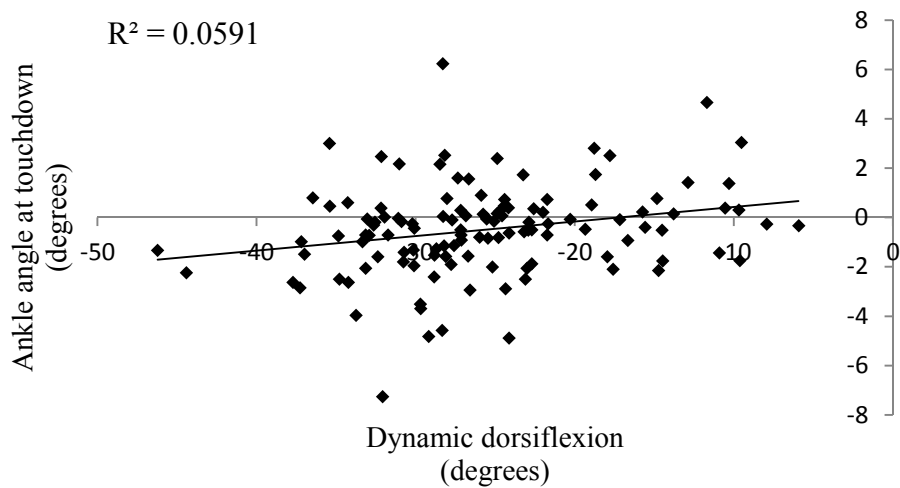


Figure 5.11. Correlations between time of peak pressure at the M3 region and: a) ankle angle at touchdown; b) time of peak dorsiflexion; c) time of heel off.



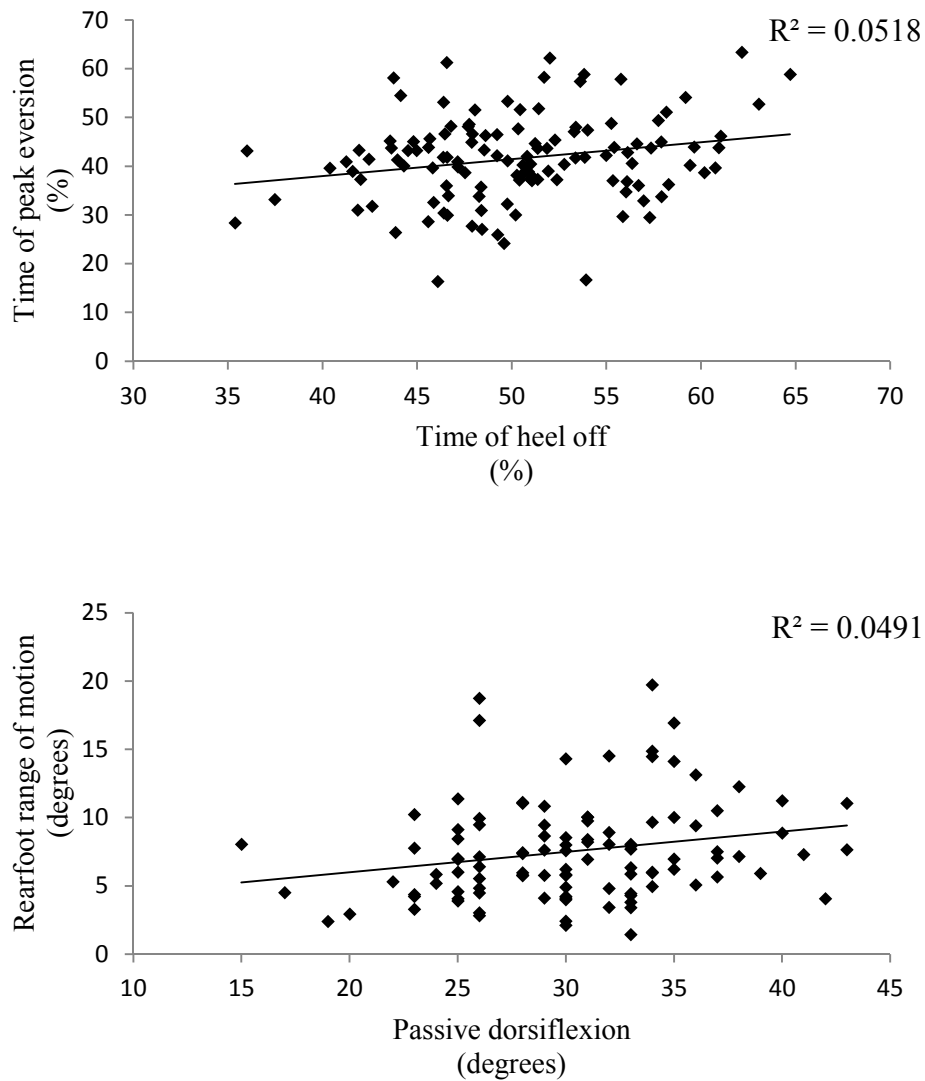


Figure 5.12. Correlations between: a) ankle angle at touchdown and dynamic dorsiflexion; b) time of peak eversion and time of heel off; c) rearfoot range of motion and passive dorsiflexion.

5.4. Discussion

This study sought to identify variables associated with TSF and MT3SF in RM recruits through a prospective cohort design and was the first of this design investigating gait variables in a military population. There were a number of statistically significant differences between stress fracture groups and controls,

providing insight into variables that may be relevant to the development of specific stress fractures. Additionally, further information on the influence of foot function on MT3SF risk was provided through analysis of correlations between variables proposed to relate to this injury. After an initial discussion of injury incidence and distribution, the following section will consider the results related to each injury in turn.

Fifteen MT3SF and 10 TSF were sustained amongst the 1065 RM recruits tested at the start of training, representing injury rates of 0.9% and 1.4% respectively. In Ross & Allsopp's (2002) report of 2091 RM recruits, a MT3SF rate of 1.7% and TSF rate of 1.4% was observed. The incidence rates reported here are therefore slightly lower than 11 years ago, however it should be noted that this is specific to this study, rather than changes in RM training injury rates. A recent study by House et al. (2013) monitored injury rates in 2004-2005, during which time a lower limb stress fracture rate of 4.4% was observed, a figure greater than in 2002. Evidence for those recruits who were not tested but passed through training during the start of the study period suggests that a greater proportion of these recruits sustained injury compared to the study recruits, potentially explaining the low numbers available for analysis. For example, a full examination of the injury records for the first 10 troops included in the study indicated that there were five TSF cases amongst the recruits that did not attend testing during that time, a rate of 2.3%. The reasons for not attending testing may be various, therefore it is unclear whether those not volunteering for the study had characteristics which placed them at high risk of

injury, however it does indicate why reported injury numbers amongst those involved in the study were slightly lower than previous reports.

Figure 5.1 shows the distribution of TSF and MT3SF with regard to the week of training in which they occurred. No injuries occurred before week 9 of training, but there were clusters of MT3SF in two main periods – weeks 11-15 (5 cases) and 24-28 (7 cases). The distribution of the 9 TSF cases for which data was available was more even, although 5 cases occurred in the middle third of training (weeks 11-21). Finestone et al. (2011) reported stress fracture distribution across a two-stage, 52 week military training programme, reporting that the large majority of stress fractures occurred at the tibia and femur prior to 26 weeks, with over 90% of stress fractures occurring to the metatarsals in the second half of training. No such pattern is in evidence here. Given the small sample size, interpretation of this distribution is limited, however the clustering of MT3SF cases in particular may be of interest. A recent report by Rice, Davey, Dixon & Fallowfield (2012) performed such an analysis and identified the rate of progression of training (e.g. sharp increases in the volume or intensity of training) as contributing to the ‘clustering’ of stress fractures at certain weeks of training. For example, 24% of TSF occurred in week 31 of training, which was the third highest week of training volume, and was preceded by the highest and second highest weeks of volume. Review of training during these weeks of high injury incidence and immediately prior to them may provide insight into loading mechanisms, or identify the need for scheduled rest periods.

5.4.1. Tibial stress fractures

A review of current literature suggested that there may be a number of factors which could feature in an increased risk of TSF. In the ten cases in the present population, differences identified for variables are consistent with previous suggestions for this injury, however some variables hypothesised to differ between groups did not. The results for lower age, bimalleolar width, BMI and corrected calf girth in TSF cases may be interpreted in relation to research suggesting the importance of tibial geometry and muscular insufficiency. The findings for tibial rotation and peak heel pressure are suggested to relate to poor impact attenuation.

5.4.1.1. Anthropometric variables

Results suggest that youth may be considered as a risk factor for TSF in military recruits, as has been demonstrated previously (Milgrom et al., 1994), supporting the hypothesis. Growing bone is particularly susceptible to damage because it is not fully formed and presents growth plates (Adirim & Cheng, 2003), therefore with the minimum age requirement for entry into RM training set at 16, there may be a number of recruits with bones that are still developing. While youth may contribute to the likelihood of sustaining a TSF on its own, this risk will likely increase when combined with other factors. If the mechanism linking youth with stress fracture risk is through low bone strength, then the addition of high external loads and/or insufficient impact attenuation mechanisms may be required in combination before a TSF develops.

BMI was lower in TSF cases compared to controls ($P<0.05$), supporting the hypothesis. Ode, Pivarnik, Reeves & Knous (2007) suggest that BMI is not an accurate predictor of fatness in athletes, therefore, given the physical fitness requirements for entry into training, it can be speculated that greater BMI in this population is associated with increased lean mass, rather than fat. Higher BMI (Felson, Zhang, Hannan et al., 1993; McGuigan, Murray, Gallagher et al., 2002) is associated with increased BMD, which is a determinant of bone strength. More specifically, the lean muscle mass component of BMI has been identified as the most important determinant of BMD (Liu, Zhao, Ning et al., 2004), with lower BMD linked directly with more rapid microcrack propagation (Carter et al., 1981) and frequently cited as a risk factor for stress fracture (see section 2.4.1.3). Bennell et al. (1996) found lower calf muscle mass to correlate with lower tibial BMD, further supporting this link. In addition to the correlation between muscle mass and BMD, calf musculature has been demonstrated to play an important role in applying loads to reduce bending moments acting on the tibia (Scott & Winter, 1990). The observation of lower calf muscle mass in the TSF group may indicate an inability to adequately perform this task, compared to controls. Results relating to calf girth corroborate the findings of Milgrom (1989) in male military recruits, but partially contradict Bennell et al. (1996) who found a similar effect in female, but not male runners. It is proposed that the data for BMI and calf girth are indicative of lower bone strength, and that the presence of lower calf girth may also relate to a reduced ability to counteract bending loads acting on the tibia.

It was found that bimalleolar width, another indicator of bone strength, was lower in recruits sustaining a TSF, supporting the hypothesis. This result was associated with a large effect size and is suggested to be an indication that these recruits had narrower tibiae than their uninjured counterparts. Beck et al. (2000) reported that male military recruits sustaining TSF had narrower tibiae. Bimalleolar width has been used in anthropometry to indicate frame size (Himes & Bouchard, 1985), and specifically as a component of tibial geometry (Franklyn et al., 2008). Franklyn et al. (2008) measured bimalleolar width with radiographs and found it to be clinically and statistically significantly smaller in subjects with medial tibial stress syndrome compared to athletic controls. There was also a non-significant trend for this width to be smaller in female subjects with a history of TSF than athletic controls. Given that this variable was significantly lower in TSF cases in the present study, it is interpreted that these recruits had narrower tibiae than controls, suggesting a lower area moment of inertia, which has been previously identified as a risk factor for TSF (Milgrom et al., 1989).

Additional measurements of the lower limb would provide greater detail regarding how the geometry of the tibia differs in those recruits at risk of TSF. Measurements of the width of the tibial plateau and the length of the tibia could be included, for example. It should be noted that without scanning techniques, these are surrogate estimates of tibial geometry. Mean values for bimalleolar width in the present study were 68 – 72 mm, compared to 53 – 54 mm in the study by Franklyn et al. (2008). This is due to the direct estimation of bone parameters from radiographs in the previous paper, compared to the current calliper measurement that included skin, subcutaneous fat and, most

influentially, the fibula. Regardless of these limitations, data suggest that this simple and quick method could be a useful tool in predicting recruits at elevated risk of TSF.

The passive range of motion of the hip was hypothesised to differ between TSF cases and controls, however no difference was reported either in internal, external or relative internal-external range of motion, rejecting the hypothesis. These results contrast with findings for Israeli military recruits by Giladi et al. (1987b; 1991), although evidence taken from wider populations reports similar null findings to the present study (Montgomery et al., 1989; Benell et al, 1996; Lun et al., 2004). It must be noted that the mean values obtained here for RM recruits are lower than those previously reported (see Table 2.3). Differences between the observed values may be due to RM recruits being less flexible than recreational runners, as well as the subjectivity involved in the testers' interpretations of 'firm resistance' in each study, however the same tester assessed all recruits in the current study, reducing the possibility of random error. Aside from the studies conducted with the Israeli defence force (Giladi et al., 1987b; Giladi et al., 1991), there is no clear relationship between passive hip range of motion and risk of TSF, largely due to an absence of evidence linking this passive measure to dynamic function. Passive hip range of motion is commonly assessed in research examining stress fracture aetiology, but results from this study suggest that it is not associated with the risk of TSF in RM recruits.

5.4.1.1. Gait variables

An inverse relationship was observed between the risk of TSF and the range of tibial rotation relative to the foot available to recruits, supporting the hypothesis. Similar peak eversion and rearfoot range of motion values between groups suggest that axial rotation of the shank accounted for this finding. The relative transverse plane movement between shank and foot has been suggested to decrease with reduced medial longitudinal arch height (Nigg et al., 1993), however there was no difference in dynamic midfoot contact between groups ($P>0.05$). Previous associations had been drawn based on static assessments of arch height, which are only weakly associated with dynamic measurements of arch function (Cavanagh et al., 1993; Kaufman et al., 1999), therefore data presented here suggest no relationship exists between dynamic midfoot contact and relative tibial rotation. As all other kinematic and plantar pressure variables were similar between groups, it is suggested that the decreased impact attenuation provided by lower tibial rotation range of motion resulted in the greater medial peak pressure seen in TSF cases ($P<0.05$). Peak lateral heel pressure was also greater in the TSF group compared to controls, and with a greater differential than seen at the medial heel, but although a moderate effect size ($d = 0.59$) was observed, this difference was not statistically significant ($P>0.05$). There is some evidence associating reduced lower limb range of motion with increased risk of stress fractures through decreased impact attenuation (e.g. a rigid, high-arched foot in Williams et al., 2001), but evidence regarding tibial rotation and impact attenuation is absent from the literature and thus requires investigation.

