<u>The kinetics, kinematics and energy requirements of distance running: Implications for footwear design.</u>

by

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A thesis submitted in partial fulfilment for the requirements for the degree of Doctor of Philosophy at the University of Central Lancashire.

Volume 1 of 1

University of Central Lancashire
December 2012

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In memeory of Donald, Margaret Sinclair and Kenneth Winter R.I.P.

Abstract

As the primary interface between the runner and the road, athletic footwear has a potentially important role to play in the reduction/management of chronic injuries and the enhancement of performance. Over the past thirty years the running shoe has changed considerably. However, despite significant advances in footwear technology, the incidence of injury in distance runners has not altered meaningfully. It has been postulated that poor footwear selection is the mechanism behind this; whereby running shoes are incorrectly selected/inappropriate for the populations and situations in which they are worn.

The investigations and results obtained from this thesis aim to attenuate this, and provide runners and footwear manufacturers with new knowledge regarding the application of footwear to different populations and conditions in order to improve both injury occurrence and performance. In addition to a significant amount of developmental analyses, four principal examinations were conducted as part of this thesis.

Study 1 aimed to determine the kinetic and 3-D kinematic differences between treadmill and overground running, in order to determine whether the treadmill replicates overground running and whether different footwear is necessary during treadmill running. It was observed specifically that treadmill running was associated with significant increases in eversion and tibial internal rotation whilst overground runners exhibited greater peak tibial accelerations. It was concluded that treadmill runners are likely to require footwear with additional medial stability properties, aimed at reducing

rearfoot eversion whilst overground runners should consider footwear with more advanced midsole cushioning properties designed to reduce the magnitude of impact transients

Study 2 examined the gender differences in the kinetics and 3-D kinematics of running in order to determine whether females require running shoes specifically tailored to their running mechanics. Females were associated with significant increases in eversion and tibial internal rotation; reaffirming the notion that they are more susceptible to overuse injuries than males. It is recommended that females select running footwear with design characteristics aimed towards the reduction of rearfoot eversion in order to reduce the incidence of injury.

Study 3 investigated the kinetics and 3-D kinematics of running: barefoot, in conventional running shoes and in barefoot inspired footwear in order to determine the efficacy of barefoot running in comparison to shod and also the ability of barefoot inspired footwear to closely mimic the 3-D kinematics of barefoot running. Barefoot running was associated with significant increases in impact parameters. It was also observed that barefoot inspired footwear does not closely mimic the 3-D kinematics of barefoot running. This leads to the final conclusion that barefoot running may not serve to reduce the incidence of injury.

Study 4 aimed to examine the influence of footwear with different shock attenuating properties on the energy requirements of distance running and to investigate the biomechanical parameters which have the strongest association with running economy using regression analyses. Whilst footwear with different shock attenuating properties did not influence running economy, it was observed that a significant proportion of the variance in running economy could be explained by kinematic and EMG parameters.

This thesis has provided information not previously available regarding the injury prevention and performance aspects of running footwear. It is clear that footwear cannot be universally prescribed and that the population and circumstances in which different shoes are to be used are key when selecting and designing appropriate running footwear.

Acknowledgements

I have been extremely fortunate throughout to benefit from excellent supervision throughout the journey towards submission.

Firstly I would like to thank my director of studies Dr Sarah Jane Hobbs; she has been the best supervisor I or anyone else could have wished for. Through our numerous and almost daily chats I learned so much from her as a mentor and in the process gained a great friend. I will forever be indebted to her. I would also like to thank the other members of my supervisory Dr Chris Edmundson and Dr Darrell Brooks to whom I am also grateful to for their constant assistance with all aspects of the thesis.

In addition to the academic support that I have received, the support and motivation that I have received from family and friends has also been a significant source of motivation. I would like to thank my mum, dad and Andrew for their encouragement throughout. I would also like to thank Dr Lindsay Bottoms, Dr Paul Taylor, Dr Andy Greenhalgh, Dr Howard Hurst and Dr Laurence Protheroe for their encouragement and support.

Finally I would like to thank my fantastically understanding girlfriend Sally Andrews for her support, encouragement and for putting up with me whilst I completed my thesis.

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2. Sinclair J, Hobbs, SJ, Protheroe L and Greenhalgh A. (2010). Utilization of an externally mounted accelerometer to determine gait events. Presented at the Bases annual conference, University of Aberystwyth, 31st March-1st April.
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Abbreviations and definitions

2-D – Two dimensional

3-D – Three dimensional

6 DOF – Six degrees of freedom

ANOVA – Analysis of variance

ASIS – Anterior super iliac spine

BF – Biceps femoris

Cardan/Euler angles - Means of representing the spatial orientation of any frame (coordinate system) as a composition of rotations from a frame of reference (coordinate system)

CAST – Calibrated anatomical systems technique. 3-D modelling technique allowing joint rotations to be measured.

.C3D – Document format specific to C-Motion (Visual 3-D)

CNS - central nervous system

COM – Centre of mass.

CO_2 – Volume of expired carbon dioxide

EMG – Electromyography, the measurement of muscular electrical activity

EV/TIR – Eversion/tibial internal ratio. Range of motion of eversion from footstrike to peak angle divided by range of motion of tibial internal rotation from footstrike to peak angle.

FFT – Fast Fourier transform

Gimbal lock –The loss of one degree of freedom in a three-dimensional space that occurs when the axes of two of the three gimbals are driven into a parallel configuration.

GA - Gastrocnemius

GRF – Ground reaction force

Helical angle – Means of quantifying rotation whereby movement from a reference position is described in terms of rotation along a single projected axis

Hz – Hertz, a measure of frequency the number of times an event occurs per second

ISB – International Society of Biomechanics

Kn – Kilonewtons

LCS – Local co-ordniate system

MANOVA – Multivariate analysis of variance

Mmol.L – Millimoles per litre

ms – Milliseconds

$m.s^{-1}$ – Metres per second

mV – Millivolts

MVC – Maximal voluntary contraction

VO_2 – Volume of inspired oxygen

POSE – Position and orientation

PSIS – Posterior super iliac spine

QTM – Qualysis track manager

ROM – Range of angular motion from footstrike to toe-off

Relative ROM (Range of motion) – Range of angular motion from footstrike to peak angle

SCS – Segment coordinate system

SENIAM - Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles

Ag/AgCI - Silver/silver chloride

TA – Tibialis anterior

VO_2 – Volume of inspired oxygen

VL –Vastus lateralis

VM – Vastus medialis

1. Introduction

This chapter provides a brief introduction to the project and rationale for the investigations, conducted as part of this thesis.

1.1 Background

In recent years distance running has become a popular physical/recreational activity (De Wit et al. 2000). The running boom which originated in the late 1970's has led distance running and the athletic footwear industry to become what it is today. Running represents a convenient low cost pass-time which contributes to a decreased mortality rate, as well as a reduction in morbidity and the development of disabilities in older adults (Curfman, 1993; Fries et al. 1994; Paffenbarger et al. 1993).

Unfortunately however, epidemiological studies analyzing the prevalence of running injuries suggest that overuse injuries are a prominent complaint for both recreational and competitive runners (Hreljac, 2004). Each year approximately 66% of runners will experience a pathology related to running (Malone, 2008). These chronic injuries serve not only to affect training, but can also instigate lasting physical and psychological effects (Shorten, 2000). These effects may make running considerably less enjoyable and consequently act as a deterrent to running training in general. Given the detrimental impact and high incidence of injuries, combined with the positive health benefits of running, research that could reduce the incidence of injury through footwear interventions could be beneficial to a large number of runners.

Typical distance running speeds may result in as many as 300 foot strikes per leg per kilometre, often on hard surfaces (Valiant, 1990). The interaction between the foot and ground during each foot strike running gait induces high impact forces, generating a transient shock wave that is transmitted through the musculoskeletal system (Shorten, 2000). Repeated impact loading of the body during gait has been linked to the development of degenerative pathologies.

In 1975 runner's world magazine referred to the running shoe as the most important piece of athletic equipment that a runner may own. Nigg (1986) attributes this article to the improved interest in the protective and performance benefits of athletic footwear in both runners and sports manufacturers. As the primary interface between the runner and the road, athletic footwear has a potentially significant role to play in reduction/management of chronic injuries and the enhancement of performance. Over the past thirty years since the origin of the running boom, the running shoe has changed significantly although its application has not. The development of running shoes has shadowed the rise of running as a popular activity. The functional role of the running shoe is to enhance performance and prevent injury to the athlete (Frederick, 1984). Footwear biomechanics is a swiftly progressing field that helps to understand the influence of running shoes and their design on injury prevention and performance. Research in the fields of orthopaedic medicine and biomechanics has focussed on analysis of the function of footwear (Nigg, 1986).

However despite significant advances in footwear technology, the overall rate of injury in distance runners has not altered significantly. Noting that running injury rates have remained unchanged over the last 30 years led researchers to investigate other paradigms such as barefoot running. These hypotheses however remain unproven at the current time. It has also been postulated that poor footwear selection is the mechanism behind the lack of improvement in injury frequency (Nigg, 2010); whereby running shoes are incorrectly selected/inappropriate for the populations and situations in which they are worn. There is currently a paucity of research examining the applicability of footwear to specific conditions which may partially explain why injury rates remain high despite the increase in knowledge and subsequent development of footwear technology.

The investigations and results conducted during this thesis aim to attenuate this, and provide runners and footwear manufacturers with new knowledge regarding the application of footwear to different populations and conditions in order to improve both injury occurrence and performance.

1.2 Aims and objectives

The overall aims of the study were to examine the efficacy and applicability of different footwear designs to different subject groups and conditions and also to examine how different footwear properties affect performance. The thesis reviews, implements and adapts previous techniques in order to identify kinetic and kinematic variables linked to overuse injuries and performance. The objectives of the thesis were:

- To determine the most effective cardan sequences for the quantification of three dimensional (3-D) kinematics of the lower extremities.

- To determine the reliability of 3-D kinematic techniques.

- To examine different methods of quantifying gait events in the absence of force platforms.

- To examine the kinetic and 3-D kinematic differences between treadmill and overground running, in order to determine whether a. the treadmill replicates overground running and b. whether different footwear should be worn during treadmill running.

- To examine the gender differences in the kinetics and 3-D kinematics of running in order to determine whether females require running shoes specifically tailored to their running mechanics.

- To investigate the kinetics and 3-D kinematics of running: barefoot, in conventional running shoes and in barefoot inspired footwear in order to determine a. the efficacy of barefoot running of in terms of its ability of reduce the proposed mechanisms of injury and b. whether shoes which aim to simulate barefoot movement patterns can closely mimic the 3-D kinematics of barefoot running.

- To determine the most effective EMG normalization technique for the muscles of the lower extremities during running.

- To examine the influence of footwear with different shock attenuating properties on the energy requirements of distance running and to examine the kinetic and 3-D kinematic parameters which have the strongest association with running economy using regression analyses.

1.3 Rationale

The treadmill is now commonly used for exercise and research, and it offers an appealing well standardized method for running gait analysis (Schache et al. 2001). The mechanics of treadmill running have been studied extensively, although a consensus has not been reached regarding its ability to simulate overground locomotion (Fellin et al. 2010a). An integrated approach that simultaneously measures the kinetics and 3-D kinematics of running is needed to evaluate and compare running overground ground and on the treadmill. Clinically this knowledge is important to gain an understanding of the susceptibility of treadmill runners to overuse injuries. If significant variations in axial impact shock and 3-D kinematics exist between treadmill and overground locomotion, there may be implications regarding the relative susceptibility of runners to lower extremity overuse injuries which would have implications for runners as it is currently unknown as to whether different footwear is necessary for treadmill running, thus runners typically use the same shoes as they would when running overground.

Interest in distance running amongst females has expanded rapidly in recent years (Atwater, 1990). Whilst studies have investigated differences between males and females (Atwater, 1990; Heiderscheit et al. 2000), the majority have been confined to the sagittal plane and kinematic analyses have predominantly used two-dimensional (2-D) video analysis as opposed to more accurate 3-D optoelectronic techniques. Females are hypothesized to be more susceptible to overuse injuries (Taunton et al. 2003). Given the potential susceptibility of females to overuse running injuries, a key issue within the discipline of footwear biomechanics that has yet to be addressed is the specific demands of athletic footwear for females. Footwear manufacturers frequently produce footwear for females on the basis of data collected using male participants. This has led to women's running shoes being habitually designed using a scaled down version of a man's shoe with all dimensions reduced proportionally according to the length of the foot. A greater understanding of the differences in running mechanics between male and female runners may also provide an insight into the aetiology of different injury patterns and how these injuries may be attenuated using appropriate footwear designs.

In recent years the concept of barefoot running has garnered much attention in footwear biomechanics literature. Barefoot locomotion presents a paradox in footwear literature; and has been used for many years both by coaches and athletes based around the supposition that running shoes are associated with an increased incidence of running injuries (Lieberman et al. 2010; Robbins and Hanna, 1987; Warburton, 2000). Based on such research and taking into account the barefoot movement's recent rise in popularity, barefoot inspired shoes have been designed in an attempt to transfer the perceived advantages of barefoot movement into a shod condition. Given the popularity of barefoot running, surprisingly few investigations have specifically examined both the impact kinetics and 3-D kinematics of the lower extremities of running barefoot and in barefoot inspired footwear in comparison to shod. Therefore a comparative investigation into the kinetics and 3-D kinematics of running: barefoot, in conventional running shoes and in barefoot inspired footwear, in order to highlight the differences among conditions would be of clinical and practical significance.

The economy of running which is a reflection of the amount of oxygen (VO_2) required to maintain a given velocity, is considered a very important factor for the determination of distance running performance. Given the influence of running economy on running performance, a number of investigations have aimed to determine the biomechanical parameters that may be related to running economy. However the observed connections are often weak and have lacked consistency between investigations. The possibility that athletic footwear can influence the economy of distance running has also been examined previously (Frederick, 1986). A number of studies allude to the assumption that it is more energetically economical to run in footwear with appropriate mechanical characteristics. Different footwear shock attenuating properties have the potential to reduce impact forces, alter running kinematics, change muscle activity, and thus potentially influence running economy (Nigg, 2003). Several investigations hypothesize that footwear midsole characteristics can influence energy expenditure (Frederick et al. 1986; Squadrone et al. 2009; Hanson et al. 2010). However, to date there have been no investigations which have examined footwear with known differences in shock attenuating properties and related their energy requirements to simultaneous measurements of 3-D kinematics, impact kinetics and muscular activation parameters.

1.4 Ethical approval

Ethical approval for all of the investigations carried out throughout this thesis was provided by an ethical panel from the School of Psychology, at the University of Central Lancashire.

1.5 Thesis Structure

This project consists of nine chapters. Following this introductory chapter is a review of relevant literature concerning the kinetics, kinematics, metabolic aspects of distance running.

Chapters 3 and 4 provide information regarding the methods and equipment used throughout the thesis and the developmental work undertaken to refine and validate these techniques Chapters 5-8 comprise the main research components of the thesis, with each one devoted to each one of the four primary research areas. Each chapter consists of an introduction to the specific study, presentation of specific pilot data, methodology, detailed description of the results, and discussion of these findings. Chapter 9 collates and summarizes the major findings and examines their implications along with recommendations for future research and development.

2. Literature Review

This chapter reviews the literature relevant to the kinetics and kinematics of running as well as literature related to the principles of different biomechanical measurement techniques.