It was hypothesised that ankle kinematics would vary between groups, due to their potential to influence impact attenuation, but this hypothesis was rejected. Evidence linking plantar pressure data to dynamic lower limb function is limited, but does suggest associations between greater medial plantar pressures and increased rearfoot eversion (Rosenbaum et al., 1994; Cornwall & McPoil, 2003; Dixon & McNally, 2008), however rearfoot motion was similar between groups despite greater medial heel pressure in the present study. The mechanisms at the ankle (Gerritsen et al., 1995; Dixon et al., 2005) and knee (McMahon et al., 1987) typically associated with impact attenuation did not differ between groups, therefore while the relative rotation of foot and shank (greater rotation providing greater attenuation) has not been previously identified as a key aspect of impact attenuation, aspects of RM training may elevate the significance of its role.

5.4.2. Third metatarsal stress fractures

The relatively high frequency of MT3SF among recruits undergoing initial training attracted attention to this under-researched overuse injury, and this prospective gait analysis study represents a significant advancement in knowledge pertaining to the aetiology of this injury. Fourteen cases of MT3SF were analysed and compared to a control group, with significant differences in anthropometric, kinematic and plantar pressure variables identified between groups. Due to the limited prior knowledge regarding this variable, particularly regarding mechanisms suggested to increase metatarsal loading, analysis of correlations between lower limb movements and pressure variables at the M3 was also undertaken.

5.4.2.1. Anthropometric variables

Recruits who sustained a MT3SF were younger and had a lower BMI than the control group, supporting the hypotheses. Although evidence linking age, BMI or lean muscle mass to metatarsal bone strength is absent from the literature, it should be assumed that a consistent relationship exists for all load-bearing long bones as is suggested for the tibia. Fatigue has been associated with increased peak pressures on the metatarsal heads (Arndt et al., 2002; Nagel et al., 2008) therefore earlier fatigue due to weaker muscles may be a mechanism for increased loading during training. This draws a speculative link between BMI in RM recruits and fatigue however, and further evidence is required to examine whether fatigue during load carriage increases metatarsal loading.

In light of previous research, it was hypothesised that a lower medial longitudinal arch would be expected to present in the MT3SF group, however no such difference was apparent and the hypothesis was rejected. Various visual or manual assessments of arch index have provided evidence of no effect of arch height on injury (Montgomery et al., 1989; Ekenman et al., 1996; Kaufman et al., 1999; Esterman & Pilotto, 2005). These studies relied on static estimates of arch height to indicate injury risk in dynamic situations, an approach which may be questionable when considering the poor correlations reported between static and dynamic descriptors of the medial longitudinal arch (Cavanagh et al., 1997; Kaufmann et al., 1999). The method of quantification

used in the present study has been demonstrated by a number of authors when classifying dynamic foot type (Cavanagh et al., 1997; De Cock et al., 2006; Arther et al., 2007), and should be considered appropriate as it described dynamic foot posture, however it is not a direct indication of arch height above the ground, rather the portion of the midfoot which contacts during stance. These methods could be improved by adding visual tracking of arch height throughout stance.

5.4.2.2. Gait variables

At the MT3 head region, the finding of greater peak pressures which occurred later in the MT3SF group, and similar impulse between groups led to partial acceptance of the hypothesis. Although the magnitude of peak pressure beneath the metatarsal heads has been proposed to be associated with stress fracture development (Arndt et al., 2002; Weist et al., 2004; Nagel et al., 2008), this is the first study to present prospective evidence of greater peak pressures in those sustaining MT3SF. Mechanical theory suggests that, for a metatarsal of given geometry, increasing the vertical load applied at the end of the bone will increase mid-shaft bending stress, which will increase the likelihood of accumulation of microdamage. Data from Study 2 (Chapter 4) indicate that each 1 N increase in plantar force results in a 0.60 MPa increase in mid-shaft compressive stress, therefore it could be inferred that for the observed difference in peak MT3 pressure of 0.54% BW (mean bodyweight = 750 N, equating to a real difference of approximately 4 N/cm²), estimated peak compressive stress at the third metatarsal mid-section was 1.60 MPa greater in

MT3SF cases than controls during each step. It was expected that impulse would be greater beneath the MT3 region, however the similarity for this variable between groups suggests that loading throughout stance only differs at the time of peak magnitude. Rather than being related to the force-time integral, results suggest that mechanisms causing an increase in the magnitude, or delay in the timing of peak pressure at the MT3 region are dominant in MT3SF development.

Peak pressures were consistently greater across all metatarsal head regions in the MT3SF group, with this difference being significant at the MT3, MT4 and MT5 head regions. As pressure is equal to force over an area, it would be expected that a rise in pressure at one area would lead to a decrease in another, assuming no change in overall force, however there was no associated decrease in peak pressure or impulse at any other assessed region. Previous research has indicated a shift in pressure from the toes to the metatarsal heads following a marathon, with proposed implications for metatarsal stress fracture risk (Nagel et al., 2008). It is possible in the present study that the toes of the control group contributed more to the distribution of plantar forces than those of the MT3SF cases, however data are required to support this. It is also interesting to note that both groups experienced the highest peak pressure at the MT3 head region, with impulse at the MT2 head slightly greater or very similar to that at the MT3 head. The only large-scale data set providing typical pressures for barefoot running was collected by De Cock and colleagues and presented over a series of papers. These authors identify the MT2 as the site of highest peak pressure during barefoot running (De Cock et al., 2006),

suggesting that differences in methodology or population may influence the relative loads observed on the metatarsal heads. Given the availability of data for up to 1065 recruits, further investigation should be performed in the future to provide reference data for loading patterns in this cohort.

The foot was found on average to be nearly 4° more abducted ($P < 0.05$) in the MT3SF group, with an associated medium effect size, supporting the hypothesis. Research conducted with people with diabetes mellitus and peripheral neuropathy found that these individuals walked with greater foot abduction (or 'toe out') during stance, which contributed to greater peak pressures at the metatarsal heads (Mueller, Hastings, Commean et al., 2003). Interestingly these authors reported that this foot position increased pressures at the MT2, MT4 and MT5 head regions, rather than the MT3, and there were no correlations between foot axis angle and plantar loading in the present data, therefore any association between foot abduction angle and M3 loading remains unclear. This foot position may have resulted in the centre of pressure pathway travelling across the MT heads in a more lateral direction relative to the longitudinal axis of the foot. Further analysis of pressure data would provide information on the location of the foot centre of pressure and the velocity with which it travels, particularly during the push-off phase of gait. Additionally, analysis of the whole-body centre of mass location relative to the MT3 head would provide an indication of moments tending to cause axial rotation of the foot, which may cause increased lateral stress to be applied to the bone. With horizontal loads proposed to be of importance in MT3SF aetiology (Arangio et al., 1998; Dixon et al., 2006) this is an area that requires further investigation.

Although there was no significant difference in either dynamic or passive dorsiflexion between groups, there was a moderate effect size and near-statistical significance associated with the result for passive dorsiflexion, with MT3SF cases demonstrating a lower mean value ($P = .061$). Further inspection of data for this variable reveals that six of the twelve MT3SF cases had lower passive dorsiflexion than the control group, to the extent that their mean value was outside the 95% CI lower limit (Figure 5.9 (H)). The distribution of cases suggests that a weak relationship exists between passive dorsiflexion and MT3SF risk, and a type 2 error has been committed. The present results are similar to those of Dixon et al. (2006), who also found a lower but non-significant range of passive motion in recruits with a history of MT3SF compared to controls. The finding of no difference in the magnitude or timing of dynamic dorsiflexion between groups is also consistent with past findings (Dixon et al., 2006; Bischof et al., 2010). Hughes (1985) discussed the increased risk of metatarsal stress fracture in participants with 10 degrees of passive dorsiflexion or less (Hughes, 1985), however in the present study, both groups had around three times this amount of flexibility. It is therefore suggested that only highly restricted (e.g. equinus deformity) ankle flexibility may influence risk of forefoot pathology. In further support of this argument, there was no correlation between passive dorsiflexion and either the magnitude or timing of peak dynamic dorsiflexion, the timing of heel off or any pressure variables at the MT3 head region. Data corroborate the only comparable previous literature, which was performed on individuals with highly restricted dorsiflexion who were assessed during walking (Orendurff, 2006; Johansen et al., 2006; Turner et al., 2011) and found little or no association between passive

dorsiflexion and dynamic function. Given the non-significant difference between MT3SF cases and controls, and the lack of correlation between passive dorsiflexion and dynamic function, no support is provided in the current study for passive ankle dorsiflexion being a risk factor for MT3SF in RM recruits.

A strong argument is presented here for the association between the timing of events during stance and the risk of MT3SF. In contrast to expectations, the timing of both heel off and peak pressure at the MT3 head was later in the MT3SF group, with moderate to large effect sizes observed for both variables. Visual inspection of the timing of MT3 peak pressure in particular is convincing, with ten of the fourteen cases showing mean peak pressure timings that exceeded the upper 95% CI bound of the control group mean (Figure 5.9 (G)). There have been few studies to investigate the claims of Hughes (1985), who stated the expectation that restricted passive dorsiflexion would cause early heel-off, but what limited evidence there is contradicts this suggestion. For example Johanson et al. (2006) found that changes in passive dorsiflexion did not influence dynamic dorsiflexion or time to heel off during gait. While the moderate positive correlation between the timings of heel off and MT3 peak pressure suggests that the later heel off in this group was important in determining the occurrence of later peak pressures, it is not clear what caused the delayed heel off, or why later occurrence of peak pressures may be associated with injury. Despite the necessity for further research in this area, the evidence regarding the timings of heel-off and peak pressure could be used in future screening to identify recruits at elevated risk of MT3SF.

The only kinematic variable that differed between groups was the angle of the ankle at touchdown, which was slightly plantarflexed in the injury group, and slightly dorsiflexed in the control group. The presence of a plantarflexed ankle in the MT3SF group appears indicative of a more anterior foot position relative to the whole body centre of gravity at touchdown, particularly as the foot angle relative to the ground was similar between groups. This is likely to have occurred due to differences in hip flexion, as knee kinematics did not vary between groups, although data are not available to confirm this. With this more anterior leg position (Figure 5.13), it is suggested that the body rotated over the supporting leg later in stance, causing the delayed timing of heel off and peak pressures seen in this group. Figure 5.11 (a) does not provide support for this theory however, as it shows a very low correlation between the time of peak pressure at the MT3 head and the ankle angle at touchdown ($P < 0.05$). This pattern is observed amongst data taken from all participants therefore it is possible that injured recruits demonstrate a different function to controls. Given that there are no other kinematic variables which explain this difference in timings between groups, it is assumed that ankle position at ground contact partially influenced the time of heel off and peak MT3 pressure in the MT3SF group.

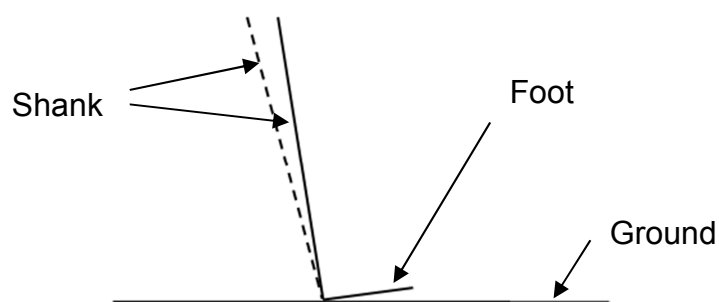


Figure 5.13. Illustrating the similar foot angle and differing shank angle at touchdown between the MT3SF (dashed line) and control (solid line) groups.

The similarity in midfoot contact area between injured recruits and controls is in agreement with some previous research (Montgomery et al., 1989; Ekenman et al., 1996; Kaufman et al., 1999; Esterman & Pilotto, 2005). The influence of footwear may be important in negating the influence of arch height, as the foot is supported relatively tightly and rigidly in the combat assault boot, especially when compared to modern running shoes, for example. It has been proposed that a pes planus foot promotes excessive pronation and increased loading of the foot structures, in addition to increased demand on muscles such as the anterior and posterior tibialis muscles to provide stability (DeLacerda, 1980). Combat boots, in addition to having a stiff sole with a defined arch, are likely to be tightly laced to prevent rubbing and blister formation, and possess a leather upper, all of which are likely to restrict subtalar joint and medial longitudinal arch movement within the boot. As suggested with the TSF group, the influence of military footwear may reduce impact attenuation mechanisms, and may also mask any relationship between arch height and MT3SF risk.

5.4.3. Limitations

This was the largest prospective study to be performed utilising gait analysis in military recruits. However only 10 eligible TSF and 14 MT3SF cases were available for analysis, providing an observed power in the range of 0.58 – 0.85 for significant results. The observed incidence was slightly lower than expected,

but reflects the inherent difficulty of performing such a study in a highly time-pressured environment. Seventy per cent of recruits enrolling between September 2010 and June 2012 were included in the study, and of those not tested, stress fracture incidence was greater than for those included in the study. As described above, the example of 5 TSF cases amongst the first 10 troops that were missed by the study had a significant impact on the numbers available for analysis. Recruits may have decided not to attend testing due to a number of reasons, including: being worried about completing their admin loads on time; fear that the detection of an injury they already carried would see them excluded; disinterest in the study. Testing was scheduled for a busy time in the second week of training, therefore to increase the chance of screening a larger proportion of recruits in future (and therefore capturing more injury cases), it is suggested that data collection be scheduled for a less time-pressured situation. However, given that many recruits voluntarily withdraw from training in the first week, and testing at a later juncture is undesirable because training adaptations may have started to occur, collection of data in the second week has associated benefits.

The statistical design employed was chosen in order to provide information on which variables are associated with individuals sustaining a stress fracture in the population of interest. Univariate analysis (independent t-tests and Mann-Whitney U tests) was the correct choice for this approach, given the robustness of these tests and use of additional exploratory plots with comparison to the control group 95% confidence intervals for further verification of findings. A strategy that could have been considered is logistic regression, which would

have incorporated an aspect of predictive modelling, an approach used in previous studies investigating risk factors for TSF (e.g. Milgrom et al., 1994; Bennell et al., 1996; Milner et al., 2006). This approach was rejected for several reasons. First, the aim of the study was to investigate which variables could be identified as being associated with injury, not predict the relative contribution of risk factors to injury risk, or to identify a model for screening out high risk recruits. One benefit of logistic regression modelling is that the equation derived from the analysis can be used to predict injury, which could then have been utilised in screening, however on a practical level any such information based on the results of 10 or 14 cases would require further validation before implementation in the field. Second, the low injury sample sizes would have resulted in weak predictions. Third, in the case of MT3SF variables, multicollinearity would have existed between a number of the variables included in the model (e.g. peak pressures and timing variables), requiring further corrections and reductions in the power of any results.

Although logistic regression was not implemented in the present study, with strong justification, future analysis investigating multiple injuries may be of benefit to CTCRM medical staff. For example, if there were consistent findings related to risk of several types of stress fractures (such as indicators of poor bone strength), greater value would be placed upon the predictive aspect of logistic regression analysis and at-risk recruits could be screened at the application stage.

Despite the novel approach demonstrated by this study, incorporating an efficient means of biomechanically assessing large numbers of RM recruits in a working environment, there were certain limitations to the methodological design. One such limitation is that a force plate was unavailable, which would have provided additional useful information on variables such as peak impact force and the rate of loading of this force, as well as horizontal forces and joint moments. Vertical ground reaction forces have been cited in TSF aetiology (please refer to section 2.4.3.1, p.88), albeit with equivocal findings (Zadpoor & Nikooyan, 2011), while horizontal ground reaction forces were demonstrated in Study 1 and in previous literature (Dixon et al., 2006) to be a potential risk factor for MT3SF development. With plantar pressure data collected at a frequency of 200 Hz in the present study, caution is advised when interpreting peak pressures associated with the high frequency loading at heel strike. Examination of Figure 5.8 (E) reduces confidence in the finding for peak medial heel pressure being greater in TSF cases than controls, and it is acknowledged that discussion of the heel-region pressure data is limited given the use of plantar pressure data to analyse impact characteristics, rather than ground reaction force data.

Although it was deemed impractical to include a podiatric assessment in the current protocol, previous research indicates that there may be some benefit to the inclusion of this approach. In addition to the quantification of static arch index, which would have provided an interesting comparison with dynamic midfoot data, measures of forefoot and rearfoot varus/valgus could have added to the study. Forefoot varus deformity, where the medial metatarsals are raised

relative to the lateral metatarsals, has been linked with lower limb stress fracture (Hughes, 1985; Korpelainen et al., 2001), however poor reliability in the measurement of this variable has been reported (Van Gheluwe, Kirby, Roosen & Phillips, 2002), in addition to null findings (Lun et al., 2004). A retrospective comparison of RM recruits with a history of MT3SF and matched controls also found no difference in forefoot varus/valgus angle (Dixon et al., 2006), so it is equally likely that this variable may not have added to the present data. As a debated risk factor however, future investigations with the resources to include podiatric assessment may benefit from doing so. Despite the absence of certain methodological aspects, this study provides strong arguments for the importance of variables that were included, and it is acknowledged in any injury study that it is not possible to cover every variable that could relate to injury risk.

5.4.4. Summary, implications and conclusions

A prospective study incorporating dynamic gait, anthropometry and flexibility measures was conducted in a military population for the first time. Despite a very large sample size, relatively few cases of TSF and MT3SF were available for analysis, compared to the potential numbers available with retrospective study designs, yet knowledge of the mechanisms for injury in this population has been improved. Relatively low age and BMI were present in injury cases, with potential effects on the strength of bone and with further implications for fatigue, given the athletic population and the rigorous demands of training. With regard to TSF risk, lower tibial width, reduced calf muscle mass and less effective impact attenuation mechanisms were implicated, emphasising the

inability of these recruits to resist the high loads involved in RM recruit training. MT3SF cases demonstrated different injury mechanisms, with the magnitude and timing of peak forefoot pressures being greater and later respectively than their injury-free counterparts. The influence of ankle position at touchdown was deemed important in influencing timings, with heel off also delayed in this group. The reason for greater peak pressures was unclear but potentially related to the foot being more abducted in the injury group, which has been shown to increase forefoot pressures, and may be of importance in determining the direction of loads applied to the forefoot. In addition to the variables associated with injury, the potential influence of footwear on injury mechanisms is acknowledged. The effect of a supportive CAB on lower limb function, particularly in relation to impact attenuation, is of interest and warrants further investigation.

The data presented provide scope for interventions to be implemented. The minimum age requirement for entry into RM recruit training is 16, with a minimum body mass requirement of 65 kg and BMI requirement 'within healthy range' which, according to the NHS calculator linked to on the information provided by the Royal Navy website (www.royalnavy.mod.uk/Careers/How-to-join/Eligibility) is 18.5 - 25 kg.m². Not only is this criterion range only suitable for the general population, rather than athletic individuals (Ode et al., 2007), but the control group and both stress fracture group means are at the upper end of this scale. It could be feasible to increase the minimum age and have a more suitable minimum BMI requirement, given the strong association of both these factors with both injuries explored herein.

The identification of high heel and forefoot loading in injury cases suggests that the issue of shock-absorbing insoles would be of benefit in reducing stress fracture incidence. There is already existing evidence of the effectiveness of this approach, with heel and forefoot pressures (Windle, Gregory & Dixon, 1999) as well as ground reaction forces (Dixon, 2007) being reduced. A recent study performed with RM recruits demonstrated that a shock-absorbing insole reduced lower limb overuse injury overall by 12.7% compared to the standard issue *Saran* insole, with significant reductions in TSF incidence in particular (House et al., 2013). This evidence is not unequivocal however, with reports that insoles can be ineffective (Gardner, Dziados, Jones et al., 1988), or may function to prevent other stress fractures such as femoral, rather than tibial or metatarsal (Milgrom, Giladi, Kashtan et al., 1985). The design of the insole is likely to be the important factor in its effectiveness, and there is evidence that custom orthotics prescribed by a barefoot plantar pressure assessment utilising the built-in 'D3D' *footscan* software could reduce impact forces and kinematic variables associated with injury (Dixon & McNally, 2008), and overall reduce the incidence of overuse lower limb injury in military training (Franklyn-Miller, Wilson, Bilzon & McCrory, 2011). This approach points to an individualised assessment strategy however, while the evidence in the present study suggests that a more time-efficient assessment could identify individuals who would benefit from a simple cushioning insole.

Key variables associated with each injury

TSF:

- Younger age
- Lower BMI
- Lower corrected calf girth
- Narrower bimalleolar width
- Lower range of tibial rotation
- Greater peak pressure at the medial heel

MT3SF:

- Younger age
- Lower BMI
- Greater peak pressure at the M3 region
- Later occurrence of peak pressure at the M3
- A more abducted foot during stance
- A later occurrence of heel off
- A more plantarflexed ankle position at touchdown

CHAPTER SIX: SUMMARY, IMPLICATIONS AND FUTURE DIRECTION

6.1. Summary of findings

The aim of this thesis was to investigate factors which may predispose RM recruits to TSF and MT3SF. These two stress fracture locations are the most common in this population, with MT3SF accounting for around 40% of all lower limb stress fractures (Ross & Allsopp, 2002). In Study 1, analysis of the footwear worn by RM recruits during training revealed that when wearing the CAB compared to the GT, plantar loading beneath the MT3 was greater and developed at a greater rate; ankle dorsiflexion was restricted; the forefoot was

subjected to more laterally applied loads and plantarflexor moments were greater, indicating increased muscular demand. These combined factors were suggested to increase the risk of MT3SF in this footwear condition. The development of a model of metatarsal bending stresses in Study 2 highlighted the contribution of individual metatarsal geometry to peak loads, as well as the role of plantar loading and the inclination of the foot. It also permitted the analysis of peak MT3 bending stresses in the standard issue footwear, which revealed significantly greater bending stresses in the CAB compared to the GT. The evidence provided by studies 1 and 2 suggests that the CAB worn by RM recruits is a factor which increases the risk of MT3SF.

The varied suggestions regarding TSF aetiology, and the general lack of understanding of MT3SF aetiology necessitated the exploration of risk factors predisposing RM recruits to these injuries. Study 3 was a prospective study which analysed selected gait, flexibility and anthropometric characteristics of 1065 RM recruits at the start of training. Those recruits sustaining a TSF or MT3SF during training were compared to a control group of 120 recruits who completed training without injury. Variables related to both the ability of the lower limb to resist loading and attenuate impacts were linked with risk of TSF, while individuals sustaining MT3SF displayed altered magnitude and timing of peak forefoot loads, suggested to be influenced by transverse and sagittal plane ankle kinematics and the timing of heel off. Both sets of injured recruits were of lower BMI and younger age than their uninjured counterparts. In light of these results, it was suggested that selection criteria for entry into the RMs could be tailored, in addition to the provision of cushioning insoles for at-risk recruits.

This thesis represents a considerable step forward in the understanding of MT3SF in particular, and provides information on specific risk factors for MT3SF and TSF in RM recruits during basic training. The implications of the results of this thesis are discussed with regard to each specific injury below.

6.2. Implications for tibial stress fracture aetiology and prevention

Study 3 adopted a prospective approach to determine whether it was possible to identify characteristics that predispose RM recruits to TSF during 32-week infantry training. A large body of research already exists with regard to TSF aetiology, however many of the risk factors previously identified are only relevant to certain populations, training environments or footwear conditions. Therefore a fairly broad approach, informed by a number of previous studies, was taken in this thesis. Several variables previously associated with TSF risk were shown to have little relevance to the RM recruit population. For example, neither rearfoot motion nor passive hip flexibility differed between groups, variables which had been identified previously as risk factors in military and athletic populations. In the population of interest, it is apparent that variables which relate to impact attenuation provided by the coupling of the rearfoot and shank; the magnitude of heel pressure and factors which influence either the ability of the lower limb to resist extrinsic loading, or which are indicative of bone strength, are important.

Reduced range of axial rotation of the shank relative to the foot has not been identified as a variable of interest in TSF aetiology before, presumably because other aspects of lower limb function such as rearfoot inversion-eversion and ankle plantar-dorsiflexion play more important roles in attenuating ground reaction forces. In the light of the findings of Study 3, it is proposed that the role of standard issue footwear is important in determining which aspects of gait contribute to injury risk. In the combat boot, which may restrict rearfoot and ankle movement through provision of a stiff leather upper, it is suggested that the importance of axial tibial rotation, which is less effectively restricted by a leather upper, becomes important. Since prospective data collected in this thesis were for barefoot running, this suggestion requires further investigation. One such approach could be through the development of prototype footwear which restricts different aspects of lower limb function, with resultant kinematics and kinetics being quantified. Given that injured recruits also displayed greater heel pressures; lower age; lower BMI; and indicators of lower tibial width and calf muscle mass, the combination of factors was proposed to increase bone loading. These results affirm the notion that TSF risk is not only multi-factorial, but is influenced by wider extrinsic factors such as footwear, as well as individual characteristics.

The observation of lower BMI, calf girth and age in injured recruits was suggested to be linked with reduced bone strength. Lean muscle mass is associated with greater bone density (and by implication, greater strength) (Felson et al., 1993; McGuigan et al., 2002). In this athletic population, it is likely that greater BMI is associated with greater lean muscle, while calf muscle mass

is also correlated specifically with tibial bone mineral density (Bennell et al., 1996). When combined with the potential presence of growth plates (Adirim & Cheng, 2003) and narrower tibia (suggested by lower bimalleolar distance), the loads experienced during RM training may be more likely to trigger the remodelling process, with the continuous nature of training causing this to escalate into further damage, leading to proliferation of microcracks and eventually stress fracture. Should the results of this investigation inform MOD policy, it may be necessary to modify selection criteria based on minimum age or BMI requirements, or in the case of recruits demonstrating characteristics of increased risk of TSF, the provision of cushioning insoles may reduce external loads and reduce the threat of injury.

Cushioning insoles have been trialled in previous military populations with mixed success (e.g. Gardner et al., 1998; Milgrom et al., 1985) being reported. For example, in the RM population, House et al. (2013) successfully reduced the incidence of TSF with randomly administered shock-absorbing insoles. Even in examples where the incidence of some stress fractures has been reduced, other injuries remain at similar levels of prevalence. It is suggested that the provision of arbitrarily designed insoles may help some individuals whose requirements happen to meet the features provided by the insole, whereas the success of an insole provision strategy could be greatly improved if the properties of insoles are tailored to the needs of the at-risk individual. While this thesis has identified risk factors for just two types of injury, a similar analysis could be provided for others, with future screening identifying those recruits in need of certain insoles. For example, a recruit demonstrating risk

factors associated with TSF could be given a cushioning insole, while a custom orthotic may be more appropriate for another recruit showing mechanisms associated with a different injury. This thesis demonstrates that it is possible to perform such a large-scale assessment on a continuous basis, while the work of Franklyn-Miller et al. (2011) demonstrated that this type of approach can significantly reduce the incidence of overuse injury in military training. Although the associated costs of performing individual assessment may be considered high, this should be weighed against the multiple millions of pounds currently spent on rehabilitation from injury.

A wide range of risk factors for TSF were able to be considered in this investigation, however the strength of some of the key findings could have been strengthened with the use of additional equipment. For example, ground reaction forces have been frequently investigated with regard to TSF risk, but the use of a force plate was not possible in the testing environment. In addition to adding information regarding vertical impact forces, loading rate of impact force and the magnitude of free moment, interpretations regarding peak pressures at the heel would also have been strengthened with the inclusion of force data. Furthermore, with ground reaction force data, full inverse dynamics solutions could have been used to indicate bending moments acting on the tibia, as demonstrated by Haris Phuah et al. (2010). Although clear and consistent associations between ground reaction forces and TSF risk are lacking in the literature (Zadpoor & Nikooyan, 2011) the addition of force plate data in future may provide important information regarding tibial loading during gait (Lafortune & Hennig, 1995).

It is assumed here that the data for BMI, age, calf girth and bimalleolar width indicate that the TSF cases had weaker and narrower tibiae than controls. Verification of these properties is possible through scanning techniques such as dual-energy x-ray absorptiometry (DEXA), and has been demonstrated previously (e.g. Beck et al., 1996, 2000; Bennell et al., 1996). However, this is an expensive process, and regarding estimates of tibial geometry, the work of Franklyn et al. (2008) reported several measurements taken from tibial radiographs that could translate to measurements of the shank taken using callipers or a tape measure. For example, in addition to the bimalleolar distance, the width of the tibial plateau and the length of the tibia could be measured and utilised to develop a more accurate estimate of the geometry of the tibia.

6.3. Implications for third metatarsal stress fracture aetiology

With previous research into MT3SF being minimal, this thesis provides a number of advances in the understanding of the aetiology of this injury. Existing suggestions linked heightened risk of MT3SF with reduced passive ankle dorsiflexion (Hughes, 1985); while recent retrospective analysis of RM recruits sustaining a MT3SF (Dixon et al., 2006), and mathematical modelling of the metatarsals (Arangio et al., 1998) highlighted that horizontal loading of the forefoot may be important. Additionally, with forefoot pathology in general

(including metatarsal stress fracture), greater plantar loading has been implicated, although there has been a tendency not consider the metatarsals individually and no prospective evidence of a direct association between plantar loading characteristics and MT3SF has been found. In fact, no prospective study examining gait-related risk factors for MT3SF has been conducted prior to this investigation. The three studies presented in this thesis have attempted to address a number of the gaps in the literature, and highlight the importance of lower limb gait characteristics in the development of this injury.