2.1 Kinematic measurement

The most common technique for the collection of kinematic information uses an image or motion capture based system to record the movements of markers placed onto a moving subject (Robertson et al. 2005). This is followed by either manual or automatic digitizing in order to determine the co-ordinates of the markers. Motion capture techniques used to record movement are usually grouped into 2-D or 3-D systems. The technique employed is governed by the requirement of the researchers and the necessary depth of analysis. 2-D motion capture requires only a single video camera orientated orthogonally with the rotational plane of interest, whereas 3-D systems are comprised of a number of cameras (usually between 6-10).

Due to factors such as price and availability it has been commonplace for researchers investigating human movement to use standard 2-D video analysis with electronically shuttered video cameras (Bartlett et al. 1992; Nigg and Herzog 2005). However, video cameras are limited by their capture frequencies as they typically operate at only 25 frames per second providing a maximum sampling rate of 50Hz when using intertwined fields. Based on the Niquist Theory of sampling these frame rates are not adequate for the quantification of faster movements (Robertson et al. 2005).

Although considerably more expensive (individual cameras can exceed £10,000) optoelectric 3-D camera configurations are considered to be advantageous as they automatically digitize the markers and are not affected by perspective and planar errors, as 2-D systems are (Richards et al. 2008). 3-D camera systems are used to track either passive retroreflective markers or active infra-red light emitting passive markers; positioned onto the participant whose motion is being examined. Passive marker systems involve the camera lens emitting infrared light to illuminate the retroreflective markers at a predetermined frequency, allowing them to be identified using specialized motion capture software. For active marker recognition, markers produce their own light and thus do not require infra-red lighting from the camera, thus the markers are identified more easily (Richards et al. 2008). 3-D motion capture systems also have the advantage of being able to capture human movement at higher frame rates (sometimes up to 1000Hz), which has made them commonplace in most specialist biomechanics laboratories.

The quantification of 3-D musculoskeletal movement via image based motion capture is conducted using a model consisting of a linked kinematic chain of segments constructed from the markers positioned on the body (Robertson et al. 2005). The objective of segmental mechanics is to obtain quantitative information that facilitates the instantaneous reconstruction of a rigid body in space (Cappozzo et al. 2005). To model the motion of each segment in 3-D two principal pieces of information are needed, the position and the orientation i.e. POSE and segment anatomy. Segments are considered to be rigid bodies for the purpose of 3-D reconstruction. The motion of rigid segments in space can be quantified using three orthogonal translational degrees of freedom and three orthogonal rotational degrees of freedom (Cappozzo et al. 1995). The quantification of the rigid-body POSE requires the determination of an anatomical frame relative to the global co-ordinate system (GCS). Within this anatomical frame is a technical frame, defined by markers that are used to track the motion of the segment. These markers may be anatomical or arbitrary, individual or clusters and mounted on the skin, on wands or on rigid plates (Manal et al. 2000).

There are a number of methods available for the computation of 3-D kinematics. Segmental kinematics reflect the 3-D angular orientation of one segment with respect to another and at least three non-colinear markers are required to track the segment POSE. The first method of quantifying the orientation of one co-ordinate system relative to another is based on the finite helical axis proposed by Woltring et al. (1985). This method involves the definition of a position and orientation vector. Movement from a reference position is defined in terms of a rotation and translation about a single axis in space. This method is rarely used however as the neutral reference position is difficult to precisely define. Grood and Suntay (1983) proposed the first method using the cardan/euler convention, known as the joint co-ordinate system (JCS). The JCS utilizes one co-ordinate axis from each local co-ordinate system (LCS) of the segments that define the joint (Grood and Suntay, 1983). Three rotation axes are defined, firstly about the proximal segment in the medio-lateral (sagittal plane) axis, then about a floating axis in the anterior-posterior (coronal plane) direction and finally about the longitudinal axis of the distal segment producing internal-external (transverse plane) rotation angles (Grood and Suntay, 1983). This method can however be subject to gimbal lock when angles approach 90°.

The final method of quantifying joint motion was developed by Cole et al. (1993) and also based on the cardan/euler convention. This method involves the definition of three angular parameters that specify the orientation of a body with respect to reference axes that would place the segment in the same position as the true movement (Hamill and Selbie, 2004). This method is also subject to gimbal lock and is dependent on the rotation sequence selected (Cappozzo et al. 2005).

2.2 Ground reaction forces in distance running

The ground reaction force (GRF) reflects the force exerted by the ground to a body in contact with it. The GRF is equal in magnitude and opposite in direction to the force that the body exerts on the supporting surface through the foot (Miller, 1990).

2.2.1 Force platforms

Force platforms have been used in biomechanical research for a number of years to quantify the external forces during human locomotion (Nigg and Herzog, 2005). Force platforms are considered to be a fundamental component of any biomechanics laboratory. Force platforms come under two categories, either strain gauge or piezoelectric. Strain gauge force platforms centre around the principle, that as a force is applied to a structure, the length of that structure will change (Thewlis, 2008). Strain reflects the change ratio between the original dimensions and the newly deformed dimensions. Strain gauge platforms contain materials that when distorted will produce a resistance, which when amplified can be related to force from the deformation in a known direction. Piezoelectric force platforms are considered to be superior as they are more sensitive and allow a greater range of measurements. Piezoelectric force platforms utilize piezoelectric crystals (usually quartz) to measure GRF. The principle behind piezoelectric force platforms is the same as strain gauge technology in that they convert deformation into force outputs. When piezoelectric crystals are subjected to deformation they generate an electric current which is proportional to the amount of deformation.

Given that force transducers regardless of their mechanism of measurement do not directly measure force, they must therefore be calibrated. Force platforms are calibrated prior to installation by the manufacturer.

2.2.2 Ground reaction force components

It is commonplace to de-compose the resultant GRF into three components, vertical, anterior-posterior and medial-lateral. The magnitude of all three GRF components is dependent on running speed, although events in the GRF curves occur at the same relative time. Thus during the process of data collection it is necessary to accurately record the running speed of each trial and to cross compare studies only when running speeds are similar (Hamill et al. 1983). Of the 3x GRF vectors the vertical component, has received by far the most attention. Its magnitude is such that the vertical component of the GRF generally dominates the force-time curve compared to the anterior-posterior or medial-lateral components (Keller et al. 1996). Thus the force time curve for this component is far more straightforward and therefore easier to quantify to allow for comparisons between studies (Miller, 1990).

Of the many external factors which can influence the GRF, running velocity has been the focus of a large number of published studies. The consistent finding amongst these studies is that the magnitude of the vertical component increases with enhanced running velocity. Munro et al. (1987), attempting to ascertain reference standards for GRF data and its relationship with running speed, collected GRF's at running speeds ranging from 3.0-5.0m.s^{-1}. The impact peak increased linearly from about 1.6 to 2.3 body weights (B.W's) over the range of speeds.

A double peaked vertical GRF configuration is traditionally characteristic of runners who exhibit a heel strike (Cavanagh and Lafortune, 1980). The first peak referred to as the impact peak is of high frequency and occurs very early in the stance phase; typically recorded values are between 6-17% of total stance time Munro et al. (1987). The second peak referred to as the active peak or thrust maximum is of lower frequency and is seen between 35-65% stance time Munro et al. (1987). Cavanagh and Lafortune, (1980) showed that the impact peak reached about 2.2 B.W. before rising slowly to an active peak of 2.8 B.W. Significantly Cavanagh and Lafortune, (1980) showed that the midfoot group did not exhibit the high frequency impact peak.

The anterior-posterior forces are separated into braking and propulsive phases (Munro et al. 1987). During the braking phase, the force opposes forward motion whilst in the latter propulsive phase it is directed in a forward motion (Miller, 1990). The force-time curve for the anterior-posterior component is relatively consistent, although the pattern of the initial braking phase can vary from runner to runner (Cavanagh and Lafortune, 1980). The anterior-posterior component comprises a small amount of the total GRF in relation to the vertical vector. Most studies show linear increases with running velocity. Nilsson and Thorstensson, (1989) found that the anterior-posterior force vector was 0.13 B.W at 1.5 $m.s^{-1}$ increasing to 0.5 B.W at 6.0 $m.s^{-1}$. Similarly Munro et al. (1987) found that the anterior-posterior force was 0.15 B.W at 3.0 $m.s^{-1}$ and 0.25 B.W at 5.0$m.s^{-1}$.

Running literature is unclear as to whether the propulsive and braking force peaks are equal or whether one exceeds the other. Miller (1990) suggests that this may depend on whether runners are accelerating or decelerating when in contact with the force platform. The point during the force-time history at which maximum braking/propulsive forces occur varies. For individuals who exhibit a single braking peak the maximum braking force occurs at around 22-24% of total stance time (Hamill et al. 1983). The point at which the transition between braking and propulsion occurs is termed the zero fore-aft shear or mid-stance (Miller, 1990). Cavanagh and Lafortune, (1980) noted that the transition from braking to propulsion occurred at approximately 48% of total stance time at a running velocity of 4.5 $m.s^{-1}$.

A number of studies have shown that both mid and forefoot striking runners exhibit a double peaked braking phase, whilst rearfoot strikers show a single peaked braking phase (Nilsson Thorstensson 1989). Hamill et al. (1983) however report that the braking force exhibited by rearfoot strikers is characterized by two peaks whereas that of midfoot strikers was single peaked. In contrast to all of the above studies Munro et al. (1987) observed single, double and multiple braking patterns in a, group comprised of rearfoot strikers. Thus it does not appear that a discernible pattern exists based on footstrike classification.

The medial-lateral component of the GRF has proved the most difficult of the three in which to identify readily quantifiable characteristics (Miller, 1990). The medial-lateral vector is by far the smallest of the three vectors of the GRF, and has shown the highest degree of variability (Cavanagh, 1982). Cavanagh and Lafortune, (1980) found peak to peak forces of 0.12 B.W, whereas Hamill et al. (1983) reported maximum medial and lateral forces of 0.15 B.W. These studies examined participants running at comparable velocities. Munro et al. (1987) found that values ranged from 0.04-0.25 B.W for the medial component and from 0.06-0.31 B.W for the lateral component.

Nilsson and Thorstensson, (1989) demonstrated that runners classified as forefoot strikers initially exhibited a medial force in the early stance phase whilst rearfoot strikers exhibit a lateral force in the early stance phase. Cavanagh and Lafortune, (1980) demonstrated that both rear and midfoot strikers exhibited lateral forces in early and mid-stance, whilst also exhibiting two medial peaks, one in the early stages of the stance phase and again immediately prior to toe off. Thus, similar to the anterior/posterior component no pattern exists based on footstrike.

2.3 Impact loading of the lower extremities

The termination of the swing phase of gait i.e. the foot striking the ground produces compressive loading of the lower limbs (Lake, 2002). One of the characteristics of the foot impacting the surface is a rapid change in momentum in which the foot is brought to rest (Whittle, 1999). This impact leads to the initiation of a transient shock through the body and carries with it the potential for injury (Shorten, 2000).

The transient shockwave is distinguishable as the short impact peak of force on the upslope of the vertical ground reaction force, in the phase immediately preceding initial contact (Collins and Whittle 1989). Given that force is proportional to the rate of change in momentum of the foot, the magnitude of the impact transmission is thus determined by the rate at which the momentum changes (Whittle, 1999). Thus, significantly Whittle, (1999) concluded that the impact shock magnitude is dependent on two key factors: the quantity of the change in momentum and the duration over which the change in momentum occurs. The velocity and mass of the foot in motion that generates the transient shockwave has a significant influence on the total change in momentum (Whittle, 1999). In addition, the duration over which the foot comes to a halt is a reflection of the compressibility of the material below the calcaneus, such as the plantar fat tissue, shoe midsole cushioning properties as well as the stiffness of the ground itself (Kim et al. 1993). By increasing the distance over which the calcaneus is decelerated, the duration over which it comes to a halt is increased and thus the change in momentum associated with the generation of the transient shock wave is reduced (Shorten, 2000). In summary, the forces imposed on the body from the foot-ground interface are dependent on both the mass and velocity of the foot as well as the thickness, elasticity and viscoelastic properties of the boundary between foot and ground (Whittle, 1999).

It was hypothesized therefore that the principal factor in determining the magnitude of the impact transient is the depth of the interface between the calcaneus and surface (Whittle, 1999). As the foot makes contact with the ground, it is brought to a halt. The shorter the distance over which the foot is brought to a halt, the shorter the duration in which deceleration takes place and the higher the force which is required to provide the necessary deceleration (Whittle, 1999). Thus the magnitude of the transient shockwave is linked to the incidence of overuse injuries and can be attenuated by increasing the duration over which the foot is decelerated (Garcia et al. 1994).

The acceleration time-curve is the only measure that can give a representative description of the impact shock transmission to the body (Hennig et al. 1993). However intercortical bone acceleration measures at the tibia have revealed high correlations between peak acceleration and GRF parameters (Hennig and Lafortune, 1991). Vertical impact peaks (R=0.7-0.85) and loading rates (R=0.87-0.99) have been shown to be strongly correlated with peak tibial accelerations (Lafortune et al. 1995). Similarly Greenhalgh and Sinclair (2012a) observed using skin mounted transducers that the impact peak, average loading rate and instantaneous loading rate of the vertical GRF were significantly correlated with peak tibial accelerations. From these experiments it was concluded that GRF measures may be adequate for the prediction of bone accelerations (Hennig et al. 1993).

2.4 Tibial accelerations

Transient tibial accelerations that are experienced by the musculoskeletal system during running are typically quantified using accelerometers which are attached to the distal tibia (Nigg and Herzog, 2005). Much like force platforms accelerometers can be either strain gauge or piezoelectric. Strain gauge accelerometers typically consist of four strain sensitive wires that are attached to a cantilevered mass element mounted to a fixed base (Nigg and Herzog, 2005). When the base is accelerated, then the mass element produces a deformation due to its own inertia. The deformation of the mass element produces a change in the strain of the wires which changes their resistance. The result of this is an electric output that is proportional to the acceleration experienced by the device. Piezoelectric accelerometers much like piezoelectric force platforms are based on the same principle as strain gauge accelerometers except that they are based on the acceleration of piezoresistive elements rather than strain gauge elements. Piezoelectric accelerometers typically use solid state transducers which alter their electrical resistance proportionally to the applied acceleration.

Accelerometers can be either uni or tri-axial. A uniaxial accelerometer will measure the acceleration component in a single direction only. Tri-axial accelerometers will measure the acceleration components in 3 orthogonal directions using three uni-axial accelerometer components. The goal of accelerometers (particularly those mounted to the distal tibia during running analyses) is to quantify bone accelerations. This is most accurately accomplished by mounting/attaching the accelerometer directly to the tibia using Hoffman pins/screws and measuring accelerations in the axial direction.

It has been observed by Lafortune and Hennig, (1995) that peak tibial accelerations quantified using skin mounted accelerometers were often twice the magnitude of those measured using the invasive bone technique. This led Nigg and Herzog, (2005) to conclude that invasive measurements will provide a more reliable representation of the acceleration time history as the resonance frequency of the pin-bone mounting is higher. However, whilst bone mounted procedures do give the best representation of bone accelerations, they are invasive and require surgical techniques in order to implement. Therefore, it has become commonplace to utilize skin mounted techniques to quantify tibial accelerations due to their increased applicability.

Previous investigations have therefore found that by securely appending a lightweight yet rigid mounting device to the skin and attaching the accelerometer that tibial impact accelerations can be accurately quantified (Kim and Voloshin, 1993). The potential for resonance can be attenuated via the utilization of skin stretching techniques, lightweight rigidly attached accelerometers and also the application of an appropriate low-pass filter (Nigg and Herzog, 2005; Sinclair et al. 2010). Sinclair et al. (2010) concluded that mounting the device securely to the skin at the anterior medial aspect of the tibia provides minimal resonance and is thus effective for the quantification of tibial accelerations.

Previous investigations have also shown that it is possible to separate the components of the acceleration signal due to tissue resonance and impact using a spectral analysis in the form of a Fast Fourier transform (FFT) (Lafortune et al. 1995). This was an important breakthrough as higher frequency accelerations have been linked to the aetiology of injury (Lafortune et al. 1995). From this the median power frequency of the acceleration signal can be obtained, allowing frequency shift in the signal that have been mediated through extrinsic parameters such as surface or footwear to be quantified and subjected to statistical analyses.

2.5 Proposed link between impact transition magnitude and overuse injuries in runners.

Valiant, (1990) states that typical distance running velocities result in excess of 300 footstrikes per foot per kilometre. Loading is necessary for maintenance of cartilage, bone, and muscle health (Stone, 1988). An optimal loading window for tissue health can be characterized by repeated impacts of certain magnitude, duration, and frequency. Impacts beyond the optimal loading window can lead to the breakdown of body tissue and overuse injuries (Hardin et al. 2003).

Animal experiments have shown that repetitive impulsive loading may be a significant contributing factor to the development of degenerative osteoarthritis (Dekel and Weismann 1978). In one of the first studies analyzing the aetiology of overuse injuries associated with distance running Radin et al. (1972) showed that osteoarthritis develops through micro-fracture of osseous tissue, resultant stiffening of the re-modelled bone, and finally a reduction in the shock absorbing capacity of the joints. Although this has never been fully established in human studies, epidemiological investigations have anecdotally linked the repetitive loading of the lower extremities during running with the progression of osteoarthritis (Valiant, 1990).

Voloshin and Wosk, (1982) found that heel strike produces a skeletal shock wave that propagates from the heel to the head, carrying with it the potential for damage. The authors attached accelerometers to the tibia, femur and the forehead of their subjects. Significantly they found a relationship linking reduced shock attenuating capacity between femur and head to the occurrence of lower back pain. They also found a relationship between reduced shock attenuation and knee joint pathology. The researchers concluded that joint pathology is associated with a reduced shock absorbing capacity of the joint and that the skeletal acceleration preceding heel strike may serve to overload the shock absorber proximal to the pathological joint, resulting in the development of osteoarthritic erosion.

Light et al. (1980) measured the peak tibial shock transmitted to the lower extremities whilst walking, by attaching a lightweight accelerometer to two Kirschner pins drilled directly into the tibia of a participant. They suggested that the acceleration magnitude measured at the tibia may contribute to the development of osteoarthritis. In support of this finding both MacLellan and Vyvyan, (1981) and Voloshin and Wosk, (1981) found that modulating peak tibial accelerations resulted in the elimination of symptoms after a 3-12 month period in subjects with pain beneath the heel or Achilles tendon. Hreljac, (2004) provided a retrorespective examination of the biomechanical and anthropometric variables that may facilitate the development of overuse injuries in runners. Comparisons were made between a group of runners who had previously sustained an overuse injury and a group who had been free of injury throughout their career (Hreljac, 2004). The injury free group exhibited significantly lower values for the vertical impact peak and peak rate of loading. The injury free group also exhibited a reduced peak eversion angle. Hreljac, (2004) suggests that runners who have developed a natural gait pattern that incorporates low impact forces and a moderate rearfoot eversion angle/angular velocity, are at a lower risk of incurring overuse injuries.

Stress fractures represent fatigue fractures of the bone and are common injuries in runners (Matheson et al. 1987). Stress fractures are amongst the most common running injuries, and may account for 6-14% of the injuries typically encountered by runners. The tibia is considered to be likely to be affected, accounting for between 35% and 56% of all stress fracture pathologies in runners (Matheson et al. 1987; Romani et al. 2002). Females are considered to be at a greater risk of developing stress fractures, both Pester and Smith, (1992) and Queen et al. (2007) reported significant increases in tibial stress fractures in comparison to males. The aetiology of tibial stress fractures is considered to be multi-factorial in nature but repetitive impact loading has been linked to their development. Milner et al. (2006) and Davis et al. 2004) found that the occurrence of tibial stress fractures was related to greater instantaneous and average vertical ground reaction force loading rates along with peak tibial acceleration compared to non-pathological controls.

Plantar fasciitis is also considered to be a common overuse foot pathology in runners, with an incidence of around 7.9% (Riddle and Schappert, 2004; Taunton et al. 2003). Although excessive pronation is believed to facilitate increases in the load on the plantar fascia, it has been proposed that external forces may also contribute to overloading of the plantar fascia (Davis et al. 2004). Because plantar fasciitis is caused by overloading of the plantar fascia, it is important to also consider the external loads that are experienced by the foot. Davis et al. (2004) reported that runners with plantar fasciitis were associated with a greater impact peak and loading rate compared with healthy control subjects. Indicating that a plantar fasciitis in runners may be linked with greater vertical ground reaction force parameters.

2.6 Rearfoot motion

During gait the foot makes contact with the ground beneath the body centre of mass (COM) (Shorten, 2000). Therefore the foot makes contact in a supinated position; this supination angle ranges from 5-10º (Edington et al. 1990). Pronation takes place as the foot rotates inwards to make level contact with the ground, reaching its peak angle around mid-stance (Shorten, 2000). Subtalar pronation unlocks the mid-tarsal and other joints in the foot, lowering the medial longitudinal arch and allowing the foot to adapt to the topography of the surface and to attenuate shock (Perry and Lafortune, 1995). Although pronation combines both eversion and abduction of the foot via subtalar rotation and ankle dorsiflexion, the dominant component of pronation is calcaneal eversion, which is traditionally measured by 2-D kinematics (Shorten, 2000). Importantly the subtalar joint is orientated in a way that links pronation to internal tibial rotation via the mitered hinge effect (Czerniecki, 1998, Figure 2.2). The relationship between pronation and internal tibial rotation is hypothesized to significantly contribute to running injuries. Previous analyses suggest that many common running injuries (Achilles tendinitis, patellar tendonitis, iliotibial band syndrome, patellofemoral pain and plantar fasciitis) may all be linked to excessive rearfoot motion (Willems et al. 2004; Lee et al. 2010; Taunton et al. 2003; Duffey et al. 2000).

Patellofemoral syndrome is believed to be associated with the excessive subtalar eversion and associated tibial rotation, causing the patella bone to deviate slightly from its neutral position relative to the femoral condyles. Therefore causing an increase in patella-femoral contact pressure which may in severe cases can lead to cartilage degradation (Shorten, 2000). When eversion about the subtalar joint occurs, the tibia (as described above) is forced to rotate internally. This causes torsion of the patellar tendon, and abnormally strain is placed on the tibia. This leads to the development of micro tears and subsequently inflammation and pain in the patellar tendon itself. Illiotibial band pathology has also been linked to excessive coronal plane ankle eversion which serves to augment tension in the illiotibial band itself which overtime leads to degradation (Noehren et al. 2006)

Additionally pronation has been hypothesized as a mechanism leading to the development of Achilles tendonitis. Excessive eversion leads to the development of micro tears in the tendon which may subsequently cause tissue degeneration (Shorten, 2000). Clement et al. (1983) suggest that internal rotation of the tibia produces torsional stress on the tendon, potentially facilitating vascular damage. These torsional stresses are also hypothesized to be linked to the development of plantar fasciitis injuries in runners (Warren 1990). Excessive pronation is also associated with some types of tibial stress syndrome. Viitasalo and Kvist (1983) suggest that over pronation causes tension in the Achilles tendon to increase, thus overloading the insertion of the soleus muscle where breakdown of the tibial surface may occur.

A potential limitation of the majority of previous research analyzing rearfoot motion is the use of a two-dimensional technique (See Figure 2.1). De Wit et al. (2000) suggest that two-dimensional angular values measured solely in the frontal plane are very sensitive to the alignment angle between the foot and the camera axis. In addition a two-dimensional measurement technique does not allow for calculation of the abduction angle between foot and lower leg, which has been linked to stability associated injuries (Areblad et al. 1990).

Given the tri-planar orientation of the subtalar joint (See figure 2.2), 3-D examinations of rearfoot pronation should become a principle research area in the future.

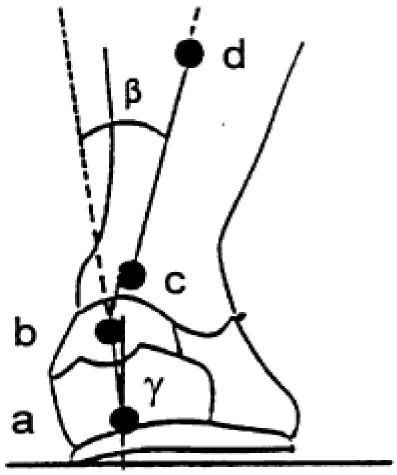

Figure 2.1; Placement of the markers for the quantification of rearfoot motion in the frontal plane. (Taken from De Wit et al. 2000).

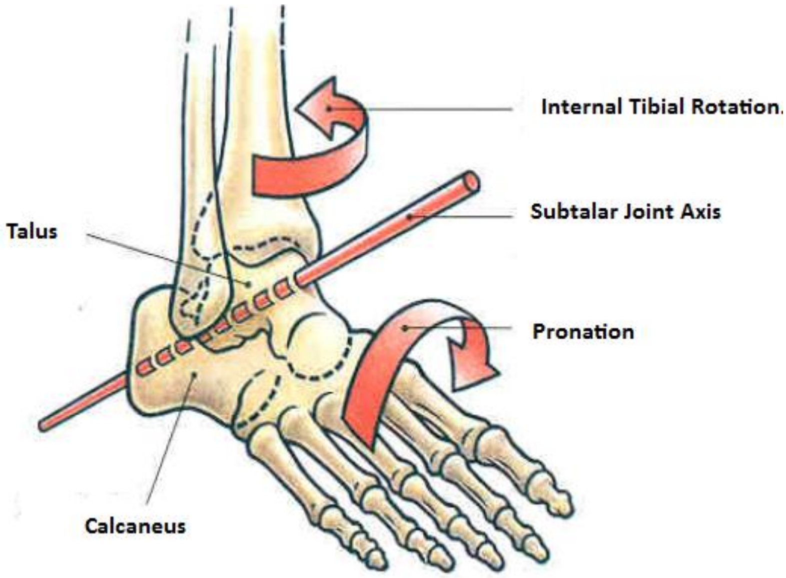

Figure 2.2: The tri-planar nature of the ankle joint complex. Taken from Kirby, (2001).

2.7 Kinematics and shock attenuation

Studies examining lower extremity kinematics have become more prevalent in biomechanics literature in recent years, as research seeks to understand the contribution of lower extremity kinematics to overuse injuries (Butler et al. 2003). These studies have focussed on lower extremity stiffness and its potential contribution to overuse injuries. In its most simple form stiffness refers to the relationship between the deformation of an object and a given force. Impact force, loading rate and stiffness are all correlated (Logan, 2007). Previous studies suggest that there is a correlation between higher loading rates, peak vertical forces, and the associated skeletal shock waves with the prevalence of running overuse injuries (Butler et al. 2003; Logan, 2007).

Experimental evidence of kinematic adaptations to impact is limited (Hardin et al. 2004), this may be because some effective kinematic adaptations are too small to be measured or because of the limited conditions under which adaptations are examined. Increased stiffness is typically associated with reduced lower extremity excursions and increased peak forces. This combination of factors typically leads to increased loading rates, which have been associated with increased shock experienced by the lower extremities (Hennig and Lafortune, 1991). Increased peak forces, loading rates, and shock are all believed to place runners at a greater risk from overuse injuries (Butler et al. 2003; Logan, 2007).

The magnitude of the vertical rate of loading during the stance phase of running can be influenced by the movement strategy that the runner employs (Lafortune et al. 1996a). All factors being equal, collisions that involve greater joint deformations are generally characterized by lower peak forces and rates of loading (Lafortune et al. 1996a). Thus

Bobbert et al. (1992) suggest that lower extremity impacts that involve greater amounts of knee and ankle flexion amplitudes should result in increased shock attenuation.

Ferris et al. (1998) suggest that runners adjust their kinematics inversely to surface stiffness in order to maintain similar loading transmission on different surfaces. It is generally thought that such adaptations mainly involve changes in knee and ankle joint angles (McMahon et al. 1987). Direct evidence of kinematic changes is scarce, however. This may be due to the fact that these adaptations are small (McNair and Marshall, 1994), although potentially important.

At footstrike the knee is not fully extended but exhibits a touchdown angle traditionally around 10-20º (Milliron and Cavanagh, 1990). Knee angle at foot ground contact is believed to have considerable influence on the body's ability to attenuate the impact shock associated with foot strike during locomotion (McMahon et al. 1987). Lafortune et al. (1996a) and Lafortune et al. (1996b) found that larger knee flexion caused significant reductions in vertical rates of loading. McMahon et al. (1987) found that participants, who ran in a groucho running style which is characterized by exaggerated knee flexion, exhibited significantly lower leg stiffness. Kersting et al. (2006) found that every one degree of increased knee flexion at foot contact resulted in a 68N reduction in impact force magnitude.

Following footstrike knee flexion occurs during what is commonly termed the cushioning phase and continues through midstance, this is followed by an extension phase that lasts until toe off (Milliron and Cavanagh, 1990). Knee excursion magnitude from foot contact to maximum flexion is hypothesized to significantly influence impact forces. Increasing the flexion ROM of the cushioning phase has been shown to significantly attenuate vertical GRF parameters (Ferris et al. 1999). Lafortune et al. (1992) attributed reductions in impact parameters to increases in knee excursion. Similarly Butler et al. (2003) demonstrated that decreased knee excursions were associated with increased leg stiffness during running. The angular velocity of this cushioning phase has also been hypothesized as a mechanism which can attenuate leg stiffness. Both Frederick, (1986) and Clarke et al. (1983) demonstrated significant negative correlations between knee flexion angular velocity and vertical GRF parameters. In concurrence, Hartveld and Chockalingam, (2004) demonstrated that when jumpers utilized full knee flexion, significantly lower vertical GRF parameters were recorded.

In running and jumping activities there is widespread verification that landing on the mid/fore foot, i.e. with more plantarflexion of the ankle on contact results in much lower vertical impact peaks and rates of loading (Hartveld and Chockalingam, 2004). Runners who exhibit a natural heel-toe gait pattern that adapt their strike pattern and run landing on the balls of their feet are traditionally subjected to significantly lower rates of loading on impact (Williams et al. 2001). A plantarflexed ankle at initial contact provides a more effective deceleration of the body. Gerritsen et al. (1995) demonstrated that impact forces were significantly influenced by plantarflexion angle at touchdown, to the magnitude of 85N per degree in foot angle. Self and Paine, (2001) studied landings with four different ankle strategies and found that the strategy that incorporated the largest amount of plantarflexion at ground contact was associated with the greatest amount of shock attenuation.

2.8 Gender issues in distance running

Interest in distance running amongst females has expanded rapidly in recent years. This is substantiated by the number of women now participating in distance running training (Nelson et al. 1995). The increase in women's running activities has stimulated many sport scientists to investigate the various aspects of female running performance. The increase in female participation in running has led to a better availability of coaching and training methods, which consequently facilitated the documented improvements in world record performances (Nelson et al. 1995).

There are several notable anatomical/physiological differences between males and females that may influence running biomechanics. The average mature male is greater in both height and mass and has a lower body fat percentage than the average female (Atwater, 1990). In a study providing anatomical reference data, Wilmore (1982) found that males are on average 0.12m taller than females and 18kg heavier, whilst carrying on average 9% less body fat. Increased muscular mass in males is attributable to the higher levels of testosterone, whilst increases in oestrogen contribute to the higher body fat percentage found in females (Wilmore, 1982).