With MT3SF accounting for an unusually large proportion of lower limb stress fractures in RM recruits, Study 1 focused on whether the footwear provided to all RM recruits could influence risk factors for this injury. The results for the CAB supported some of the previous suggestions regarding injury, including restricting ankle dorsiflexion and increasing plantar forefoot loading. Study 2 provided a mathematical model which was able to verify that peak MT3 bending stress was significantly greater in the CAB when compared to the GT, which was representative of a neutral cushioned trainer. With Study 3 highlighting greater peak pressures in the forefoot of injured recruits compared to controls, it would appear that forefoot loading is a key variable determining MT3SF risk. Although a given individual will experience greater MT3 stress with increased magnitude of plantar loading, Study 2 also serves to highlight that individual metatarsal geometry is important in determining the extent to which this translates to strain, and ultimately whether it reaches a damaging level. Even within a group of males with the same sized feet, differences in metatarsal geometry were largely responsible for a wide range of peak MT3 bending

stresses. Within a group of 1065 recruits of varying physical characteristics, metatarsal geometry is likely to vary widely, therefore caution is urged when considering only forefoot peak pressures in MT3SF risk. As with TSF, the causes of injury are likely to be multi-factorial, with factors affecting both the nature of the applied load and the ability of the metatarsal to withstand that load being of importance.

In the MT3SF group in Study 3, the presence of greater peak forefoot pressures was suggested to occur due to certain kinematic characteristics, while others risk factors were associated with influencing whether loads are damaging. For example, the presence of a more abducted foot and more plantarflexed ankle at touchdown was linked with greater magnitude and later occurrence of peak pressures, while lower age and BMI was linked with possible lower bone strength and earlier fatigue, causing loads to be more damaging. If it is assumed that recruits who are younger and have a lower BMI than controls are at elevated risk of a stress fracture, then the combination of their gait characteristics and training in the CAB may cause a tendency for MT3SF. In particular, it is unlikely that the tendency for a more abducted foot during stance would be influenced by the CAB, however the results from Study 1 suggest that the already-elevated forefoot loads in injured recruits may have been exacerbated by training in this footwear condition (compared to the GT). With forefoot loading appearing to have great importance in MT3SF aetiology, risk of this injury may be reduced with the provision of cushioning insoles. Specifically, insoles have been shown to reduce forefoot peak plantar pressures by 24% when worn in the combat assault boot (House et al., 2002), although metatarsal

stress fracture incidence has not been reduced in RM recruits through the provision of such insoles (House et al., 2013). This may reflect the need to target other variables linked with metatarsal stress fracture risk, as well as reducing forefoot peak pressures.

One limitation of only obtaining data on vertical loads in Study 3 is that the relative contribution of the nature of horizontal braking forces to injury risk could not be assessed. Previous literature (Arangio et al., 1998; Dixon et al., 2006) suggests that the magnitude and angle of application of horizontal loads could be of importance due to the relative weakness to lateral loads demonstrated by the MT3. Study 1 indicated that the CAB encourages a more laterally applied horizontal ground reaction force vector at the time of peak braking, supporting these suggestions. While Study 2 confirms that the third metatarsals of all participants were less well adapted to resist horizontal bending loads, Study 3 does not add further evidence to the discussion due to the absence of horizontal force data. Although this thesis does implicate the magnitude of vertical loads in MT3SF aetiology, it is suggested that horizontal loads should continue to be considered in future research. As with vertical loads, horizontal loads may also be modified through adaptations to footwear and such a strategy should be investigated further.

This thesis challenges the proposal by Hughes (1985) that restricted passive dorsiflexion leads to early heel off and subsequent earlier and greater loading of the forefoot. It must be considered that participants in the present work had 'normal' ranges of passive ankle flexibility, compared to the participants

presenting equinus deformity in the work of Hughes (1985), therefore within normal ranges of passive ankle function, such a relationship has not been shown here. There was no correlation between passive dorsiflexion and any dynamic function other than rearfoot range of motion. Furthermore, the contention that the timing of heel off influences the magnitude and timing of peak pressure was not conclusively supported by results obtained in this thesis. Although Study 3 identified a significant correlation between the timing of heel off and the timing of peak pressure at the M3 region, later heel off but earlier timing of peak pressure was observed in the CAB compared to the GT in Study 1. Additionally, Study 3 showed no correlation at all between the timing of heel off and the magnitude of peak pressure. Future research investigating kinematic aspects influencing the magnitude of forefoot loading in individuals without equinus deformity should therefore look beyond passive and dynamic ankle flexibility, with initial ankle position and foot abduction during stance suggested to be of importance here.

6.4. Further research

In light of the work presented in this thesis, further work should be undertaken to continue the pursuit of reduced stress fracture incidence during RM recruit training. The results of Chapter 3 suggest some clear areas for further research were identified, particularly with regard to improving design aspects of standard issue footwear. Development of more suitable footwear for RM training may include laboratory-based studies of design aspects which could be modified to, for example, allow greater range of motion at the ankle and reduce forefoot

loading. Any successful footwear designs could be investigated using a randomised control trial, and the influence on injury rates observed.

Some clear areas for improving the model presented in Chapter 4 were identified, including FE modelling and the integration of plantar attachments, however the potential application of the model to RM recruit training is limited. The model demonstrated the important influence of metatarsal geometry on peak stresses therefore this data would need to be obtained for any individual to whom the model is applied. This would be impractical for a large cohort, although one avenue for investigation may be to use portable scanning techniques to estimate bone properties, as demonstrated by Davey (2013) and Hazell, Vanstone, Rodd, Rauch, & Weiler (2013). Were bone geometry to be estimated with sufficient accuracy, kinematic and kinetic data obtained during screening could be used to model predicted MT3 stresses and potentially identify recruits at elevated risk of MT3SF. If effective, similar approaches could be used to model tibial bending stresses in future.

Further verification of the results of the prospective study is necessary, given that conclusions are drawn from a relatively low number of cases. Having established an effective strategy for screening recruits, and identified areas for investigation, the recommended improvements to the methods should be implemented and applied over a long-term data collection period. This would include repeating data collection using improved methods (e.g. bone scanning, force plate data) until greater numbers of stress fractures are obtained. With increased case numbers, logistic regression modelling can be performed to

provide a predictive model for injury, identifying the most important variables of interest. With this approach, the desired level of 'risk' can be identified (e.g. 60% or 80% chance of TSF or MT3SF) along with the range of values for each variable which classifies the individual as likely to be injured.

Once a more robust data set has been established, giving greater confidence in the variables predisposing to injury, interventions can be applied. These can be taken from the literature, or developed in smaller-scale laboratory studies. For example, different designs of cushioning insole could be developed to specifically target variables associated with each type of stress fracture. Interventions should be introduced at the same stage of training as the testing is performed. In practice, the established screening protocol would be used to screen recruits, following which interventions could be provided to recruits who are identified as 'at risk' by the predictive model. The effectiveness of the intervention may be tested by randomly assigning interventions to half of the 'at risk' recruits, and monitoring injury incidence in the intervention and control groups. This study design may face an ethical issue however, as it requires the investigators to actively withhold an intervention from an individual they consider to be at risk of injury. A defence to this approach may be that the interventions have not yet been proven to work in the population, therefore the control group are not being deliberately disadvantaged during training.

6.5. Summary of key findings for each injury

Tibial stress fracture

Compared with recruits who are likely to pass through training without injury, the characteristics of RM recruits at increased risk of TSF during training are:

- Younger age
- Lower BMI
- Lower corrected calf girth
- Narrower bimalleolar width
- Lower range of tibial rotation
- Greater peak pressure at the medial heel

These variables provide strong indications that poor bone strength, muscle insufficiency and poor impact attenuation are associated with TSF risk in this population. In response to this information, where future screening identifies recruits at high risk of injury, the prescription of cushioning insoles may be beneficial in reducing risk of TSF.

Third metatarsal stress fracture

Both individual characteristics and footwear have been implicated in the high rate of MT3SF amongst RM recruits. An analysis of the footwear issued to recruits as standard suggested that wearing the CAB modified aspects of gait in a manner that increased risk of MT3SF. Mathematical modelling of the influence of this footwear on bending stresses acting on the MT3 reinforced this suggestion. It also highlighted the importance of individual variation in bone

geometry in determining the extent to which external loads may reach damaging levels of bone stress. The influence of the combat assault boot on risk factors associated with MT3SF risk included:

- Increased plantar pressure at the M3 region
- More rapid and more laterally applied loading of the forefoot
- Restricted ankle dorsiflexion
- Suggested greater demand on plantarflexor muscles
- Increased bending stress placed on the MT3 mid shaft

Prospective analysis of individual characteristics predisposing RM recruits to risk of MT3SF suggests that the following variables are associated with this injury:

- Younger age
- Lower BMI
- Greater and later occurrence of peak pressure at the M3 region
- A more abducted foot during stance
- A later occurrence of heel off
- A more plantarflexed ankle position at touchdown

This thesis highlights ankle kinematics and plantar loading as risk factors associated with MT3SF. These risk factors may be influenced by footwear, and in the case of RM recruits, it is suggested that the CAB may exacerbate risk factors for MT3SF which present in recruits at the start of training. This was able to be demonstrated most clearly by mathematical modelling, which demonstrated that bending stresses acting on the metatarsal are significantly increased by wearing the CAB. Recommendations for the reduction of MT3SF risk therefore relate to the provision of devices to reduce forefoot loading.

APPENDICES

Appendix A: Evaluation of the mathematical model of bending stresses acting on the third metatarsal.

Introduction

Study 2 presents the results of an analysis of stress acting on the MT3 of five participants, estimated using a mathematical model which combines data for bone geometry, external load and metatarsal inclination. In order to assess the

validity of claims made in Study 2, the mathematical model must be evaluated as part of its development (Yeadon & King, in Payton & Bartlett, 2007). Where possible, the data produced by a model should be compared to actual values, such as those obtained using *in vivo* strain gauges. However, although strain data have been produced for the second metatarsal in particular (e.g. Arndt et al., 2003; Milgrom et al., 2011), no data exist in the literature for MT3 bending stress or strain during running. Although there is research suggesting that second and third metatarsal strains are of the same order (Gross & Bunch, 1989), the geometric properties of the two metatarsals which would influence peak stresses have been shown to differ (Griffin & Richmond, 2005). Additionally, it would not be appropriate to use the data obtained for walking in *in vivo* studies (e.g. Arndt et al., 2003; Milgrom et al., 2011) with estimates for running in the present model. Consequently, while magnitude of reported values can be compared with the model proposed by Gross & Bunch (1989), the key components of the model (geometric properties and plantar loading) presented here require evaluation by comparison to literature values. Additionally, assessment of the reliability of the model is necessary to determine the likelihood of error due to digitisation of MRI data, while information about the sensitivity of the model to changes in external conditions (load and angle of inclination) would provide useful information regarding the influence of error or natural variation in these variables.

Method

In order to assess the accuracy of the geometric properties obtained from digitisation of MRI scans, the values obtained for the five participants in the study were compared to reference data available in the literature. Given the propensity towards analysing the mid-point of the metatarsal in previous literature, values for CA, I_x , I_y and I_z at the mid-point were compared with the mid-point data presented by Griffin & Richmond (2005). These authors used CT scanning to obtain the geometric properties of the mid-point of 40 MT3s from male cadaver feet (Griffin & Richmond, 2005).

In order to assess the reliability of the geometric properties obtained by the digitising process, the MRI scans for participant 4 were digitised a total of three times and all stress calculations performed for the resulting data sets. The 3D model of the metatarsal produced by each repeat was reconstructed for visual inspection, and the mean, standard deviation (SD) and coefficient of variance (CV%) were calculated for each variable produced by the model (peak axial stress; peak vertical and horizontal compressive and tensile stresses; peak torsional stress; location of peak stresses; peak mid-point compressive, tensile and torsional stresses).

A sensitivity analysis was performed using the data for participant 4. The kinematic and kinetic data for this participant's dynamic trials were modified systematically to investigate the influence of changes in either peak plantar force or the orientation of the MT3 (inclination relative to the vertical) on calculated peak stresses. Three of the barefoot running trials (trials 1, 5 and 8 of 10) were independently manipulated and vertical compressive stress calculated

for each manipulation. Relative to the actual produced data, one-unit increments and decrements of up to 5 N of peak force and 5 degrees of MT3 angle were introduced, providing a range of 10 N and 10 degrees in total. These increments were chosen to represent deviation typically experienced between running trials. The mean change per unit was calculated and time histories were produced for the mid-point of the MT3 for dynamic trial number 8, highlighting the affect of the manipulation.

Results

Table 1 shows the comparison of geometric properties produced by digitisation of MRI scans, compared to reference values from Griffin & Richmond (2005). The mean mid-point CA in the present study is greater than the mean presented by Griffin & Richmond (2005), but sits comfortably within the ranges they present, as does the range of CA values. Similarly, the inertial properties reported in the present study all fall within the boundaries of the reference value ranges, although they are all slightly below the means presented by Griffin & Richmond (2005). The range of values presented here is also comfortably within the reference range.

Table 1. Comparison of MT3 mid-point properties calculated for the five participants in the present study with reference values from Griffin & Richmond (2005). Values are the mean (SD), with range in parentheses below.

| Property (units) | Present study value | Reference value (Griffin & Richmond, 2005) |
|------------------|---------------------|--|
|------------------|---------------------|--|

| | | |
|--------------------------------------|---------------------------------|---------------------------------|
| CA (mm ²) | 45.5 (4.3) [38.7 – 47.7] | 35.3 (8.7) [21.7 – 63.2] |
| I _x (mm ⁴) | 166.5 (41.4) [133.3 – 238.9] | 216.0 (105.2) [52.9 – 535.0] |
| I _y (mm ⁴) | 106.5 (25.7) [73.9 – 145.6] | 172.4 (91.2) [45.2 – 453.2] |
| I _z (mm ⁴) | 273.0 (64.4) [224.6 – 384.5] | 388.4 (190.8) [98.2 – 988.2] |

Figure 1 shows the 3D reconstructed images of the MT3 for the three repeats, demonstrating that visually the characteristics of the bone are similarly represented each time.

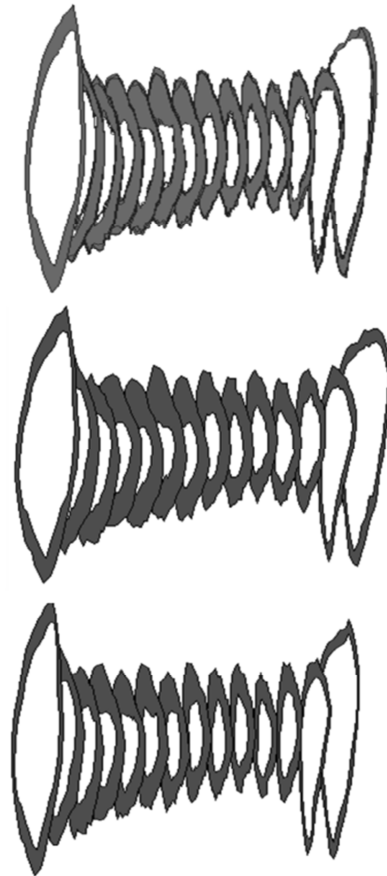


Figure 1. Reconstruction of the digitised slices of MRI scans obtained for participant 4 over three repeats (in sequence top to bottom). The proximal end of the metatarsal is orientated to the left.

Table 2 summarises the maximum, minimum and mid-point cross-sectional area (CA) for each repeat, and the resulting maximum axial stress (σ_{ax}). In each repeat, the location of minimum and maximum CA was the same, as was the location of peak σ_{ax} . Variation was largest (12.86%) for the calculation of maximum CA, which in each case was at the most proximal slice of the MT3. The values for mid-point and minimum CA, as well as peak σ_{ax} showed low variation (3.01 – 7.51%) giving confidence in their reliability.

Table 2. For each of three repeats, data are presented for the magnitudes and locations of minimum and maximum cross sectional area (CA); CA at mid-point; mean (SD) magnitude and location of peak axial stress (σ_{ax}). Locations are the distance from the distal end of the third metatarsal. The mean, SD, coefficient of variance (CV%) and range are presented for the three repeats of each variable.

| Repeat | Min CA (mm ²) | Location (mm) | Max CA (mm ²) | Location (mm) | Mid CA (mm ²) | Peak σ_{ax} (MPa) | Location (mm) |
|--------------|---------------------------|---------------|---------------------------|---------------|---------------------------|--------------------------|---------------|
| 1 | 32.46 | 15 | 94.73 | 65 | 46.19 | 4.09 | 15 |
| 2 | 35.59 | 15 | 94.16 | 65 | 43.85 | 3.73 | 15 |
| 3 | 30.70 | 15 | 74.87 | 65 | 43.08 | 4.33 | 15 |
| Mean | 32.91 | - | 87.92 | - | 43.37 | 4.05 | - |
| SD | 2.47 | - | 11.30 | - | 1.62 | 0.30 | - |
| CV% | 7.52 | - | 12.86 | - | 3.66 | 7.46 | - |
| Range | 6.89 | - | 19.86 | - | 3.11 | 0.60 | - |

The maximum vertical and torsional stresses are presented in Table 3, and the maximum horizontal stresses in Table 4. Values for the mid-point of the MT3 are shown in Table 5. For both maximal and mid-point peak stresses, variation was extremely small (CV% < 4, ranges < 4 MPa). The location of peak stress varies across the repeats however, probably due to errors in digitisation influencing the distances y and R in particular. Figure 2 showing the maximum values for y and R for each slice of the MT3 for the three repeats, highlighting the variation between trials.

Table 3. For each of three repeats, means (SD) are presented for the magnitudes and locations of maximum vertical stresses. Peak torsional stress is also included. Locations are the distance from the distal end of the third metatarsal. The mean, SD, coefficient of variance (CV%) and range are presented for the three repeats of each variable.

| Repeat | Peak σ_c (MPa) | Peak σ_t (MPa) | Location (mm) | Peak σ_{tor} (MPa) | Location (mm) |
|--------------|--------------------------|--------------------------|------------------|------------------------------|------------------|
| 1 | 77.35 (22.16) | -72.23 (20.78) | 30 mm | 34.76 (9.98) | 30 mm |
| 2 | 77.79 (21.58) | -72.44 (20.18) | 30 mm | 34.92 (9.71) | 30 mm |
| 3 | 78.09 (21.67) | -73.70 (20.53) | 40 mm | 37.03 (10.29) | 45 mm |
| Mean | 77.74 | -72.79 | - | 35.57 | - |
| SD | 0.37 | 0.80 | - | 1.27 | - |
| CV% | 0.48 | 1.09 | - | 3.56 | - |
| Range | 0.74 | 1.47 | - | 2.27 | - |

σ_c = compressive stress; σ_t = tensile stress; σ_{tor} = torsional stress.

Table 4. For each of three repeats, means (SD) are presented for the magnitudes and locations of maximum horizontal stresses. Peak torsional stress is also included. Locations are the distance from the distal end of the third metatarsal. The mean, SD, coefficient of variance (CV%) and range are presented for the three repeats of each variable.

| Repeat | Peak σ_c (MPa) | Peak σ_t (MPa) | Location (mm) |
|--------------|--------------------------|--------------------------|------------------|
| 1 | 89.82 (25.75) | -86.07 (24.74) | 45 mm |
| 2 | 90.30 (25.07) | -86.36 (24.04) | 45 mm |
| 3 | 93.67 (23.18) | -89.68 (22.28) | 40 mm |
| Mean | 91.26 | -87.37 | - |
| SD | 3.10 | 2.01 | - |
| CV% | 2.30 | 2.30 | - |
| Range | 3.85 | 3.61 | - |

Table 5. For each of three repeats, means (SD) are presented for the magnitudes of maximum axial, vertical, horizontal and torsional stresses at the mid-point of the MT3. The mean, SD, coefficient of variance (CV%) and range are presented for the three repeats of each variable.

| Repeat | Vertical | | | | Horizontal | |
|--------------|-----------------------------|------------------------------|--------------------------|--------------------------|--------------------------|--------------------------|
| | Peak σ_{ax} (MPa) | Peak σ_{tor} (MPa) | Peak σ_c (MPa) | Peak σ_t (MPa) | Peak σ_c (MPa) | Peak σ_t (MPa) |
| 1 | 2.00 (0.71) | 33.74 (9.65) | 71.03 (20.28) | -65.52 (18.79) | 85.19 (24.33) | -79.58 (22.82) |
| 2 | 2.10 (0.74) | 34.69 (9.73) | 75.99 (21.89) | -68.24 (19.36) | 87.30 (24.37) | -81.13 (22.64) |
| 3 | 2.02 (0.86) | 33.52 (9.18) | 72.65 (19.66) | -69.49 (19.55) | 83.36 (22.91) | -78.44 (21.85) |
| Mean | 2.04 | 33.98 | 73.22 | -67.75 | 85.28 | -79.72 |
| SD | 0.05 | 0.62 | 2.53 | 0.03 | 1.97 | 1.35 |
| CV% | 2.57 | 1.83 | 3.45 | 3.00 | 2.31 | 1.69 |
| Range | 0.10 | 0.22 | 1.62 | 3.97 | 1.83 | 1.14 |

σ_{ax} = axial stress; σ_c = compressive stress; σ_t = tensile stress; σ_{tor} = torsional stress.

Inspection of Figure 2 suggests that variation increases for slices located towards either end of the metatarsal. The mid-section of the MT3 shaft, which is commonly reported, shows minimal variation, as do the peak stress values.

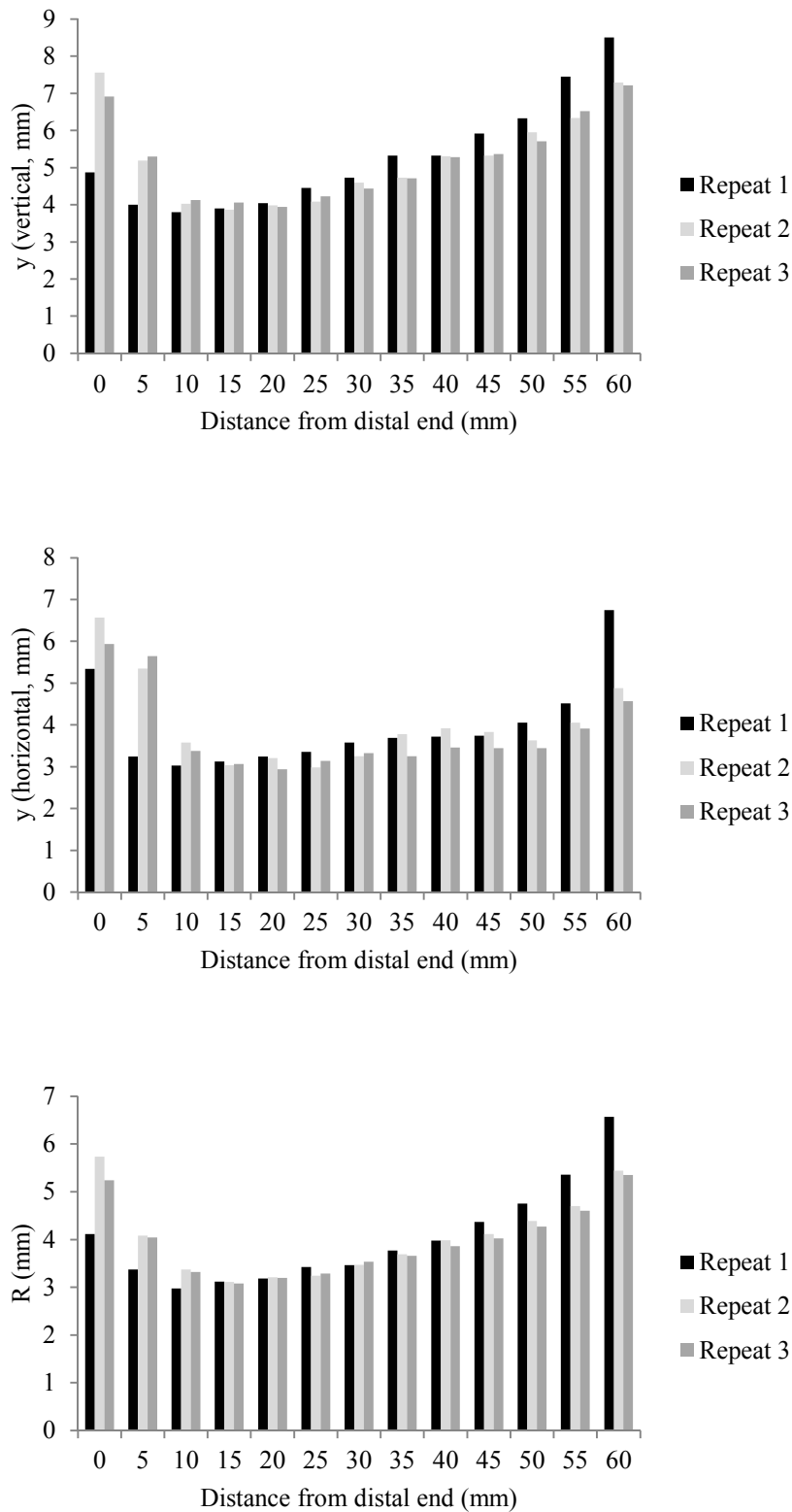


Figure 2. Moment arms influencing stress calculations for each slice of the third metatarsal of participant 4. For each direction (dorsi-plantar (top), medio-lateral (middle), rotational (bottom)) the result obtained for each of three repeats is presented.

Figure 3 displays time histories produced for the mid-point of the MT3 for dynamic trial 8, including the influence of manipulations of ± 5 units of force or angle of inclination of the MT3 respectively. Table 6 provides a summary of the peak stresses calculated at the ranges for the three randomly selected trials (trials 1, 5 and 8). Manipulation of force and angle data resulted in changes in compressive stress. Results across the three trials indicate that the effects are not consistent in magnitude. It is clear that an increase in angle of inclination of the MT3 to the vertical results in a decrease in peak stress, through reduction of the perpendicular distance between the MT3 head and the mid-section of the MT3 (see Figure 4). Increasing the force applied at the MT3 head also increases the peak compressive stress, with an average increase of 0.6 MPa per Newton of additional force. This relates to an increase in strain of approximately $35\mu\epsilon$ per Newton.

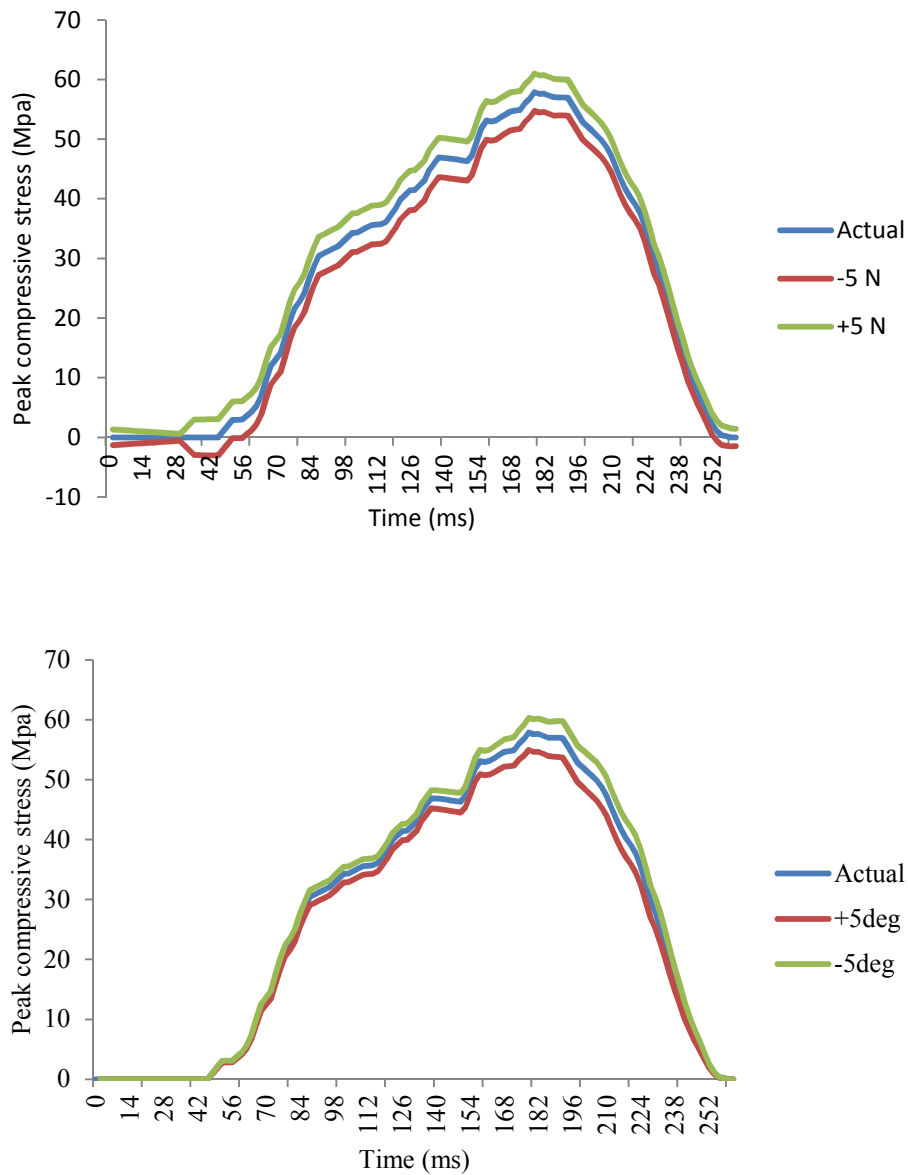


Figure 3. Time histories for compressive stress at the mid-point of the MT3 when the plantar force (top) or angle of inclination of the MT3 (bottom) are manipulated for trial 8 of participant 4. The actual data presented by the model is shown with the effects of increasing or reducing the independent variable by 5 units. Note: An increase in angle represents steeper inclination to the ground.