Females are almost twice as likely to sustain a running injury in comparison to their male counterparts (Geraci and Brown, 2005; Taunton et al. 2003). Stress fractures are a fairly common athletic injury that affects a number of runners (Queen et al. 2007). Stress fractures in runners occur under conditions of repetitive cyclic loading of the weight bearing bones (Atwater, 1990). Studies have shown that the bones of the lower leg and foot are the most frequent sites for stress fractures in runners (Gudas, 1980). The tibia is considered to be likely to be affected, accounting for between 35% and 56% of all stress fracture pathologies in runners (Matheson et al. 1987; Romani et al. 2002). Females have been found to be more susceptible to stress fractures. Previous analyses suggest that females are as much as 2-12 times more likely to develop a stress fracture (Pester and Smith, 1992; Queen et al. 2007).

Anatomical factors specific to females have been linked to the development of osteoarthritis. The increased Q-angle is generally believed to predispose female runners to degenerative osteoarthritis at the knee although the exact rationale for this remains undetermined (Atwater, 1990). Gender specific differences in foot shape have also been documented. Anatomical research suggests that females may have greater medial longitudinal arch heights, shorter total foot lengths and smaller instep circumferences (Wunderlich and Cavanagh, 2001). Unger and Rosenbaum, (2004) suggested that these variations in foot geometry can influence the distribution of the load applied to the plantar surface.

Horton and Hall, (1989) dispelled the long held notion that females are associated with a wider pelvis than males. It was observed however that females have a larger hip width to femoral length ratio than do males which it was believed would contribute to a greater adduction of the hip during the stance phase. These structural variances exhibited by females at the hip and knee may influence their movement patterns during running. A limited number of investigations have examined the differences in lower extremity joint kinematic parameters between genders during running. Malinzak et al. (2001) studied frontal and sagittal plane motion of the knee in male and female runners. They reported that, while the frontal plane waveforms were similar, females exhibited significantly more knee abduction throughout the entire stance phase. Females were also were found to exhibit reduced peak knee flexion and also reduced knee flexion ROM in compared to men. Ferber et al. (2003) performed a similar analysis; they also observed that females were associated with increased knee abduction in comparison to males. They also observed significant increases in hip adduction and internal rotation in female runners. The structural combination of increased hip adduction, hip internal rotation, and knee abduction may produce the larger Q-angle that is evidenced in females (Horton and Hall, 1989; Hsu et al. 1990). Increases in the Q-angle have been found to be associated with an increase in patellar contact forces (Mizuno et al. 2001). Thus increased Q-angle magnitude is believed to play role in the incidence of patellofemoral disorders experienced by females (Ferber et al. 2003).

2.9 Running shoe influences

The primary function of athletic footwear as described by (Luethi and Stacoff, 1987) is to provide shock attenuation. Functional footwear is accomplished through appropriate design of the shoe itself and with precisely chosen materials and components. Athletic footwear traditionally comprises an upper, lining section and a sole. Running shoes also traditionally feature a midsole and foot-bed. Various materials and design characteristics are utilized as supportive aspects of the upper and sole. These materials and shoe components contribute to the comfort, fit and overall performance of the shoe.

The properties of athletic footwear have been linked to the prevention of running injuries (Nigg, 2010). Excessive rearfoot motion and impact forces are recognised as factors associated with running injuries. It is assumed that these factors can be influenced by appropriate running shoe construction (Milani et al. 1997). Bates, (1985) stated that the two key functions of running shoes are to provide cushioning and stability. The density of the shoe midsole has a significant influence on both of these functions. There is evidence that amplitude of vertical impact forces in sports involving running accounts for many injuries Andrish, (1985) and Clement et al. (1981). Thus, it seems natural for researchers to attempt to reduce the magnitude of impact forces through footwear design (Robbins et al. 1998). However literature has provided conflicting results regarding the effects of midsole density on both shock attenuation and stability. Some studies have shown that shoes with softer midsoles produce lower vertical force and tibial acceleration parameters (Devita and Bates, 1988; Aerts and De Clercq, 1983).

However, other studies have shown that softer running shoes do not reduce the impact forces associated with footstrike during running. Snel et al. (1985) examined vertical impact forces in runners wearing one of nine pairs of the shoes selected by the authors. It was concluded that no difference in impact force parameters was detectable between different levels of shoe cushioning. Clarke et al. (1982) measured vertical impact forces of participants who ran over a force platform whilst wearing both soft and hard soles. They found that impact forces were not significantly different between conditions, thus they concluded that cushioning did not affect impact parameters. Later, Clarke et al. (1983) examined vertical impact forces in runners who ran over a force platform whilst wearing shoes that represented both extremes of hardness. Once again midsole cushioning failed to reduce impact loading.

Kaelin et al. (1985) investigated vertical impact forces in runners who each ran whilst wearing one of 12 pairs of custom shoes with varying sole thicknesses and midsole hardness's. Once again the conclusion drawn was that differences in cushioning did not influence impact loading. Aguinaldo and Mahar, (2003) evaluated the effects of running shoes with 3 different types of cushioning column systems (Hard, medium and soft) on impact force patterns during running. Participants exhibited significantly lower impact peaks and loading rates when wearing the hard and medium shoes compared to the soft cushioning system.

Some studies have shown that running shoes with a softer midsole may actually produce higher vertical GRF parameters than harder shoes. Nigg et al. (1987) examined impact forces in runners wearing shoes with midsoles that spanned a wide range of hardness. Impact loading was significantly higher when wearing the softest soled shoes. Nigg, (1986) suggested that higher vertical impact forces exhibited whilst running in shoes with the softest midsole materials (Shore A 30 and below) may be attributable to a bottoming out effect. This occurred when the material was too soft, it therefore totally compressed in a relatively short space of time, thus eliciting high force amplitudes.

Clarke et al. (1983) suggested that different findings between in vitro material tests and in vivo participant tests using force platform analyses may also be due to fact that participants alter their running kinematics in response to perceived hardness or softness of the shoe. This adaption would then alter the recorded GRF pattern, consequently hiding any shock attenuating properties that may have been provided by the shoes mechanical characteristics.

It was hypothesized by Frederick, (1984) that runners exhibit a plantar sensory response to compensate for the perceived lack of cushioning whilst running barefoot and in harder shoes. Frederick, (1984) proposed a three component system in which the surface, shoe and body interact. The body was described as a dynamic adapting component of this system. Ignoring any component of the system is an error as it blinds researchers to the crucial role of the kinematic adjustments that the body makes in response to the summary mechanical components of the system (Frederick, 1984). Under load, plantar feedback is important for the perception of impact and neuromuscular adaptations in kinematics. When running barefoot or in harder shoes, previous studies indicate that runners maximise shock attenuation by adopting touchdown kinematics in favour of deceleration (Robbins and Gouw, 1991). Robbins and Gouw further hypothesized that the cushioning properties of modern footwear create a perceptual underestimation of impact severity. Therefore in highly cushioned shoes this plantar sensory mechanism may be significantly reduced, resulting in a sharp reduction in shock absorbing behaviour (Frederick, 1986).

There have been a number of studies analyzing the effect of shoe design parameters on pronation features. Midsole hardness has been shown to significantly influence pronation. The general consensus is that softer midsoles allow more maximum pronation and total rearfoot movement (Clarke et al. 1983; Stacoff et al. 1988). The effect of heel flare has also been discussed in running shoes literature. Clarke et al. (1983) demonstrated that increases in flare angle resulted in a significant reduction in maximum pronation and total rearfoot movement. Although angles in excess of 15º showed only minimal improvements.

Since the very early 1980's most technical running shoes have incorporated anti-pronation or subtalar control devices. Anti-pronation devices are designed to work in two ways. Some are designed to stiffen the shoe and thus physically restrain the movement of the subtalar joint, whereas others modify the shoe cushioning to reduce the lever arm of the GRF in an attempt to decrease the amount of torque that leads to pronation (Shorten, 2000). Such devices include harder cushioning, heel counters, insole boards, medially posted midsoles and vagus wedges (Shorten, 2000).

Running shoes have been shown to alter both running kinetics and kinematics compared to barefoot running. Clarke et al. (1983) describe barefoot trials as a baseline for conventional running research. Running shoes have been shown to increase total pronation, Shorten (2000) proposes that the softer midsoles found in cushioned shoes, do not adequately restrain the movement of the subtalar joint. Bates et al. (1978) as cited in Edington et al. (1990) determined that removing the running shoe resulted in a significant increase in the time to maximum pronation in addition to a significant reduction in total pronation angle. Stacoff et al. (1991) examined total pronation angles during barefoot and shod running. They found that shod running was associated with a significantly higher angle of pronation compared to barefoot trials.

2.10 Barefoot running

In recent years the concept of barefoot running has experienced a resurgence in footwear biomechanics literature. A number of well-known athletes have competed barefoot, most notably Zola Budd-Pieterse and the Abebe Bikila who both held world records for distance running disciplines. Thus barefoot running does not appear to prevent athletes from completing at the highest levels (Warburton, 2000).

The majority of daily activities in contemporary populations take place in some degree of footwear. Shoes are traditionally introduced at an early age and are designed to provide a safer, more comfortable environment for the foot. However, Staheli, (1991) proposes that in general shoes may impair the natural growth and development of the feet. Wearing shoes has been revealed to initiate the progressive tapering of the forefoot area (Hoffman, 1905) and to be connected with the development of hallux valgus deformity (Sim-Fook and Hodgson, 1958). It is hypothesized that these disorders associated with the forefoot region of the foot may transpire because the shoe may cause constriction of the toes during the developmental years and into adulthood (Bonney and Macnab, 1952).

In countries where both barefoot and shod populations reside, it is commonly reported that the incidence of injury to the lower extremities is considerably higher in shod inhabitants (Robbins and Hanna, 1987). In addition, overuse injuries to the lower extremities that may be attributed to running are uncommon in developing countries, where the majority of residents live barefoot (Robbins and Hanna, 1987). The study revealed that there were no reported instances amongst the barefoot subjects of Onychrocryptosis, Hallux Valgus, Hallux Varus, and Bursitis at the first or fifth metatarso-phalangeal articulations (Shulman, 1949).

Shulman, (1949) suggested that those who do not use footwear develop few foot problems, the majority of which are non-debilitating. Their overall foot ranges of motion were also found to be greater than the non-shod population, allowing complete foot activity. Shulman, (1949) thus drew the conclusion that footwear is not required for healthy feet and are the cause of most aliments. A review by Robbins et al. (1987) reported a particularly low frequency of running related injuries in barefoot populations in comparison to shod populations. It was noted that despite the design characteristics of the modern running shoe, running injuries were common amongst even runners with a relatively low average weekly mileage. The frequency of lower extremity injuries associated with protective footwear and relative resistance to injury in the barefoot state presents a paradox in footwear literature (Robbins et al. 1987). To rationalize this paradox Robbins et al. (1987) hypothesized that the kinematic adaptations associated with barefoot running provide impact attenuation and security from running injuries.

An adjustment involving deflection of the medial longitudinal arch during the impact phase on foot-ground contact is theorized to be a significant adaptation providing impact absorption (Robbins et al. 1987). The arch support, which is found in all running footwear, may interfere with the downward deflection of the arch during loading. The most frequently cited acute injuries associated with running are ankle sprains, Warburton, (2000) proposed that the majority of these injuries are inversion injuries causing damage to the anterior talofibular ligament. Footwear increases the incidence of these sprains, Warburton, (2000) hypothesizes that the mechanism for this occurrence was attributable to either a diminished awareness of foot orientation via feedback from the plantar surface mechanoreceptors interface from the ground, or by increasing the lever arm about the subtalar joint itself due to the shoe cushioning system that consequently increases the twisting torque.

The majority of epidemiological research concerning barefoot vs. shod running has concerned overuse/chronic injuries. Common overuse injuries include shin splints, plantar fasciitis, Patella femoral pain syndrome and iliotibial band syndrome and are traditionally attributed to excessive rearfoot motion and impact forces (Taunton et al. 2003).

Epidemiological research suggests that plantar fasciitis is one of the most common overuse injuries in runners (Taunton et al. 2003). Robbins and Hanna, (1987) hypothesize that the plantar fascia supports the medial longitudinal arch, and that the inevitable strain placed on the tissue attachment points during the loading phase of footstrike can lead to the development of plantar fasciitis (Robbins and Hanna, 1987). Warburton, (2000) theorized that running barefoot necessitates kinematic adaptations that serve to redistribute the impact to the surrounding musculature therefore the potentially harmful mechanism by which the plantar fascia tissue may incur damage is reduced.

In direct contrast to early biomechanical assumptions and the results obtained from in vitro analyses, it seems that running shoes do not always attenuate impact forces during running (Shorten, 2002). Both Robbins et al. (1988) and Robbins and Gouw, (1991) hypothesize that shoe midsole cushioning properties serve to reduce the perceived impact load experienced by the foot itself, thus leading to a perceived underestimation the actual impact shock being applied to the lower extremity. Warburton, (2000) proposed that an underestimation of actual loads can be a causal feature in the development of overuse injuries.

Manufacturers of high-end/expensive running shoes traditionally claim that their shoes serve to reduce pronation and provide shock attenuation. Robbins and Gouw, (1991) however found that wearers of expensive running shoes suffer more running overuse injuries. Robbins and Waked (1997) also indicated that expensive shoes accounted for twice as many running related injuries as cheaper shoes, a finding that led the authors to conclude that deceptive athletic footwear may present a public health hazard. Anthony, (1987) suggested that running shoes be regarded as protective devices as opposed corrective devices, as their capacity for shock attenuation and ability to reduce of rearfoot motion is limited. A study by Yessis, (2000) suggested that the structures of the foot are destabilized by long term use of footwear meaning that people must therefore rely on the external support of the footwear, which fails to match that provided by an effective foot. This lead Shorten, (2002) to conclude that perceived positive effects of shoe midsole cushioning systems are illusory and that footwear is the greatest enemy of the human foot.

Given the popularity of barefoot running as a proposed injury prevention mechanism there have been surprisingly few comparative analyses of barefoot and shod running, and of those reported the results of which have often been conflicting. De Wit et al. (2000) observed that barefoot running was associated with a significant increase in loading rates and runners landed with a significantly flatter foot position in comparison to shod running. In concurrence, Dickinson et al. (1985) found differences in ground reaction forces between barefoot and shod running. Results from their study indicated that a running shoe without a shock absorber increases the forces on the body. Hamill et al. (2011) also observed increases in loading rates and reductions in time to impact peak when running barefoot in comparison to shod. However in contrast Liebermann et al. (2010) observed that barefoot runners were associated with significant reductions in impact kinetics in comparison to shod. Squadrone and Gallozzi, (2009) observed similar increases in impact kinetics in shod running and also noted that barefoot runners landed with significantly more plantarflexion and used a higher stride frequency.

2.11 Treadmill running

The treadmill is now a common mode of exercise, and is becoming more and more popular (Corey, 2005). Treadmills were traditionally used in clinical and laboratory research, but are now used extensively in both fitness suites and homes. Treadmills allow users to adjust variables such as speed and elevation to vary the difficulty of running.

Since the early 1980's the sport of running has changed dramatically, with a significant increase in the number of treadmill runners (Milgrom et al. 2003). Runners' World suggests that, 40 million people in the U.S run using treadmills, and 70% of home treadmill owners are females (Milgrom et al. 2003). Many previous studies analyzing the kinematics of distance running have been conducted using treadmill running. Schache et al. (2001) suggest that the treadmill may be a more appropriate method for gait analysis. The convenience of the treadmill means that it is an appealing instrument for the analysis of human gait as it allows the velocity of movement and gradient to be standardized (Schache et al. 2001).