Figure 4. Example of the influence of metatarsal inclination on the moment arm influencing the magnitude of bending stresses. The horizontal dashed line represents the perpendicular moment arm between the point of force application (metatarsal head) and the section of bone (approximate mid-point). The moment arm increases as the foot moves from a steep inclination (top) to shallow inclination (bottom).

Table 6. The influence of manipulating plantar force and MT3 angle of inclination on compressive stress at the metatarsal mid-point of participant 4. The mean change per unit change is shown for force and angle manipulations, the peak value for the normal and maximal manipulations is included, as well as the range of values over the 10 unit change. Manipulations for three trials are shown.

| Trial 1 | | | |
|--------------|-------------------|-------------------|-------------------|
| Force | Peak stress (MPa) | Angle | Peak stress (MPa) |
| Actual | 40.46 | Actual | 40.46 |
| +5 N | 43.47 | +5 degrees | 38.14 |
| -5 N | 37.45 | -5 degrees | 42.48 |
| Range | 6.02 | Range | 4.34 |
| Change per N | ±0.60 | Change per degree | ±0.43 |
| Trial 5 | | | |
| Actual | 66.44 | Actual | 66.44 |
| +5 N | 69.31 | +5 degrees | 62.06 |
| -5 N | 63.57 | -5 degrees | 70.33 |
| Range | 5.74 | Range | 8.27 |
| Change per N | ±0.57 | Change per degree | ±0.83 |
| Trial 8 | | | |
| Actual | 58.48 | Actual | 58.48 |
| +5 N | 61.62 | +5 degrees | 55.63 |
| -5 N | 55.34 | -5 degrees | 60.90 |
| Range | 6.28 | Range | 5.27 |
| Change per N | ±0.63 | Change per degree | ±0.53 |

Note: An increase in MT3 angle represents steeper inclination to the ground.

Discussion

Values for the geometric properties of the MT3 produced by the model relate well with reference data from the literature. Peak stresses are highly repeatable; however digitisation errors may be significant, potentially influencing the location of peak stresses in particular. These errors appear notable towards either end of the bone, but minimal in the mid-shaft, which is the area where maximum bending stresses occur. Results for the mid-shaft and peak bending stresses provide confidence in the results presented in Study 2.

Further comparison with previous research shows good agreement with values presented in the current model. The maximum MT3 bending moment presented by Gross & Bunch (1989) was 4.74 Nm, compared with a mean [min – max] for one participant in the present study of 3.57 Nm [1.83 – 6.49]. At 200 N, the peak plantar force measured at the MT3 head by Gross & Bunch (1989) is greater than the value presented here (147 N), offering an explanation for the greater bending moments these authors estimated. Metatarsal geometry was simplified to that of a hollow ellipse by Gross & Bunch (1989), with a major metatarsal radius of 4.55 mm indicated. This compares well with the maximum vertical (5.32 mm) and horizontal (3.69 mm) radii at the mid-section of the representative participant in the present study. It is not clear how the moment of inertia was calculated by Gross & Bunch (1989). Although the model is detailed in their paper, using the available data provides a result incongruent with their published peak strains. However, as moment of inertia calculations presented here are in agreement with those reported by Griffin & Richmond (2005), it is

concluded that the values used at each step of the model are within acceptable ranges and valid for calculation of bending stress. An example of the calculation of peak compressive stress in the mid-section of the MT3 of participant 4 (running trial 8) is shown below:

Axial stress = Force / Area

$$= 94.70 \text{ N} / 50.54 \text{ mm}^2$$

$$= \underline{1.87 \text{ MPa}}$$

Bending moment = Force * perpendicular distance from intersection to point of application (sample angle = 27 degrees)

$$= 94.70 \text{ N} * ((\cos 27^\circ) * 35 \text{ mm})$$

$$= \underline{2953.69 \text{ N.mm}}$$

Maximum vertical bending stress = ((bending moment (N.m) * maximum vertical outer radius (m)) / moment of inertia about x-axis through centroid (m⁴))

$$= ((2953.69 \text{ N.mm} * 5.32 \text{ mm}) / 312.6 \text{ mm}^4)$$

$$= 15713.63 / 312.6$$

$$= \underline{50.27 \text{ N/mm}^2 (50.27 \text{ MPa})}$$

Maximum compressive stress = axial stress + bending stress

$$= 1.87 \text{ MPa} + 50.28 \text{ MPa}$$

$$= \underline{52.14 \text{ MPa}}$$

Estimated peak compressive strain can be calculated by dividing the peak stress by Young's modulus, with units in parts per million (microstrain – $\mu\epsilon$).

Young's modulus has been reported to lie in the region of 17 GPa for cortical

bone (Reilly, Burstein & Frankel, 1974), a value also used by Gross & Bunch (1989). For the example above:

$$\begin{aligned}\text{Peak strain} &= (52.14 \text{ MPa} / 17000 \text{ MPa}) * 1000000 \\ &= \underline{3067 \mu\epsilon}\end{aligned}$$

Gross & Bunch (1989) estimated a peak strain of 5160 $\mu\epsilon$, which is in the same order of magnitude of the result provided here. It is clear from the stepwise progression of the present model that differences in peak plantar force and estimation of geometry may account for differences in the obtained values. Furthermore, the previous model accounted for contact forces with the toes, as well as the influence of plantar musculature, which will influence calculated strains (Gross & Bunch, 1989).

Decreasing the inclination of the metatarsal to the ground or increasing the load applied causes increases in peak calculated stress, although the systematic modification of these variables did not result in a systematic influence on calculated values. The influence of changes in force was seen to be greater than the influence of changes in angle. Given that natural variation between running trials will cause perturbations in the angle of the foot and peak force under the metatarsal head, the use of several gait trials to provide a mean peak stress is beneficial.

Summary

In the absence of reference values for peak bending stress/strain to compare to, the model presented in Study 2 was evaluated in relation to reference values for geometric properties and against the values estimated by the most comparable model in the literature (Gross & Bunch, 1989). Compared to reference values for the cross-sectional area and inertial properties, the data produced through analysis of MRI scans were within acceptable ranges, giving confidence in these values. Values for peak stress were not only in the same order of the model of Gross & Bunch (1989), but were repeated with a good degree of reliability when the modelling process was repeated three times for one participant. The model is sensitive to perturbations in the angle of inclination of the metatarsal and the peak plantar force acting on the metatarsal head, such that a decrease in angle and increase in force result in greater peak stresses. The natural variation in foot angle and plantar forces seen between running strides therefore necessitates the use of multiple trials to produce reliable mean data with this model.

Appendix B: Determination of control group size for use in Study 3

Study three involved the comparison of injury groups with a control group. While injury group size was determined by injury occurrence and availability of data, eligible controls were those who completed training with their original troop. As such, a relatively large number of controls were available for analysis (n=419). To determine the required size of a representative sub-sample of controls, the following analyses were performed.

Stage 1 – Estimate of representative sample

There are standard approaches to estimating the sample size required to estimate a population mean, in this case the control group representing the population of recruits passing out of training with their original troop. For a given precision, a specific estimate of the sample size required (n) can be produced using the following equation (from Putt, Shaw, Woods, Tyler & James, 1987):

$$n = \frac{N * 1.96^2 * S^2}{N * d^2 + 1.96^2 * S^2}$$

Where N = total population size (419); d = maximum difference to be tolerated between the sample mean and the true mean; S = standard deviation of

sample. 1.96 is used as the multiplier in the equation in order to ensure 95% confidence that d will be met.

This calculation was performed for the following precision levels: a) within 10% of S ; b) within 15% of S ; c) within 20% of S ; d) within 25% of S . Level d) represents the standard deviation criterion suggested by Bates et al. (1983).

Table 1 displays the suggested sample size for each precision level.

Table 1. Suggested number (n) required for representation of population mean for each of four levels of precision.

| Precision level | n |
|-----------------|-----|
| A | 201 |
| B | 122 |
| C | 79 |
| D | 54 |

Given the available cases, a precision level of 10% of the group standard deviation could be achieved. Although this analysis is a useful reference for estimating the precision of the control group data, it is strictly an *a priori* analysis, and the prediction equation does not account for the levels of variation in different variables. Stage 2 of the analysis utilised a more detailed approach to determination of the control group size.

Stage 2 – Cumulative means analysis

Stage 2 used a cumulative means procedure, based on that suggested by Bates et al. (1983) for the determination of trial numbers. The mean of 150 cases was calculated for each of 132 variables summarised in Table 2. This group of data was used as a stable representation of all passed out recruits. Previous studies utilising gait variables in prospective cohort designs have typically used smaller sample sizes, for example 39 controls (Bennell et al., 2010) and 20 controls (Rice et al., 2013), although Williams et al. (2006) utilised 167 control legs when analysing plantar pressure data. The use of 150 controls was deemed a suitable representation of the control cohort.

For each variable, the criterion mean was calculated as the mean of 150 recruits. The criterion difference value was calculated as 0.25 of the standard deviation of 150 recruits. Cumulative means were then calculated for $n=1$ up to $n=150$ for each variable, and the difference between this cumulative mean and the criterion mean was calculated. The point at which this difference no longer exceeded the criterion difference value was recorded. This represented the minimum number of controls above which inclusion of additional controls would not significantly change the group mean.

The results of this analysis indicated that at most, 100 controls are required. This was the maximum number of controls required to achieve stability in any variable (hip range of motion). Figure 1 displays the cumulative means graph for

this variable. Six further variables required 89 or 90 cases for stability, while all other variable was stable after 68 cases were included.

Table 2. Variables included in the analysis.

| Variable group | Region | Specific variables included |
|------------------|----------------------------|---|
| Kinematics | Foot | Angle at touchdown |
| | Ankle | Angle at touchdown |
| | Rearfoot | Initial plantarflexion (ankle) |
| | Knee | Peak flexion |
| | Tibia | Range of motion |
| | | Time of peak flexion |
| | | Rate of peak flexion |
| | Other | Stride length (m) Stride length (% height) |
| Plantar pressure | Hallux | Peak pressure (masks/zones) |
| | Metatarsal heads 1-5 | Impulse (masks/zones) |
| | | Time of peak pressure (masks/zones) |
| | Medial heel, lateral heel | Relative pressure of each region (masks/zone) |
| | | Peak force (zones) |
| | | Time of peak force (zones) |
| | Whole foot | Ground contact time |
| | | Rearfoot surface contact % |
| | | Midfoot surface contact % |
| | | Forefoot surface contact % |
| | | Rearfoot surface impulse % |
| | Midfoot surface impulse % | |
| | Forefoot surface impulse % | |
| Range of motion | Foot axis angle | |

| | | |
|---------------|-----------------|-------------------------------------|
| | | Total abduction-adduction |
| | | Subtalar joint flexibility |
| | | Subtalar joint minimum angle |
| | | Subtalar joint maximum angle |
| Anthropometry | Flexibility | Passive dorsiflexion |
| | | Hip internal rotation |
| | | Hip external rotation |
| | | Relative internal-external rotation |
| | Characteristics | Age |
| | | Height |
| | | Mass |
| | | BMI |
| | | Calf girth |
| | | Corrected calf girth |
| | | Ankle width |

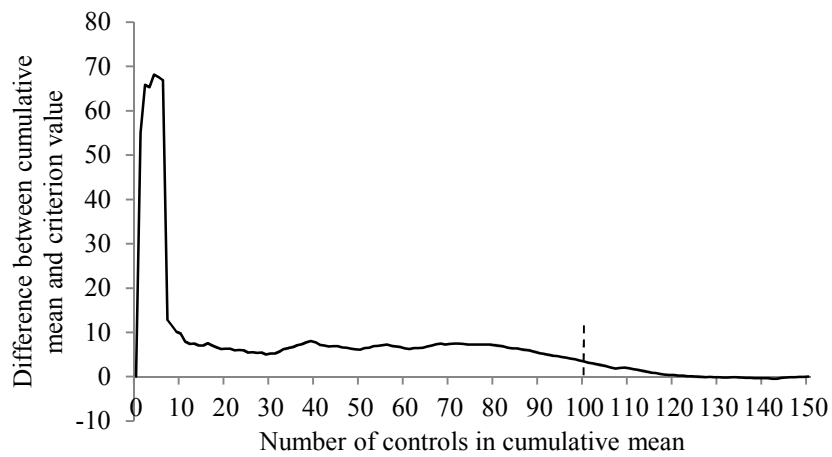


Figure 1. Cumulative means analysis for hip range of motion. The dotted line shows the point at which the cumulative mean consistently passes within 0.25 of the standard deviation of the mean of 150 cases (n=101).

Stage 3 – Confidence interval analysis

Although useful, the Bates et al. (1983) procedure is intended for use in the calculation of the number of trials required for stable data. In order to determine whether the mean of a sub-sample of available controls is representative of this population, a more suitable approach might be to investigate whether the mean of that sub-sample is within the 95% confidence intervals (95%CI) for that variable. In the second stage of analysis, 95% CI were calculated for each of 132 available variables summarised in Table 2 using the formula:

$$\text{Mean of variable} \pm 1.96 * SE$$

Where SE is standard error, calculated using:

$$SE = SD * \sqrt{\frac{1-f}{n}}$$

Where SD is the standard deviation of the sample, f is the proportion of the population included in the sample (150/419), and n is the sample size (150). Upper and lower confidence intervals were calculated for each variable. The cumulative mean of each variable was then calculated and assessed to identify whether it was between the 95% CI limits. The number at which any further

addition to the sample size no longer moved the mean away from the limits of confidence was noted for each variable.