However the validity of treadmill studies particularly epidemiological analyses regarding the prevention of running injuries is limited if they cannot be generalized to overground running (which despite the increase in treadmill utilization still strongly predominates). Opinion differs regarding the legitimacy of the assumption that treadmill locomotion is comparable to overground locomotion.

Elliott and Blanksby, (1976) provided an early kinematic comparison of overground and treadmill jogging. They reported that no significant differences existed in terms of the stride length, stride frequency, stance time or recovery time when running at both 3.3 and $4.8 m.s^{-1}$.

Nigg et al. (1995) also examined subjects running overground and on the treadmill at velocities of 3.0 and 4.5m.s^{-1}. Sagittal plane kinematics of the lower extremities were studied using high speed cinematography. Based on their collected data they concluded that the measured differences in kinematic variables between treadmill and overground running showed an inconsistent pattern, which they further suggested, was dependant on the individual landing style and running speed.

Van Schenau, (1980), utilized theoretical modelling to compare treadmill and overground running and determined that the mechanics of running on a treadmill are similar so long as the belt velocity is consistent. They suggested that differences in running mechanics originate from environmental factors such as air resistance and visual/spatial information rather than mechanical factors due to the different running conditions (Van Schenau, 1980). Frishberg (1983) used a similar method to examine the differences between overground and treadmill running at higher velocities. This study suggested that overground and treadmill locomotion are mechanically different, although to what extent the variations were due to mechanical or psychological factors could not be adequately determined.

Nelson et al. (1972) compared the running kinematics of treadmill vs. overground running. They used video analysis to record runners at three speeds of 3.5, 4.8 and 6.40m.s^{-1} and on three different gradients. Their results showed that there were no significant differences between the two conditions when the running surface was horizontal. Stolze et al. (1997) indicated that at identical velocities, there were decreases in both stride length and stance time whilst running on a treadmill in addition to an increase in swing phase duration. They also reported that reductions in stride parameters almost disappeared after a period of ten minutes, suggesting that there is an accommodation period associated with treadmill running.

Despite this evidence it is still not fully understood how humans adapt their running mechanics to treadmill running (Bagestiero, 1999). Some studies also suggest that treadmill gait is not comparable to that of overground locomotion, and the correspondence between the two conditions remains a controversial topic. Kobylarz, (1990) provided an extensive comparison of overground and treadmill running. The results of the analysis suggest that significant differences exist in lower limb mechanics during treadmill ambulation in comparison to overground. The results showed increases in stance time, stride length cadence as well as alterations in knee joint kinematics. Alton et al. (1998) provided a comparison of treadmill and overground walking using both male and female subjects. For the female group, the maximum flexion angle of the hip joint was found to be significantly greater during treadmill walking. For males, significant increases were noted for cadence and maximum knee flexion angle during treadmill walking. When the two genders were analyzed cumulatively, significant increases in total hip joint ROM, maximum hip flexion and stride frequency were detected during treadmill locomotion, whilst a significant decrease was observed in stance time.

Parvataneni, (2009) provided an extensive and rigorous comparison of overground and treadmill running as part of a doctoral thesis. The report confirmed that at comparable velocities, a significant decrease in stance time and consequently an increase in swing duration was recorded during treadmill running in comparison to overground. Significant differences were also reported between overground and treadmill running in terms of the total ranges of motion about the hip, knee or ankle joints.

A 1998 examination by Wank compared both the kinematics and muscle activities associated with overground and treadmill running. Muscle activation profiles and several kinematic factors showed systematic changes between the two conditions. Treadmill running was associated with significant decreases in total bilateral swing phase duration, vertical COM displacement and the variance in joint kinematics. The majority of participants exhibited a reduction in stride length and consequently an increase in stride frequency. In addition stance times during treadmill running were found to be shorter than those recorded during overground running. Although the electromyographic (EMG) signals obtained were in the main similar between the two conditions, a number of small variations were repeatedly identifiable. Based on the results from both the kinematic and EMG analysis, Wank, (1998) hypothesized that during treadmill running, participants selected a style of locomotion that provided them with a greater level of security in an unfamiliar running environment.

Although the sagittal plane kinematics of treadmill running have received attention in biomechanics literature, very little is known about the kinetics of treadmill running. Milani et al. (1988) acknowledged that it is likely that there are kinetic differences between treadmill and overground locomotion. White et al. (1998) compared vertical ground reaction forces during overground and treadmill locomotion at three velocities 1.04,1.42 and 1.68m.s^{-1} Their results showed that although the patterns of the vertical reaction forces for the two forms of locomotion were nearly identical, small but significant differences in selected force magnitudes were evident between the two conditions. However, due to the low velocities utilized in this study it is unlikely that the results can be generalized to higher velocities more commonly associated with distance running.

The 3-D kinematics of treadmill and overground running have been examined, but to a lesser extent than traditional 2-D analyses. There is currently a paucity of comprehensive comparisons regarding the 3-D kinematics of the lower extremities from treadmill and overground running during the stance phase. Riley et al. (2008) examined the differences in hip, knee and ankle joint kinematics from both treadmill and overground motion. They examined maximum and minimum angles from the full gait cycle. Significant differences were reported in maximum and minimum knee sagittal plane parameters, with overground associated with greater peak flexion and the treadmill with increased minimum flexion angles. However, Riley et al. (2008) examined only maximum and minimum angles of the full gait cycle, therefore as the majority of these occurred during the swing phase; angles during the stance phase were not compared. Fellin et al. (2010a) investigated 3-D lower extremity motion during both treadmill and overground locomotion; their examination utilized a trend symmetry design which is an effective method of comparing the similarities between kinematic curves. They found that the overall average trend symmetry was high, R^2=0.94, yet the knee coronal and transverse plane waveforms exhibited lower similarity R^2=0.86-0.90. Unlike overground running, there are no epidemiological figures for overuse injuries associated with treadmill running (Milgrom et al. 2002). No studies have been conducted analyzing the impact characterisers associated with treadmill running in comparison to overground running.

2.12 Electromyography

An electromyogram is a reflection of the electrical potential that was generated by multiple activated muscle fibres (Nigg and Herzog, 2005). Electromyography (EMG) measures the effects of temporal imbalances of ions surrounding the muscle fibres around the position of the electrode.

2.12.1 EMG signal

Each muscle even at rest is an excitable tissue, with an electrical potential of around -90mV. This voltage potential at rest is a result of different concentration of sodium, potassium and chloride ions across the muscle sarcolemma (Kamen, 2005). To produce a contraction of a muscle, muscle fibres must receive a stimulus from a motor neuron. Once the motor neuron is stimulated, the electrical impulse produced via the central nervous system (CNS) is proliferated to the muscle motor end-plates which culminate in the generation of a muscle fibre action potential (Kamen, 2005). The EMG signal is produced when the electrical impulse reaches the pre-synaptic terminal causing the release of the neurotransmitter acetylcholine which diffuses across the synaptic cleft and binds to receptors at the motor end plate. This facilitates an alteration in membrane permeability by opening ion channels (Nigg and Herzog, 2005) allowing an influx of sodium ions which causes the polarity of the electrical potential to shift to around 30mV. Skeletal muscles contract in response to this stimulus (Nigg and Herzog, 2005).

2.12.2 SENIAM Recommendations

The Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) guidelines for surface EMG aimed to generate agreement on key aspects such as electrode design, placement and data processing so that EMG analyses and techniques could be contrasted fairly against similar studies. The SENIAM recommendations suggest that silver/silver chloride (Ag/AgCI) electrodes of 8-10mm in diameter, with a pre-gelled surface should be used in order to obtain the cleanest signal. They further recommended an electrode spacing i.e. inter-electrode distance of 20mm and placed on the belly of the muscle near the innervation zone and orientated in the direction of the muscle pennation angle. The construction of the electrode itself is not considered to be important although the mass should be minimized in order that movement artefact is avoided due to the inertial properties of the device. The SENIAM guidelines also recommend that a reference electrode should be positioned in a location such as the wrist or ankle with minimal muscle activity.

2.12.3 Surface Electromyography

EMG signals are usually measured using electrodes that quantify the differences in electrical potential between two salient points (Nigg and Herzog, 2005). There are two fundamental methods of obtaining EMG data, through surface EMG which involves placing the electrodes on the skin overlying the muscle and intramuscular EMG which involves pushing an electrode inside the muscle belly itself (Richards et al. 2008); this technique is not used widely due to ethical concerns.

2.12.4 Collecting surface EMG data

Surface EMG electrodes are available in a number of varieties. Currently the most utilized surface electrodes are the widely available Ag/AgCl electrodes. Prior to the commencement of data collection impedance of the skin must be reduced by shaving the electrode placement site, rubbing with abrasive paper to remove dead cells and finally wiping with alcohol to remove oils (Nigg and Herzog, 2005). EMG electrode recordings can be obtained using either a monopolar or bipolar electrode array. The monopolar configuration quantifies the electrical potential at the position of the electrode, with respect to the voltage potential of the ground reference. Monopolar electrode constructions are rarely used however as they can sometimes record fluctuations induced in the tissue between electrode and ground reference, which can be larger than the action potential voltage potential itself (Nigg and Herzog, 2005). Bipolar electrode configurations use two electrical contacts in order to measure the voltage potential difference between electrodes and ground reference (Nigg and Herzog, 2005). Bipolar EMG must be used with a differential amplifier. Because bipolar electrodes have two contacts that are not connected to one another, one contact will be used for positive input, and the other will be used for a negative input for the differential amplifier (Nigg and Herzog, 2005).

2.12.5 Factors influencing the surface EMG signal recording

2.12.5 (i) Muscle-tendon interface

The motor unit action potential responsible for EMG signal dissipates at the musculo-tendon interface (Nigg and Herzog, 2005). If electrodes are positioned too close to the musculo-tendon interface this may result in a distortion of the motor unit conduction properties and thus negatively disrupt the signal (Clarys 2000). As a result of this the area local to the muscle-tendon interface is typically avoided for placement of the electrode.

2.12.5 (ii) Crosstalk

Crosstalk can influence the obtained EMG signal when the activity from one of the adjacent muscles contributes to the recorded signal of the muscle under investigation (Clarys 2000). Crosstalk occurs when the signal source and detection point are not defined properly. Crosstalk is a key concern in most EMG studies that is difficult to detect and even more difficult to separate it from the true signal (Nigg and Herzog, 2005).

2.12.5 (iii) Movement Artefact

An additional noise source, the movement artefact noise, also originates at the electrode-skin interface. It is generated when: (a) the muscle moves underneath the skin, and (b) when a force impulse travels through the muscle and skin underlying the sensor causing a movement at the electrode-skin interface. The resulting time-varying voltage produced across the two electrodes can be the most troublesome of noise sources and requires the most attention. In addition to appropriate skin preparation techniques the SENIAM guidelines recommend that all electrode cables be taped to avoid pulling artefacts.

2.13 Muscle tuning

During distance running the musculoskeletal system experiences externally applied forces from the foot ground interface (Nigg and Wakeling, 2001). The application of these forces to the human musculoskeletal system generates vibrations and oscillations within the body. The primary illustration of this is the transient shockwave that is encountered by the lower extremities during the impact phase of ground contact (Valiant, 1990).Vertical impact forces during running can reach a peak of up to three body weights (Cavanagh and Lafortune, 1980). Impact forces have been linked to the development of running related overuse injuries (Hreljac, 2004).

Extreme impact loading may expose the lower extremities to overuse injuries (Nigg and Wakeling, 2001). A fundamental loading window is believed to exist, whereby tissues respond positively to impact forces of a certain appropriate magnitude (Wakeling et al. 2002). Taking into account the current knowledge of the body tissues and their responses to impact loading

Boyer and Nigg (2004) propose that new ideas are necessary regarding the effect that impact forces have on the body during locomotive movement. The impact peak during traditional heel-toe running has a frequency content of around 10-20 Hz (Nigg and Wakeling 2001). This impact is anticipated by sensory receptors, and information is then relayed to the central nervous system (Nigg and Wakeling, 2001).

This impact causes the soft tissue packages of the lower extremities to vibrate, following which the tissues continue vibrating, with the vibration magnitude deteriorating due to dampening within the tissues (Nigg and Wakeling, 2001). Several body structures serve to attenuate the magnitude of the impact shock wave through the body (Nigg and Wakeling, 2001). Alterations in joint alignment and muscular activity can be controlled and are utilized in order for the body to alter the vibration response to heel strike impact transmission during running (Wakeling et al. 2002).

Prolonged exposure to vibrations about the soft tissues has been hypothesized to detrimentally affect the soft tissues themselves (Wakeling et al. 2003). It has been suggested in recent years that the body is able to tune muscular activity to reduce vibrations in the soft tissues in order to reduce these adverse effects (Nigg 1997). This notion therefore suggests that the level of muscle activity necessary during gait is to a certain extent reliant on the magnitude of the impact shockwave transmitted to the musculoskeletal system: a concept known commonly as muscle tuning (Nigg and Wakeling, 2001).

2.14 Measuring energy expenditure

The most common method to assess differences between individuals with regards to the economy of movement is to quantify the steady state VO_2 during exercise at a confined power output or velocity (McCardle et al. 2007). This approach is only appropriate to steady state expenditure where oxygen uptake mirrors energy expenditure (Frederick, 1984).

An understanding of energy expenditure is an essential element with regards to the effect that running shoes have on the mechanics of running (Frederick, 1984). Accurately quantifying the influence of shoe design characteristics on the metabolic costs of running is of particular relevance in footwear biomechanics research.

The preferred method for the measurement of VO_2 during running is open circuit spirometry. Although this method is limited in the sense that it is unable to resolve very small alterations in VO_2 it remains the most effective technique to evaluate steady state running economy (McCardle et al. 2007).

2.15 Footwear influencing VO_2

Ergonomic characteristics can have a significant influence in the design of running shoes (Frederick, 1984). In addition to providing shock attenuation and moderating rearfoot motion, the functional role of the running shoe is to improve performance (Frederick, 1984). The mechanical characteristics of running shoes can influence the energetic cost of running which may facilitate improvements in distance running performance. It appears logical that a runner who is able to maintain a given pace with lower energy expenditure than his/her competitors maintains a competitive advantage (Frederick et al. 1983).

Catlin and Dressendorfer, (1979) suggested however that little evidence exists to suggest that footwear, within normal confines, has an influence on performance that is large enough to outweigh the base physiological attributes of the runner. Morgan et al. (1989) propose that the oxygen cost of running at a given velocity may be influenced by the shoe. The alterations in potential and translational kinetic energy of the foot during running are substantial (Chapman and Caldwell, 1983; Williams and Cavanagh, 1983). During a complete gait cycle at typical distance running velocities each foot is moved approximately 0.5 metres and accelerated to roughly twice the velocity of the torso before being decelerated to zero upon foot contact (Frederick, 1984). This suggests that carrying even small increases in weight on the feet may have a sizeable influence in the economy of running (Frederick, 1986).

Studies examining the influence of footwear on running economy commenced during the 1940's, Russell and Belding, (1946) as cited in (Frederick, 1986) were commissioned by the U.S army during World War II in order to determine the effects that military footwear had on the energy cost of walking. Their study compared energy consumption when walking barefoot and a range of army boots selected due to their vast variations in weight. As hypothesized the heavier boot necessitated significantly more energy expenditure. It was also reported that at higher velocities and when walking on gradients the relative energy cost of locomotion increased. It was concluded that due to the significant alterations in potential and rotational kinetic energies of the foot during locomotion that the energy cost of carrying additional weight on the shoe will have a greater relative influence than carrying the same weight about the torso (Frederick, 1986).