Results of this analysis highlighted rearfoot rate of eversion as the variable requiring the highest number of recruits for stability, with 114 recruits required. Hip variables stabilised at 110 recruits, while six other variables required between 97-104 cases for stability. All other variables were stable after 84 cases.

Discussion and conclusion

Stage 1 of the analysis utilised a standard approach to predict the required sample size for estimation of the whole group mean. This estimated that for the given desired precision levels, between 54 and 201 recruits could be included in the analysis. Further analysis in stages 2 and 3 revealed that a sample size of at least 114 would be required to ensure stable representative data for all variables. Based on this, it was decided to include 120 recruits in the control group. Referring again to Stage 1 analysis, this would mean that the group mean would be within approximately 15% of the group standard deviation from the population mean. These analyses give confidence that this sub-sample of 120 of the 419 recruits provides a stable and accurate representation of the population.

Appendix C: Analysis of the accuracy and reliability of the Coda mpx30 system

Introduction

It is necessary to assess the quality of data obtained through scientific research. Data published in research is often used to inform clinical practice, as well as making meaningful contributions to the knowledge base, however there are inherent errors in almost all data obtained. The accuracy and precision of the testing system will determine the value of results obtained (Payton & Bartlett, 2007), making quantification of the errors produced important.

Kinematic analysis of gait is commonly used in clinical and research settings (Montgomery & Connolly, 2002), and a number of studies have been performed to assess the reliability of 3D gait analysis systems. A review of the reliability of 3D gait analysis systems by McGinley, Baker, Wolfe & Morris (2009) concluded that acceptable errors can be achieved when comparing different systems and different methods of assessing system reliability. The bilateral Cartesian Optoelectronic Dynamic Anthropometer (Coda) mpx30 setup (Charnwood Dynamics Ltd., Leicestershire, UK) used to capture 3D kinematic data in study three has previously been assessed in terms of reliability and accuracy (Maynard, Bakheit, Oldham, & Freeman, 2003; Monaghan, Delahunt & Caulfield, 2007; Birch & Deschamps, 2011).

Maynard et al. (2003) assessed the repeatability of measurements obtained during walking with a marker protocol utilising active markers placed both directly on the skin and attached to wands. Of the kinematic variables assessed, joint angles of the hip, knee and ankle were included. Only one trial was obtained for each of ten subjects, with data collection repeated later on the same day, and a week later. Two bilaterally aligned Coda mpx30 units were used to collect marker positions. In terms of the inter-rater reliability reported in the study, generally weak intra-class correlation (ICC) coefficients were reported (<0.75) were reported between sessions. The design of this study points to three potential causes of variability in results: system error; subject error; rater error.

Monaghan et al. (2007) assessed gait parameters in a group of ten volunteers on two sessions, one week apart. Participants were required to walk along a runway, and were assessed using a single Coda mpx30 unit. Markers were attached both to the skin and to wands, and ten trials were obtained in each session. The means of two, four, six, eight and ten trials collected on days one and two were compared across a number of variables, including hip, knee and ankle angles. ICC coefficients were generally greater than those reported in Maynard et al. (2003), with a trend for values to improve with increasing trial numbers, although very good (>0.85) ratings were achieved at the ankle from just four trials. The use of one skilled rater in this study reduced the likelihood of variability due to rater error, however system and subject errors may still have occurred between sessions. A more rigorous assessment of the accuracy and reliability of the Coda mpx30 system was performed by Birch & Deschamps in 2011.

Birch & Deschamps (2011) performed three assessment protocols without the influence of subject or rater variability. Their first assessment focused on the capture of a single, static marker, with ten sets of trials (5 seconds capture at 200 Hz) performed on each of five consecutive days. The second assessment tested the measurement of an aluminium frame with two arms set at 90 degrees. The frame was placed at four different orientations and fifty one-second trials were collected. The final approach utilised active markers placed on a model of the lower limb, with marker sets placed on the foot and shank segments. Fifty trials at 200 Hz were collected, for 5 seconds each, with markers in a static location. Results showed that repeatable data were produced for marker locations and angle calculations. The Coda performed worst in the y-direction (parallel to the direction of travel, perpendicular to the mpx30 units), with some concerns raised about the potential knock-on effect of variations in marker position estimates when angles are calculated.

Previous studies have attempted to quantify the reliability and accuracy of the Coda mpx30 system, however it is clear that there are a number of confounding factors that can influence the results obtained in terms of the subjects, system and raters. Subject and rater issues have been addressed in the literature and discussed in Chapter 2. The accuracy of the Coda system was of interest in this study. Due to the relative age of the system, and the wear-and-tear likely encountered by a portable system, it was important to ascertain whether the errors produced in the system were acceptable.

Methods

Procedure

A simple procedure was designed to assess the accuracy and repeatability of measurements acquired in a single session. Testing took place prior to a data collection session, with the full equipment setup in place. Please refer to for section 5.2 for a full description of the equipment setup. The Coda mpx30 was aligned prior to testing, using a three-marker axis definition. The y-axis was defined as the direction of travel.

Five active markers were attached to a rigid object 0.5m long (Figure 1) using a combination of 2- and 4-marker drive boxes. The object was selected as it allowed a 'handle' to be gripped without obstructing the line of sight between the markers and the Coda units. Markers were placed in positions deemed representative of the typical proximities observed in the marker protocol used in Study 3, with the mix of drive box types used for the same reason. Markers and drive boxes were attached using micro pore tape. Three segments were defined using the marker locations, and the distances between them measured using a Silverflex anatomical metal tape measure (Rabone Chesterman, England) placed flush against the object. Measurements were taken from the centre of each marker dome, and reported to the nearest 1 mm. The segment lengths are shown in Table 1.

Static and dynamic measurements were taken of the object. Static trials were collected with the object placed in the centre of the testing area. Five trials of 5

seconds at 200 Hz were collected. Dynamic trials were collected with the test object being held at the end without markers attached by a tester. The tester walked along the length of the runway, holding the test object in front of them and moving it in multiple directions. Medial-lateral, anterior-posterior and vertical movements were carried out for 5 seconds at 200 Hz. Ten successful trials were collected. A trial was deemed successful if all markers were visible for a minimum of 95% of the trial, as confirmed by the Coda in-view summary. 3D coordinate data were exported from all static and dynamic trials for data analysis.

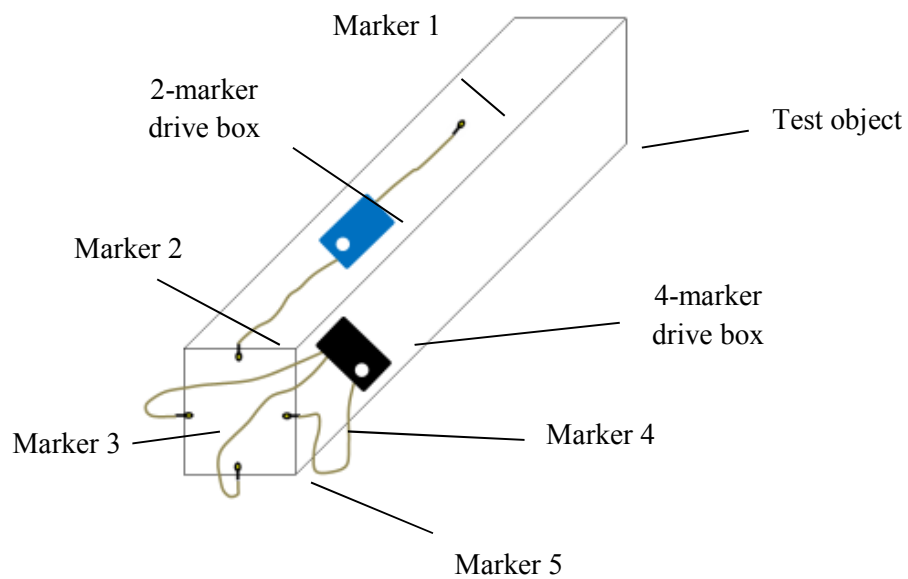


Figure 1. Location of active markers on test object.

Table 1. Segment definitions and lengths

| Segment | Definition | Length (mm) |
|-----------|----------------------|-------------|
| Long axis | Marker 1 to Marker 2 | 285 |
| Face 1 | Marker 2 to Marker 5 | 93 |
| Face 2 | Marker 3 to Marker 4 | 93 |

Data Analysis

To maintain validity with data analysis methods used in Study 3, raw coordinates were filtered using a fourth order, low pass Butterworth filter with a 12 Hz cutoff frequency in Matlab (v2011b, The Mathworks, USA). The three segments defined in Table 1 were then calculated in three dimensions, providing resultant segment lengths for the static and dynamic trials. These values were compared to the known lengths (Table 1) for accuracy, then between-trials for reliability. The mean difference between the measured and the true values was first calculated to provide an initial indication of accuracy. The root mean square error (RMSE) was then calculated. The RMSE calculation is considered to be a conservative assessment criterion (Payton & Bartlett, 2007), and is calculated by:

$$\text{RMSE} = \sqrt{\sum(X-\alpha)^2/(n-1)}$$

Where:

X = Measured Value

α = True Value

n = Number of times measurement was taken

The value achieved describes the overall residual error between data points, highlighting the accuracy of the measurement system (Mow & Huiskes, 2004). Coefficient of variation (CV%) was also calculated for the five static and ten dynamic trials, as the standard deviation divided by the mean, expressed as a percentage.

Results

Table 2 displays the results for the five static trials. The length of each segment is shown, as well as the difference between the known and measured length of each segment. The CV% and RMSE for each segment are also displayed.

Table 3 displays the same results for the ten dynamic trials.

Table 2. Results for the five static trials. The observed segment length and the difference from the measured segment are displayed for each segment (mm). The mean and SD of the five observed lengths are displayed, and the CV% and RMSE reported for the five trials.

| Trial | Long axis (mm) (actual = 285 mm) | | Face 1 (mm) (actual = 93 mm) | | Face 2 (mm) (actual = 93 mm) | |
|--------------|-------------------------------------|--------------|---------------------------------|-------------|---------------------------------|--------------|
| | Observed | Difference | Observed | Difference | Observed | Difference |
| 1 | 284.64 | -0.36 | 93.37 | 0.37 | 92.79 | -0.21 |
| 2 | 284.70 | -0.3 | 93.15 | 0.15 | 92.44 | -0.56 |
| 3 | 284.79 | -0.21 | 93.22 | 0.22 | 92.40 | -0.6 |
| 4 | 284.73 | -0.27 | 93.22 | 0.22 | 92.51 | -0.49 |
| 5 | 284.72 | -0.28 | 93.26 | 0.26 | 92.57 | -0.43 |
| Mean | 284.72 | -0.28 | 93.24 | 0.24 | 92.54 | -0.46 |
| SD | 0.05 | - | 0.08 | - | 0.15 | - |
| Range | 0.15 | - | 0.22 | - | 0.39 | - |
| CV% | 0.02 | - | 0.09 | - | 0.17 | - |
| RMSE | 0.36 | - | 0.31 | - | 0.57 | - |

Table 3. Results for the ten dynamic trials. The observed segment length and the difference from the measured segment are displayed for each segment (mm). The mean and SD of the five observed lengths are displayed, and the CV% and RMSE reported for the ten trials.

| | Long axis (mm) (actual = 285 mm) (static mean = 284.72) | | | Face 1 (mm) (actual = 93 mm) (static mean = 93.24) | | | Face 2 (mm) (actual = 93 mm) (static mean = 92.54) | | |
|-------|---|------------------------------|---------------------------|--|------------------------------|---------------------------|--|------------------------------|---------------------------|
| Trial | Observed | Difference from actual | Difference from static | Observed | Difference from actual | Difference from static | Observed | Difference from actual | Difference from static |
| 1 | 284.78 | -0.22 | -0.06 | 94.44 | 1.44 | -1.2 | 93.93 | 0.93 | -1.39 |
| 2 | 284.16 | -0.84 | 0.56 | 95.72 | 2.72 | -2.48 | 93.23 | 0.23 | -0.69 |
| 3 | 284.67 | -0.33 | 0.05 | 94.70 | 1.70 | -1.46 | 93.32 | 0.33 | -0.78 |
| 4 | 284.34 | -0.66 | 0.38 | 94.31 | 1.31 | -1.07 | 92.36 | -0.64 | 0.18 |
| 5 | 284.00 | -1.00 | 0.72 | 95.19 | 2.19 | -1.95 | 92.30 | -0.70 | 0.24 |
| 6 | 284.72 | -0.28 | 0 | 93.18 | 0.17 | 0.06 | 94.10 | 1.10 | -1.56 |
| 7 | 284.25 | -0.75 | 0.47 | 96.16 | 3.16 | -2.92 | 92.71 | -0.29 | -0.17 |
| 8 | 283.88 | -1.12 | 0.84 | 93.63 | 0.63 | -0.39 | 91.54 | -1.46 | 1 |
| 9 | 283.75 | -1.25 | 0.97 | 95.99 | 2.99 | -2.75 | 92.14 | -0.86 | 0.4 |
| 10 | 283.95 | -1.05 | 0.77 | 95.22 | 2.22 | -1.98 | 93.72 | 0.72 | -1.18 |
| Mean | 284.25 | 0.75 | 0.47 | 94.85 | 1.85 | -1.61 | 92.93 | -0.07 | -0.40 |
| SD | 0.37 | - | | 0.99 | - | | 0.85 | - | |
| Range | 1.03 | | | 2.98 | | | 2.56 | | |
| CV% | 0.13 | - | | 1.04 | - | | 0.92 | - | |
| RMSE | 0.83 | - | | 2.06 | - | | 0.07 | - | |

Discussion

When considering the precision of the Coda mpx30, it is important to consider that the object used in this study was not measured to a high degree of precision, and it is likely that the statically obtained values provide a more precise measure of its dimensions. A more valid focus is that pertaining to the accuracy and reliability of values provided when comparing the dynamically and statically reported data.

Static assessment

The mean difference for the supposed actual dimensions of the object and those reported by the Coda was 0.46 mm at most, with this value reported for the 'Face 2' dimension. The biggest difference for any trial was 0.6 mm, again for 'Face 2'. It is likely that any differences in these values is as a result of the measured value being incorrect. The greatest range of reported segment length was 0.39 mm, which compares favourably with the biggest range (0.72 mm) obtained in a static assessment by Birch & Deschamps (2011).