In a further investigation Russell and Belding, (1946) evaluated the influence of adding weight to standard combat footwear compared to adding the same weight near the body COM. The results indicated that carrying additional weight away from the COM was four times more costly in terms of energy expenditure. Catlin and Dressendorter, (1979) analyzed the energy expenditure of athletes wearing two different footwear conditions. The two models differed by 350g per pair. Their results indicated that the two models differed significantly from one another in terms of energy expenditure, with the heaviest shoe necessitating on average 3.3% more energy. It was concluded that the difference in energy expenditure was attributable to the discrepancy in weight between shoes.

However, when more rigorous studies examining the economy of locomotion began to be conducted it was established that this was not entirely accurate (Frederick, 1984). Frederick et al. (1984) studied the energy expended when carrying additional weight on the feet during running as opposed to wearing shoes of identical additional weight. Once again a weight difference of 350g was observed between conditions. They found that adding additional weight on the feet was less energetically costly than Catlin and Dressendorter, (1979) in that wearing weighted shoes necessitated only an additional 1.8% increase in energy expenditure. Thus it appears that only a portion of the difference in energy cost between conditions can be explained by weight alone. Frederick, (1984) hypothesized that the discrepancy in energy expenditure between the two studies is attributable to the fact that the shoes utilized by Catlin and Dressendorter, (1979) were different enough in design characteristics to produce alterations in running economy beyond the influence of weight alone.

A number of investigations have found that numerous non-weight effects aspects of footwear design can influence both running kinetics and kinematics of running (Clarke et al. 1983). Thus Frederick, (1986) suggested that it is not unexpected that variations in energy expenditure have been reported between footwear with various midsole characteristics. It appears reasonably straightforward that changes in kinematics influence running economy, as alterations in kinematics necessitate changes in muscle activation; therefore the energy cost associated with locomotion is a direct result of skeletal muscle metabolism (Williams, 1990). Running kinematics have also been found to influence running economy. Cushioned footwear is hypothesized to reduce the oxygen cost of running by 2-5% in comparison to barefoot and hard soles (Shorten, 2000). The mechanism in which cushioning influences oxygen consumption is related to the adjustments runners make in running kinematics to compensate for the lack of cushioning. Runners adopt their gait pattern by increasing knee flexion and ankle plantarflexion at touchdown (De Wit et al. 2000). These adjustments may necessitate greater muscular mechanical work, thus the body incurs an energy penalty, referred to as the cost of cushioning (Frederick et al. 1983).

Frederick et al. (1980) noted a significant difference in energy expenditure in a group of male participants who ran in both air and non-air soled shoes. The air soled shoes required a total of 2.8% less VO_2 despite weighing slightly more than the non-air shoes. In concurrence Frederick et al. (1986) compared the energy demands of running in footwear with varied midsole cushioning properties. Participants completed treadmill running at $4.13 m.s^{-1}$; the energy expended per kilometre was quantified using ten experienced distance runners. The shoes used in the research were conventional and available to purchase commercially. The first shoe was a conventional running shoe with EVA midsole, whilst the second was similar in design but featured a midsole that utilized a 1cm air cushion thus making it noticeably softer (Frederick et al. 1986). The results indicated that the O_2 demands of running were significantly lower when wearing shoes with soft midsole cushioning.

Frederick et al. (1980) also indicated a significant reduction in energy expenditure when running in soft soled footwear. They found that VO_2 was significantly decreased in elite runners completing treadmill running at 5.37 $m.s^{-1}$. The runners utilized 194.0ml.O_2.kg.km when wearing conventional lightweight shoes and 191.5ml.O_2.kg.km when wear a pair of much heavier soft soled footwear. This study revealed a criticism within the methodological designs of the majority of studies, in that other variables aside from the midsole properties were not controlled for (Frederick, 1986).

Although economy does not provide information regarding the mechanism of how various segmental movements influence energy cost (Williams, 1990), the assumption that reducing oxygen consumption by optimising running mechanics will improve performance is a rational one (Frederick, 1986; Novacheck, 1998; Williams, 1990). Despite the central association between running mechanics and energy cost, research has yet to establish a clear mechanical description of the economic runner.

3. General Methods

This section provides a summary of the equipment and techniques used throughout all of the experimental testing in this thesis. More specific methods will however be provided in the methods section of each chapter and in the validation of experimental techniques chapter.

Publications

1. Sinclair J, Taylor PJ and Hobbs SJ (2012). Alpha level adjustments for multiple dependent variable analyses and their applicability – A review. International Journal of Sport Science and Engineering, Vol.06, No. 03, pp. 134-142.
2. Sinclair, J, Hobbs, SJ, Morley, A and Taylor, PJ (2012). The appropriateness of Multi (MANOVA) and single-variate examinations, for statistical analysis of human movement: A Review. Sport Science Reviews (In press).

3.1 Introduction

For all of the investigations carried out during the course of this thesis the same fundamental equipment was utilized, these being a 3-D motion capture system, force platform, tibial accelerometry and electromyography. These systems are effective in that they allow synchronized data to be collected and quantified.

3.2 3-D motion capture

3.2.1 Qualisys Oqus 310 motion capture system

The Qualisys motion capture system (Qualisys Gothenburg, Sweden) uses passive infrared technology to capture retro reflective markers positioned on a body. The cameras used throughout were the same model Oqus 310 series.

Figure 3.1 Qualisys Oqus motion capture camera

3.2.2 Calibration

In order to identify the extrinsic parameters of the camera, the camera pose with respect to the lab co-ordinate system must be identified (Richards et al. 2008).

Figure 3.2: Static reference L-frame and wand used for dynamic calibration of 3-D Oqus system.

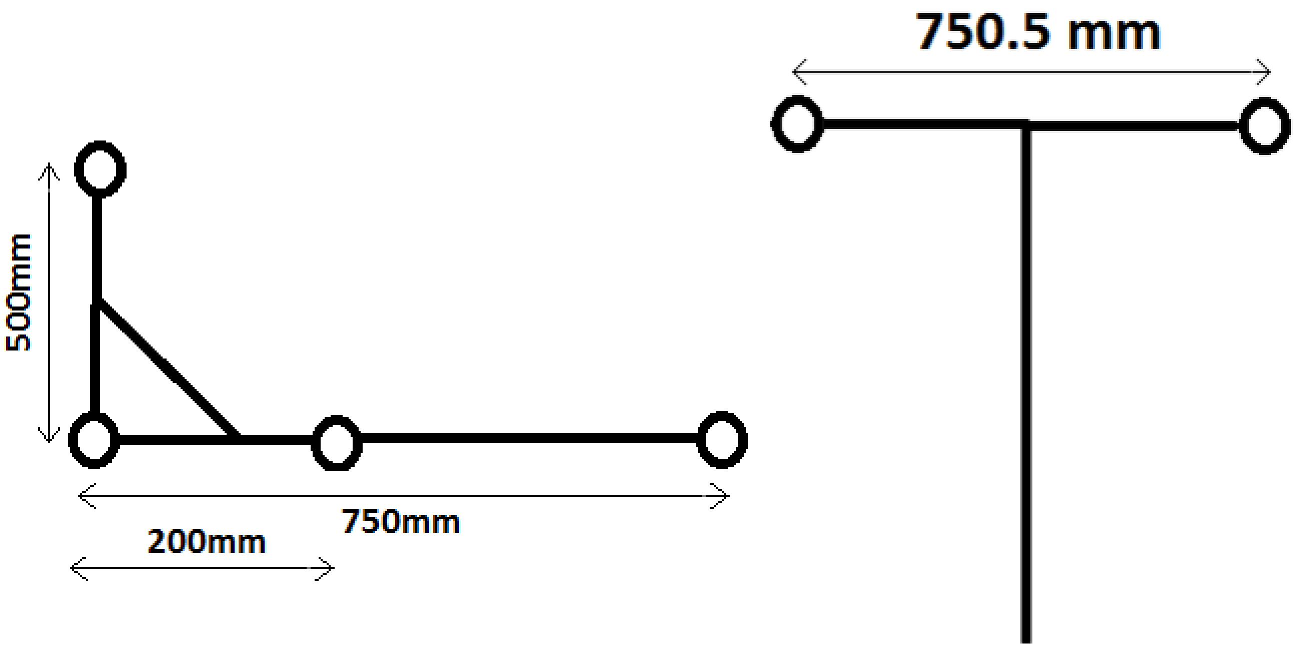

Figure 3.3 Schematics of the L-frame and wand

To determine the extrinsic properties of the camera system, the lab global co-ordinate system (GCS) must be defined. This is accomplished using a static reference L-frame which defines the origin of the GCS (Richards et al. 2008). In addition to the static reference L-frame a T shaped wand (figure 3.2) is moved through the anticipated movement volume. The camera system will obtain 500 data points from the calibration which are included into a bundle adjustment to obtain the position and orientation of the camera's and wand.

Throughout the course of this project the Qualisys motion capture system was calibrated prior to any data collection for a total of 30 seconds. From each calibration two key factors were extrapolated. 1). Norms of residuals are calculated which refers to the error associated with the camera system and 2). Standard deviation of the known wand length which provided information regarding the potential errors in the quantification and spatial representation of marker positioning.

3.2.3 Camera placement

The Qualisys motion capture system was configured as per figure 3.4 in order to track stance phase kinematic motions during running.

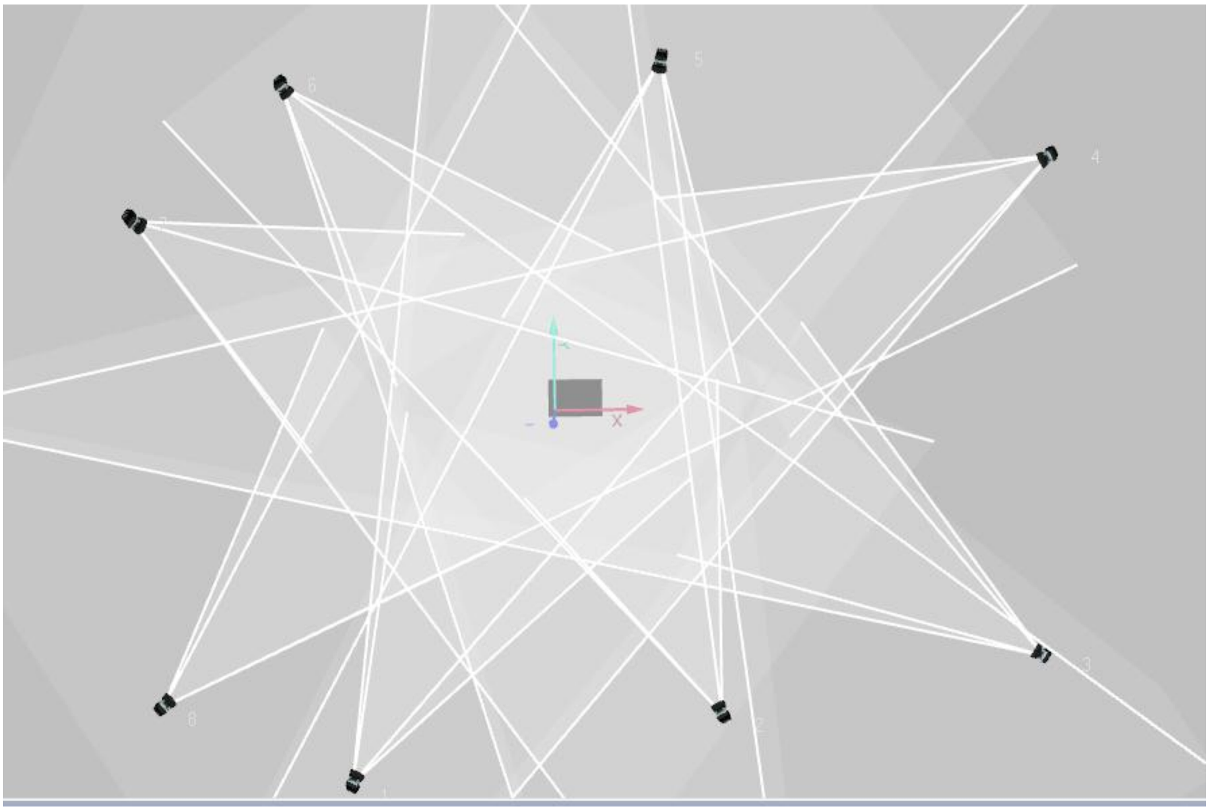

Figure 3.4 Camera positioning around the force platform using eight camera set-up.

3.2.4 Calculation of 3-D kinematics

3.2.5 Methods of Quantifying 3-D kinematics

Throughout the course of this thesis 3-D tracking/modelling of specific segments were achieved using the Calibrated Anatomical System Technique (CAST) developed by Cappozzo et al. (1995). The CAST technique allows segments to be modelled in 6 degrees of freedom (6 DOF) and involves the identification of an anatomical frame for each segment via the determination of specific anatomical landmarks and a technical frame using tracking markers (Richards et al. 2008). These segments are defined as below.

3.2.5 (i) Foot

The foot modelled and tracked throughout this thesis was considered to be a single rigid segment. To delineate the anatomical axes of the foot, anatomical landmarks were placed over the 1st metatarsal and 5th metatarsal to define the distal end over the medial and lateral malleoli to define the proximal end. For the foot the orientation of the segment co-ordinate system axes were defined as the mid-point of the malleoli markers.

3.2.5 (ii) Shank

To model the shank, anatomical landmarks were placed over the medial and lateral epicondyles of the femur to define the proximal end and over the medial and lateral malleoli to define the distal end. The segment co-ordinate system axes were defined as the mid-point of the femoral epicondyle markers.

3.2.5 (iii) Thigh

The thigh segment was defined at the proximal end using the hip joint centre and by the medial and lateral epicondyles of the femur at the distal end. The hip joint centre was determined/estimated based on the Bell et al. (1989) regression equation. The location of the hip joint centre was defined as 0.36* Distance between ASIS markers medial to the ASIS marker, distance, 0.19* Distance between ASIS markers posterior to the ASIS marker, 0.3* Distance between ASIS markers inferior to the ASIS marker. The segment co-ordinate system axes origin for the thigh segment was at the hip joint centre.

3.2.5 (iv) Pelvis

The pelvis was constructed using the CODA option in visual 3-D via the left and right PSIS markers and left and right ASIS markers. The segment co-ordinate system axes origin for the pelvis segment was defined as the midpoint between the ASIS markers.

The positioning of the anatomical markers allows the proximal and distal ends of the segments to be defined, providing an anatomical co-ordinate system for each segment. Definition of the medial and lateral aspects of the segment end points allows the midpoint between the markers to be defined as the origin of the segment anatomical co-ordinate system. Two conjoined segment endpoints produce a joint. Once the anatomical co-ordinate system axes have been defined the segment co-ordinate system is then referenced in relation to the position of a technical cluster array of markers following acquisition of a standing static calibration of the model.

To track and establish the technical frame of each segment participants were instrumented with segmental clusters made of specifically moulded carbon fibre (See figure 3.7). The clusters were designed in accordance with the Cappozzo et al. (1997) guidelines with a long and short axis. The femoral cluster had a length to width ratio of 2.05:1 and the tibial cluster had a length to width ratio of 1.5:1. This falls within the guidelines provide by Cappozzo et al, (1997) of 1.5-2.1:1. The anatomical location of clusters of markers utilized to track segmental rotations is not crucial as the CAST method requires the positions of the markers during the static trial to define segment endpoints (Richards and Thewlis, 2008).

Technical frame position can be anatomical or arbitrary either in the form of individual markers or clusters and can be attached to the skin or using specific rigid cluster plates (Richards and Thewlis, 2008). However, these markers must be placed in such a position that allows them to be tracked effectively. A minimum of three non-collinear markers is necessary to track the segment POSE in six degrees of freedom (Cappozzo et al. 2005). Based on the work by Manal et al. (2000) clusters of four markers mounted to a lightweight shell are the most effective method for segmental tracking. As the orientation of the cluster is important, the longest principal axis of the marker cluster distribution was oriented toward the relevant anatomical landmark position (Cappozzo et al. 1997).

3.2.6 Establishing the anatomical segment co-ordinate system

In order to define the anatomical co-ordinate system axes for each segment the position of three anatomical points are necessary (Richards et al. 2008). Throughout this project the SCS Z axis is determined by the unit vector directed from the distal segment end to the proximal segment end. Next, the SCS Y axis is determined by the unit vector that is perpendicular to both the frontal plane and the Z axis. Finally, the SCS X axis is determined by the application of the right hand rule. Thus, the SCS Z axis is directed from distal to proximal, the SCS Y axis is directed from posterior to anterior, and the SCS X axis is medial-lateral in orientation (Figure 3.5 and 3.6). Joint rotations were calculated based on the notion that X is flexion-extension; Y is ab-adduction and is Z is internal-external rotation.

3.2.7 Calculating 3-D angular kinematics

In order to quantify 3-D joint angles and velocities, segmental rotations and translations are utilized. This process is achieved by establishing the pose of each segment. All joint angles and velocities were calculated using the cardan/euler technique described by Grood and Suntay (1983). These angles are represented by the projection of orthogonal vectors from one co-ordinate system axes relative to the orthogonal planes of another. In order for the segment pose to be reconstructed and a joint angle to be quantified; two key pieces of information are required, a position vector and a rotation matrix. Rotation matrices describe the axes i.e. xyz of the of the segment co-ordinate system and the position vector defines a pivot point between two segments, allowing the orientation of the segments to be known and joint angles to be obtained provided that a proximal reference segment is delineated. The orientation of a segment co-ordinate system is determined using independent projection angles that relate to three rotational degrees of freedom. These angles however must be performed in an ordered sequence as they are not commutative. The order in which the xyz rotations are placed in the sequence may ultimately effect the orientation of the segment axes and thus the resultant joint angles.

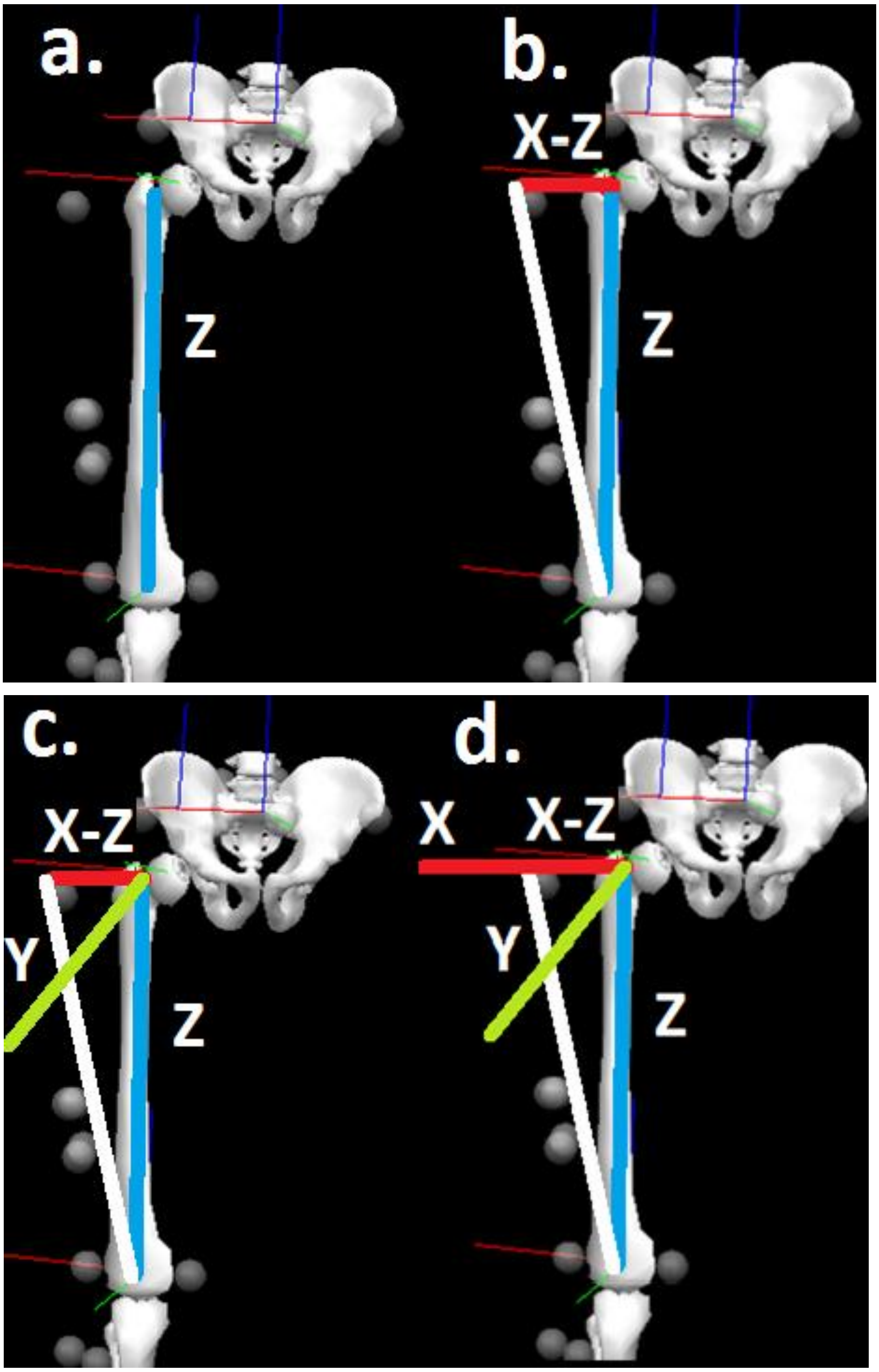

Figure 3.5: Construction of anatomical frame segment co-ordinate system.

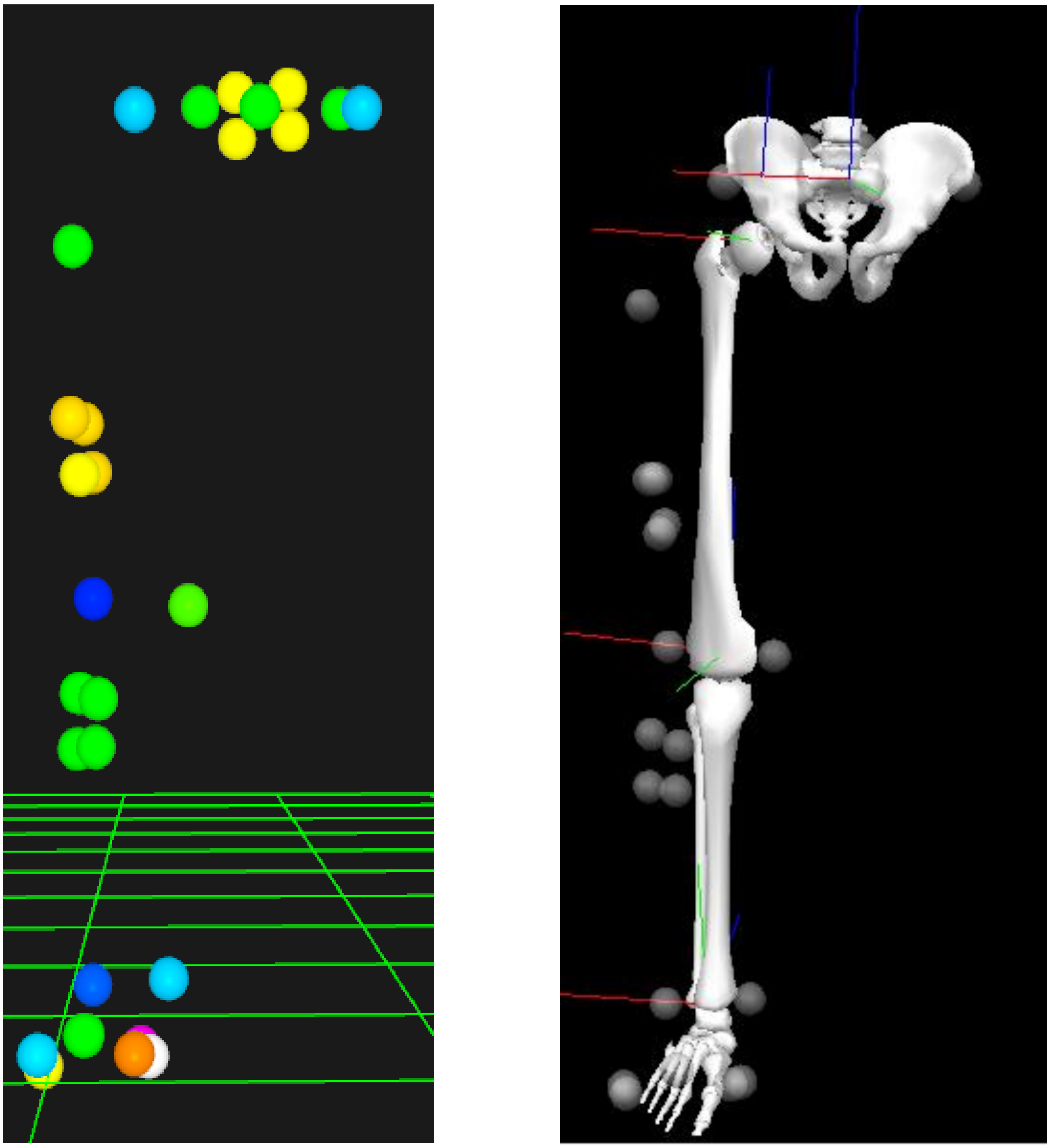

Figure 3.6: Positioning of lower extremity anatomical and technical tracking markers.

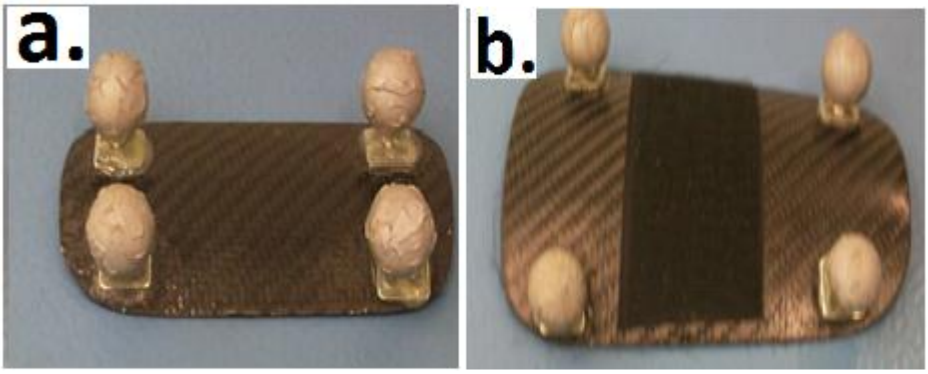

Figure 3.7: Rigid carbon fibre tracking clusters (a. = tibial cluster and b = thigh cluster).

3.2.8 Sampling frequencies

All kinematic information throughout the project were obtained at 250 Hz.

3.2.9 Extraction of salient points from the stance phase of running

Key gait events of footstrike and toe-off were established during overground analyses via the threshold recognition function within Visual 3-D. This process involved using vertical force platform information to define these events; a threshold of 20N was used allowing the stance phase to be delineated and any information outside these events to be ignored. However, for analyses conducted using the treadmill additional work was necessary to determine the most appropriate technique. Furthermore, it was also necessary to determine the peak angle during the stance phase. These measures allowed further discrete variables of peak angle, range of motion (ROM) from footstrike to toe-off and relative ROM from footstrike to peak angle to be extracted (Figure 3.8).

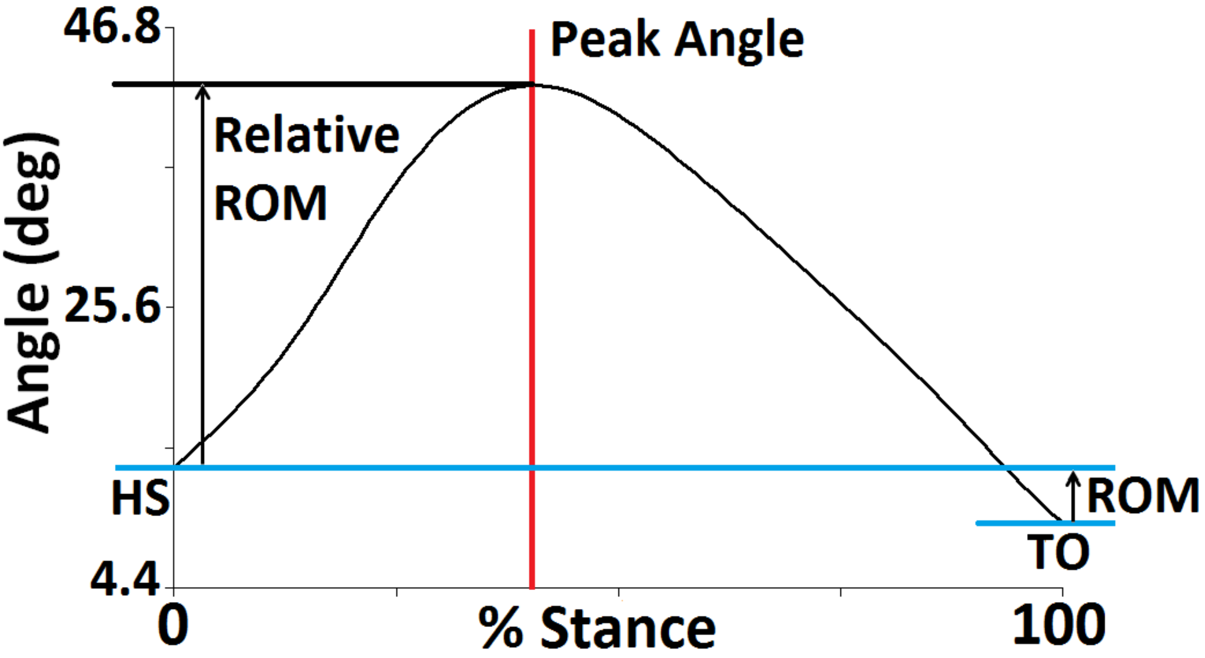

Figure 3.8: Identification of the discrete angular variables from the stance phase (HS=Footstrike and TO = Toe-off).

3.2.10 Data collection, format and analysis

All data were collected using Qualisys track manager (QTM) as .QTM files. These files included kinematic and analog information from the measured trials. Once the trials had been digitized and the appropriate markers had been identified, the .QTM files were exported as.C3D files. This format allowed the files to be imported into Visual 3-D (C-motion Inc, Germantown, MD, USA) where calculation of the outcome measures was undertaken. Further, data analysis and tabulation was undertaken using Microsoft excel (Microsoft Corp., Redmond, WA, USA) and SPSS (SPSS Inc., Chicago, IL, USA).

3.3 Force platform and quantification of forces

3.3.1 Kistler force platform specifications and details

The force platform utilized throughout this project is a piezoelectric Kistler 9281CA model. The force platform has dimensions of 60 mm length by 40 mm width. All force platform information was collected and interfaced through Qualisys track manager software, allowing synchronous 3-D kinematics and force platform information to be obtained. The force platform gain was set to as per the recommendations of the manufacturer. Furthermore, the plate was calibrated on installation by the manufacturer, allowing the precise position of the force platform with respect to the top of the laboratory flooring to be calculated.