A CV% of at worst 0.17 is extremely low, especially when compared to Birch & Deschamps (2011), who reported a best CV% of 2.58 and worst of 35.36. The RMSE values, reporting a maximum value of 0.57 mm, are also extremely low. A report of the accuracy and reliability of the Vicon Peak Motus optical tracking system, considered a 'gold standard' system, reported RMSE values ranging from <0.5 to 2 mm (Scheirman & Aoki, 1999), although a slightly different approach was taken in this case.

The variation evident for the estimation of static dimensions was greater for the smaller dimensions, which was expected given the potential error when markers are clustered together. The maximum summed error between two segments was 0.93 mm, which can be considered acceptable (Birch & Deschamps, 2011). However, the effect this could potentially have on calculated angles would be greater for those composed of smaller segments, such as rearfoot eversion angle, than larger segments such as knee flexion angle.

Dynamic assessment

Assessment of dynamically obtained values was performed against both the known and statically obtained segment lengths. Results varied between segments, with the lowest mean difference from the actual length reported as 0.07 mm, for 'Face 2' and the largest at 1.85 mm for 'Face 1'. As stated previously, the measurement of the actual values may have been inaccurate, however these results vary from those obtained statically, and suggest that the dynamically obtained values were less accurate. This was confirmed when considering the difference between statically and dynamically obtained segment lengths, with 'Face 1' varying by 1.61 mm and 'Face 2' by 0.4 mm.

The largest variation between statically and dynamically obtained values was 2.92 mm, with ranges of observed values of 2.98 mm and 2.56 mm for 'Face 1' and 'Face 2' respectively. Values for the two face vectors varied more than those for the 'Long axis' vector, potentially due again to the proximity of markers at the end of the object. However, CV% remained extremely low, at a maximum of 1.04%, while the maximum RMSE of 2.06 mm still compares with the values obtained for the Vicon Peak system (Scheirman & Aoki, 1999), suggesting excellent reliability.

Summing the maximum error for any two segments however (difference from static values), provides a potential 3.48 mm error, which could translate into fairly large errors when considering angles between small segments. When considering that the difference in rearfoot eversion between TSF cases and

controls in Study 3 was 0.19 degrees, an error of 3.48 mm when reporting the relative position of the rearfoot and shank in the medio-lateral direction could equate to a large error in the reported angle.

Conclusions

A simple accuracy and reliability study was performed, and demonstrated that the Coda mpx30 system used in study 3 was able to produce repeatable data. The absolute accuracy of data acquired cannot be fully assessed, as the lengths of the three vectors were not measured to a degree of accuracy or precision comparable to that of the Coda mpx30 system. It should be noted that the percentage of time 'in view' of markers in this procedure does not accurately represent the typical time in view during a standard running trial, therefore error due to interpolation of points may be greater in actual data collected. However, as with all motion capture systems, sources of systematic and random error are present within data collection. The present results, combined with previous research, give a high level of confidence in the accuracy and reliability of data obtained using the Coda mpx30 system.

Appendix D: Forms

Appendix D(i): Study 1 + 2 information sheet



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“Investigating the effect of military boots on stress fractures of the foot in military recruits”

Please read this information sheet carefully before deciding whether or not to participate in the above study.

Background

The intense and repetitive nature of Royal Marines training lends itself to the development of stress fractures of the metatarsals. These occur where bones are repeatedly exposed to damaging forces without enough time for recovery. Once a stress fracture arises, the most effective management strategy is rest, often leading to a 6-12 week period before training can resume. This delay severely reduces the chances of military recruits completing their training, and thus costs the ministry of defence in terms of money and soldier numbers. Third metatarsal (the long bone in your foot in line with your third toe) stress fractures are more common amongst military recruits than other populations, despite often similar workloads. It is suggested that the standard issue combat assault

boot (CAB) may be the reason for the unusually high incidence of 3rd metatarsal stress fractures in military recruits.

There is a lack of research into 3rd metatarsal stress fractures, with this site often overlooked in favour of the more common 2nd metatarsal stress fracture. There is even less research that considers the size and shape of the bone, or an attempt to calculate the force acting upon it during running.

The aim of this investigation

The aim of the investigation is to use magnetic resonance imaging (MRI) to collect individual bone geometry and motion data with a view to developing a mathematical model of the forces acting upon an individual's 3rd metatarsal during gait. This will be used to compare the effects of the CAB and other footwear types on stresses experienced by the 3rd metatarsal.

What participants are needed?

Ten male volunteers, aged 18-24, with size 11 feet.

What are the procedures?

There are two sections to the testing procedure. First, volunteers will be required to perform 8 short runs of 15m at a controlled speed of 3.8m/s for each of three footwear conditions: the CAB, a cross trainer and the CAB with an insole. In each instance, the volunteers will also be required to wear a pressure insole within their shoe. During each run, the volunteers must land one foot on a force plate set in the laboratory floor.

Participants will be allowed as much tuition and time as is necessary to practice running with the different footwear conditions, at the desired speed, whilst planting their foot on the force plate. This section of the study is anticipated to take approximately 1 hour.

The first part of the testing requires each volunteer to be scanned using MRI. This will take place at the University of Exeter's Peninsula Medical School, on St. Luke's Campus. For this session, participants will be required to sit with their right leg in a confined space for a total of approximately 30-45 minutes. Music may be played to make the volunteer feel more comfortable and refreshments can be supplied as required. Participants may wear clothing of their choice as long as it does not have any metal in it such as zips or buttons.

Exclusion criteria

Participants should not have suffered a significant lower limb injury in the last 3 months prior to testing. A detailed exclusion criteria for MRI testing accompanies this document, however in summary volunteers should not have and metallic objects about their person. This includes metal implants such as plates, pacemakers and fillings.

Information handling

The information attained from the investigation will only be seen by Mr Michael Nunns, Dr Sharon Dixon and MRI staff present at the scanning. The data are to be used in the development of a mathematical model and then statistically analysed and presented in the PhD thesis of Mr Michael Nunns. Findings may also be published, with the data included being kept anonymous. Participants are welcome to attain the results of the study if they so wish.

Mr Michael Nunns and Dr Sharon Dixon are responsible for ensuring that any personal data which they hold are kept securely and are not disclosed either orally or in writing and by accident or otherwise to any unauthorised third party. All data will be destroyed after a period of approximately 4 years following any publications. This may mean that data is stored for up to 10 years.

Throughout the duration of the project you are welcome to contact Mr Michael Nunns with any questions or worries you may have regarding the project. Contact can be made via mobile phone (07779449812) or via email at mn227@ex.ac.uk .

Withdrawal

You will be completely free to end your participation in the project at any time without fear of prejudice.

The Ethics Committee of the School of Sport and Health Sciences have reviewed and approved this project.

Thank you for taking the time to consider this project.

Appendix D(ii): MRI safety form

Participant Safety Checklist

Name:.....
Weight:
.....

Date of Birth:
Study Name/Volunteer Number:

Please check the following list carefully, answering all appropriate questions. Please do not hesitate to ask staff, if you have any queries regarding these questions.

1.Do you have a pacemaker, artificial heart valve or coronary stent?
Yes No

2.Have you ever had major surgery?
Yes No

If yes, please give brief details:

3.Do you have any aneurysm clips (clips put around blood vessels during surgery)?
Yes No

4.Do you have any implants in your body?

Yes No Joint replacements, pins or wires

Yes No Implanted cardioverter defibrillator (ICD)

Yes No Electronic implant or device

Yes No Magnetically-activated implant or device

Yes No Neurostimulation system

Yes No Spinal cord stimulator

Yes No Insulin or infusion pump

Yes No Implanted drug infusion pump

Yes No Internal electrodes or wires

Yes No Bone growth/bone fusion stimulator

Yes No Any type of prosthesis

Yes No Heart valve prosthesis

Yes No Eyelid spring or wire

Yes No Metallic stent, filter or coil

Yes No Shunt (spinal or intraventricular)

Yes No Vascular access port and/or catheter

Yes No Wire mesh implant

Yes No Bone/joint pin, screw, nail, wire, plate etc.

Yes No Other Implant

5. Do you have an artificial limb, calliper or surgical corset?

Yes No

6. Do you have any shrapnel or metal fragments, for example from working in a machine tool shop?

Yes No

7. Do you have a cochlear implant?

Yes No

8. Do you wear dentures, plate or a hearing aid?

Yes No

9. Are you wearing a skin patch (e.g. anti-smoking medication), have any tattoos, body piercing, permanent makeup or coloured contact lenses?

Yes No

10. Are you aware of any metal objects present within or about your body, other than those described above?

Yes No

11. Are you susceptible to claustrophobia?

Yes No

12. Do you suffer from blackout, diabetes, epilepsy or fits?

Yes No

For women:

13. Are you pregnant or experiencing a late menstrual period?

Yes No

14. Do you have an intra-uterine contraceptive device fitted?

Yes No

15. Are you taking any type of fertility medication or having fertility treatment?

Yes No

Important Instructions

Remove all metallic objects before entering the scanner room including hearing aids, mobile phones, keys, glasses, hair pins, jewellery, watches, safety pins, paperclips, credit cards, magnetic strip cards, coins, pens, pocket knives, nail clippers, steel-toed boots/shoes and all tools. Loose metallic objects are especially prohibited within the MR environment.

I have understood the above questions and have marked the answers correctly.

Signature
(Participant/Parent/Guardian)

Date

MR Centre Staff Signature

Appendix D(iii): Study 3 information sheet

Title of Study: Identification of Biomechanical Predictors of Lower Limb Stress Fracture Susceptibility

We would like to invite you to participate in this research project being undertaken by the **University of Exeter** and the **Institute of Naval Medicine (INM)**. You should only participate if you want to; choosing not to take part will not disadvantage you in any way. Before you decide whether you want to take part, it is important for you to understand why the research is being undertaken and what your participation will involve. Please take time to read the following information carefully and discuss it with others if you wish. Ask the project team if there is anything that is not clear or if you would like more information. If you would like to take part, please inform the project team know if you have been involved in any other study during the last year.

The purpose of this study is to identify biomechanical (anatomical and movement) variables associated with the development of lower limb stress fracture in RM recruits, which in turn will allow the identification of interventions for reducing stress fracture incidence during RM training.

Prior to the study, you will undergo medical screening by trained personnel, including the completion of a health history questionnaire.

The study involves the measurement of anthropometric (body mass, height, and lower limb – calf – girth) and anatomical (ankle dorsi-flexion and hip range of motion) variables, as well as undertaking synchronised movement (CodaMotion, Charnwood Dynamics, UK) and pressure (RSscan International, Belgium) measures during barefoot running.

An independent medical officer will attend study briefings with participants and will be on call in the **Medical Centre** throughout all parts of the study. Her sole function is to act independently of the study team to ensure your safety and well-being. She may terminate your participation in the trial on medical grounds at any time, and you may consult with her at any time.

You may, at any time, withdraw from the experiment without giving a reason. If you ever require any further explanation, please do not hesitate to ask.

Any information obtained during this trial will remain confidential as to your identity: if it can be specifically identified with you, your permission will be sought in writing before it will be published. Other material, which cannot be identified with you, will be published or presented at meetings with the aim of benefiting others. You have a right to obtain copies of all papers, reports, transcripts, summaries and other material so published or presented on request to the Project Officer. All information will be subject to the current conditions of the Data Protection Act 1998.

In the event of you suffering any adverse effects as a consequence of your participation in this study, you will be eligible to apply for compensation under the MoD's 'No Fault Compensation Scheme' (see separate sheets for details).

Experimental records, including paper records and computer files, will be held for a minimum of 100 years in conditions appropriate for the storage of personal information. You have right of access to your records at any time.

A full scientific protocol for this research has been approved by the Ministry of Defence Research Ethics Committee. This study complies and at all times will comply with the Declaration of Helsinki¹ as adopted at the 52nd WMA General Assembly, Edinburgh, October 2000 and with the Additional Protocol to the Convention on Human Rights and Biomedicine, concerning Biomedical Research, (Strasbourg 25.1.2005). Please ask the Project Officer if you would like further details of the approval or to see a copy of the full protocol.

Name and contact details of Independent Medical Officer:

Dr Caroline Hunter (Civilian Medical Practitioner, CTC RM),
Medical Centre, Commando Training Centre Royal Marines, Lymington,
Exmouth,
Devon. EX8 5AR

Telephone: Mil. 93785 4123 Civ. (01392) 414123

Name and contact details of Principal Investigator:

Dr Sharon Jane Dixon (Senior Lecturer)

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Name and contact details of MoD/RN Liaison (Project Team Member):

Dr Joanne L. Fallowfield (Head of Applied Physiology)

Environmental Medicine Unit, Institute of Naval Medicine, Crescent Road,
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¹ World Medical Association (2000) Declaration of Helsinki. Ethical principles for medical research involving human subjects. 52nd World Medical Association General Assembly, Edinburgh, Scotland October 2000.

Appendix D(iv): Study 3 consent form

Title of Study: Identification of Biomechanical Predictors of Lower Limb Stress Fracture Susceptibility

Ministry of Defence Research Ethics Committee Reference:

- The nature, aims and risks of the research have been explained to me. I have read and understood the *Participant Information Sheet* and understand what is expected of me. All my questions have been fully answered to my satisfaction.
- I understand that if I decide at any time during the research that I no longer wish to participate in this project, I can notify the researchers involved and be withdrawn from it immediately without having to give a reason for my withdrawal. I also understand that I may be withdrawn from the study at any time, and that in neither case will this be held against me in subsequent dealings with the Ministry of Defence.
- I understand that the screening process to decide if I am suitable to be selected as a research participant may include completing a medical screening questionnaire and/or a physical examination by a medical officer and I consent to this.
- I consent to the processing of my personal information for the purposes of this research study. I understand that such information will be treated as strictly confidential and handled in accordance with the provisions of the Data Protection Act 1998.
- I agree to volunteer as a research participant for the study described in the information sheet and I give full consent to my participation in this study.
- This consent is specific to the particular experiment described in the *Participant Information Sheet* (attached) and shall not be taken to imply my consent to participate in any subsequent study/experiment or deviation from that detailed here.
- I understand that in the event of my sustaining injury, illness or death as a result of participating as a volunteer in Ministry of Defence research, I or my dependants may enter a claim with the Ministry of Defence for compensation under the provisions of the no-fault compensation scheme, details of which are attached.

Participant's Statement:

I

agree that the research project named above has been explained to me to my satisfaction and I agree to take part in the study. I have read both the notes written above and the *Participant Information Sheet* about the project, and understand what the research study involves.

Signed

Date

Witness

Name

Signature

Investigator's Statement:

I

confirm that I have carefully explained the nature, demands and any foreseeable risks (where applicable) of the proposed research to the Participant.

Signed

Date

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