3.3.2 Sampling frequencies

All force platform information throughout the project were obtained at 1000 Hz.

3.4 Accelerometer and quantification of tibial accelerations

3.4.1 Biometrics accelerometer specifications and details

The device used throughout this thesis to quantify tibial accelerations was a tri-axial accelerometer (Biometrics ACL 300, Gwent, United Kingdom). The accelerometer has dimensions of length 15.0mm, height 12.7mm and width 10.0 mm and a measurement range of ±500 g. The accelerometer was mounted to the skin using a custom piece of carbon fibre. The combined mass of the accelerometer and mounting piece was 9 grams. All accelerometer information was interfaced through Qualisys track manager software, allowing synchronous 3-D kinematics and tibial acceleration information to be obtained.

3.4.2 Sampling frequencies

All tibial acceleration information throughout the project were obtained at 1000 Hz.

3.5 Electromyography (EMG) collection of data

Muscle activity throughout this thesis was obtained via surface electrodes (Biometrics SX230-1000). The electrodes were capable of measuring frequencies between 20 and 450Hz over ± 4.8V and featured inbuilt amplification to interference noise. The Biometrics electrodes are housed in a 35.0 x 19.8 x 5.4mm polycarbonate casing, which feature a contoured surface in order to minimize the interference between casings and the electrode bars as well as maximizing skin adhesion. The electrodes use two circular parallel bars 10mm by 1 mm made of 99.9% pure silver, with an inter electrode distance of 20 mm, as advised by Basmajian (1974) and SENIAM, giving a detection area of $10mm^2$. All EMG information was interfaced through Qualisys track manager software, allowing synchronous 3-D kinematics and EMG information to be obtained.

3.5.1 Sampling frequencies

All EMG information throughout the project were obtained at 1000 Hz.

3.6 Gas analysis

3.6.1 Metalyzer® 3B

Breath-by-breath measurements of respiratory gases were made using a MetaLyser 3B® (Cortex Biophysic, Leipzig, Germany). Before each testing session the system was calibrated by inputting the atmospheric pressure, following which the pneumotach volume sensor was also calibrated using a 3 litre syringe (Hans Rudolph Inc, Kansas city USA). Lastly, the gas sensors were calibrated using ambient air and known gas concentrations of 5.09% O_2 and 14.46% CO_2. A Hans-Rudolph mask (Hans Rudolph, Inc, USA) was held securely around the face using a mask harness. The mask was attached to the volume analyser using a sample line via a volume sensor, thus allowing respiratory gases to be obtained.

3.7 Normalization of data

Four principal normalization techniques are employed throughout this project. Firstly, all time dependant information were normalized between events established using Visual 3-D. This was performed using a linear interpolation technique whereby the data was normalized to 101 data points. The second normalization technique that is employed refers to the normalization of GRF's. This was employed by dividing the recorded forces by each participant's body weight in newton's, allowing the forces to be reported in body weights. The third method employed in this project is the normalization of EMG amplitude. This was performed by dividing the recorded value by a reference value called a maximum voluntary contraction (MVC). Finally the inspired and expired gases were expressed relative to size by dividing the obtained values by each participant's body mass in kg.

3.8 Statistical Analyses

3.8.1 Descriptive statistics

For each study and pilot study used throughout this work, descriptive statistics of means and standard deviations were used to describe not only the outcome measures but also the characteristics of the participants i.e. age, height, mass.

3.8.2 Inferential statistics

The principal aims of the investigations in this thesis were to examine differences between footwear conditions or groups in order to make informative choices regarding appropriate footwear selection. All inferential statistical analyses were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA) with statistical significane accepted at the $p \leq 0.05$ level. Given the number of dependent variables or outcome measures that were examined in each study this would typically be examined using multivariate statistics in the form of a MANOVA. However, that the varialbes examined are highly inter-related would violate one of the prinicpal underlying assumptions of multivarite analyses (Sinclair et al. 2012). Therefore, it was determined that the most appropriate technique would be to perform examinations for each dependent variable or outcome measure which avoids the problem of inter-relations between multiple variables yet still allows comparisons between footwear/conditions to be made using either t-tests of ANOVA depending on the number of independent variables being contrasted.

This technique however may raise questions to so scientists particularly from the behavioural or social sciences regarding the adjustment of the alpha level to control for multiple comparisons. However, whilst this is common place in some fields; in epidemiological and semi epidemiological examinations such as the current project Rothman (1990) and Sinclair et al. (2012) state that no corrections for multiple comparisons are neccesary when examining main effects as it may lead to potentially important clinical observations being missed.

3.8.3 Tests of normality

The Normal Distribution is an important consideration when drawing statistical inferences from any data set. The normal distribution maximizes information entropy among all distributions with known mean and variance, which makes it the natural choice of underlying distribution for data summarized in terms of sample mean and variance. The normal distribution is the most widely used family of distributions in statistics and traditional statistical analyses center around the assumption of normality. Therefore, it is important to run tests for normality of all outcome measures before conducting any inferential statistical analyses. Normality was examined throughout this thesis using a Shapiro-Wilk test, which is considered to be the most effective test for normality and designed for sample sizes of less than fifty. An alpha level of ≤0.05 denotes a data set that has violated the normailty assumption.

3.8.4 Statistical power estimations

Previous investigations have found significant differences between kinetic and kinematic measures with as little as eight participants. To ensure sufficient participants were recruited for the investigations throughout this thesis an a priori statistical power analysis was conducted in order to avoid type II error. This was performed using custom software via the Hopkins technique (Sport sci.org) and utilized an alpha level of $\alpha=0.05$ for type I error and a beta level of $\beta=0.80$ and was based around a moderate effect size of 0.6. The analysis revealed that a sample size of twelve participants would be sufficient in order to provide 80% statistical power.

4. Development and accuracy/validity of experimental techniques

This chapter describes a series of experimental pilot investigations that were conducted in order to establish/determine the most effective techniques for the collection and processing of the outcome parameters.

Publications

1. Sinclair, J., Taylor, P.J., Edmundson, C.J., Brooks, D., and Hobbs, S.J., (2012). Influence of the helical and six available cardan sequences on 3-D ankle joint kinematic parameters. Sports Biomechanics, 11, 430-437.
2. Sinclair, J., Taylor, P.J., Edmundson, C.J., Brooks, D., and Hobbs SJ (2012). The influence of two different hip joint centre identification techniques on 3-D hip joint kinematics, Journal of Sports Sciences, S15.
3. Sinclair, J, Taylor, P.J., Edmundson, C.J., Brooks, D., and Hobbs (2012). Optimal frequency cut-off for filtering kinematic information during running – an examination of recommended techniques. Journal of Sports Sciences, S56.
4. Sinclair, J. Taylor P.J, Greenhalgh A, Edmundson CJ, Brooks, D and Hobbs S.J (2012). The test-retest reliability of anatomical co-ordinate axes definition for the quantification of lower extremity kinematics during running. Journal of Human Kinetics, 35, 27-40.

5. Sinclair, J, Hobbs, SJ, Taylor, PJ, Currigan, G and Grenhalgh, A (2012). The influence of different force measuring transducers on lower extremity kinematics, Journal of Applied Biomechanics (In press).
6. Sinclair, J, Hobbs SJ, Currigan G and Taylor PJ (2012). The reliability of three hip joint centre location techniques. Acta of Bioengineering and Biomechanics, 14, I First.

Conference Presentations

7. Sinclair J, Hobbs SJ, Edmundson CJ and Brooks D (2010). The influence of different cardan sequence of rotations on ankle joint kinematics. Bases Annual Biomechanics interest group. University of Bath, 21st April.
8. Sinclair J, Hobbs, SJ, Protheroe L and Greenhalgh A (2010). Utilization of an externally mounted accelerometer to determine gait events. Presented at the Bases annual conference, University of Aberystwyth, 31st March-1st April.
9. Sinclair J, Edmundson, CJ, Brooks and Hobbs SJ (2011). Methods of quantifying lower extremity angular kinematics during running. European Conference of Sports Sciences, Liverpool, John Moores University ECSS.
10. Sinclair, J., Taylor, P.J., Edmundson, C.J., Brooks, D., and Hobbs (2012). The influence of two different hip joint centre identification techniques on 3-D hip joint kinematics, International Convention on Science, Education and Medicine in Sport annual conference, Glasgow, UK.
11. Sinclair, J., Taylor, P.J., Edmundson, C.J., Brooks, D., and Hobbs (2012). Optimal frequency cut-off for filtering kinematic information during running – an examination

of recommended techniques. International Convention on Science, Education and Medicine in Sport annual conference, Glasgow, UK.

4.1 Pilot study 1: Optimal calibration of Oqus 310 system.

In order to obtain accurate kinematic data and standardize the procedure used throughout the project it is important that an optimal calibration method be established. Using the equipment described in chapter 3 a total of twelve different methods of performing the calibration were evaluated using different durations (15, 30, 45 and 60's), subjective velocities (slow, medium and fast all 30's duration) actions (spin, push and combination) and operator movement strategies (static and dynamic both 30's duration) (Table 4.1).

Table 4.1 Number of points and average error residuals obtained using the different calibration techniques.

Calibration Technique	Points	Average Residuals
15's	2255.88	0.85
30's	4453.38	0.82
45's	6959.75	0.93
60's	7166.81	1.00
Brush	4185.69	0.87
Rotational	4138.31	1.25
Combination	5364.56	0.79
Slow	4173.19	1.11
Medium	4545.44	0.89
Fast	4511.56	1.03
Static	5373.06	0.89
Dynamic	5148.50	1.13

The results indicate that a 30's second calibration time, using a medium subjective velocity for wand motion in conjunction with a combination of brushing and spinning motions with the operator remaining static produced the most accurate calibration.

4.2 Pilot Study 2 - Determination of motion capture system accuracy

4.2.1 Introduction

In order to ensure that the findings from the kinematic analysis are dependable it is necessary to determine whether kinematic data obtained using the motion capture system is accurate. Therefore the aim of this pilot investigation was to determine the accuracy of the Qualisys^{TM} system in terms of its ability to spatially reconstruct marker data.

4.2.2 Methods

The calibration wand was used as a reference frame for the analysis. The calibration wand had a marker-marker distance of 750.5mm. Before the analysis the camera volume was calibrated using the same acceptance criteria outlined previously for data collection.

Data was captured for 20 seconds at the standard calibration frequency of 100Hz giving a total of 2000 data points. The 20 seconds of capture time involved 5 seconds of vertical motion, 5 seconds of medial lateral motion, 5 seconds of circular motion and 5 seconds of random motion. A total of three data sets were captured at three points within the calibration volume 1.) located at the centre of the calibration volume, 2.) at the extreme right of the volume (positive X in the lab co-ordinate system) and 3.) extreme distal end of the volume (positive Y in the lab co-ordinate system).

4.2.3 Results

The data was analyzed and accuracy of the reconstructed marker data was determined using a simple mean error analysis between the measured and actual wand lengths (Table 4.2 and Figure 4.1).

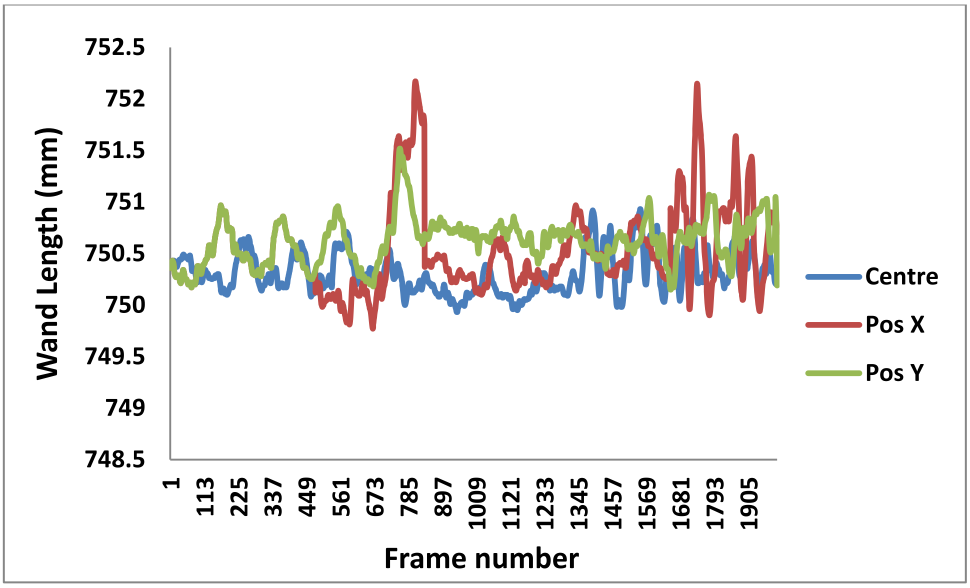

Figure 4.1: Measured wand length over 2000 frames at different areas of the calibration volume.

Table 4.2: Mean and standard deviation marker reconstruction error.

	Central	Positive X	Positive Y
Mean (mm)	750.44	750.62	750.59
Mean error (mm)	0.06	0.12	0.09
Std. Dev	0.20	0.56	0.22

4.2.4 Discussion

The results of this pilot investigation suggest that measurements taken centrally with respect to the calibrated volume are the most accurate producing mean errors of 0.06mm, whilst the positive X is associated with the greatest amount of error 0.12mm. These findings suggest that the Qualisys^{TM} camera system used in this thesis is low in measurement error allowing conclusions drawn from the inferential analyses to be dependable (Thewlis 2008).

4.3 Pilot study 3 – test-retest reliability of anatomical co-ordinate axes definition

4.3.1 Introduction

3-D kinematic analyses are used widely in both sport and clinical examinations. The computer aided movement analysis in a rehabilitation group (Leo, 1995) proposed recommendations for anatomical landmarks used to define the anatomical frame of the lower extremity segments. This was borne out of the work by Cappozzo et al. (1995) and was designed to increase the efficacy of future studies in modelling lower extremity segments.

The CAST technique involves the quantification of an anatomical co-ordinate system axes for each segment via the identification of anatomical landmarks through external palpation which is then calibrated with respect to corresponding arrays of technical tracking clusters (Richards and Thewlis, 2008). This technique is currently considered to be the gold standard for 3-D kinematic analyses (Richards and Thewlis, 2008). However, anatomical landmark identification by manual palpation and corresponding marker placement is not an error-free technique, and mal-positioning of anatomical landmarks may result in improperly defined segment co-ordinate system axes which will result in erroneous joint rotations (Kabada et al. 1989; Ferber et al. 2002). Analyses using 3-D motion capture systems are now common place in biomechanics research and reliability is of paramount importance, particularly in epidemiological or aetiological analyses when clinical decisions are made.

In sport and clinical research, where multiple participants are examined or patient's gait must be assessed over time it is essential to ensure that the identification of the relevant joint centres is reproducible. Reliable segment co-ordinate system axes are important as they provide dependable and consistent movement interpretation. Kadaba et al. (1989) and Della Croce et al. (1999), suggest that even small differences in the orientation and placement of markers forming the segment co-ordinate system can lead to sizeable differences in the calculation of joint angular parameters which may in turn inhibit the interpretation of the collected data.

Therefore analyses utilizing 3-D motion capture techniques clearly necessitate the accurate palpation of anatomical landmarks to produce repeatable, segmental anatomical co-ordinate systems (Della Croce et al. 2005). However, Della Croce et al. (2005) suggest it is difficult to place anatomical markers in exactly the same location and the determination of their location can lack accuracy and precision. Previous investigations have been conducted examining the reliability of 3-D kinematic techniques (McGinley et al. 2009; Rothstein and Echternach, 1993; Pohl et al. 2010); however the majority of these have examined either inter-session or inter-assessor reliability between sessions. Whilst these factors are clearly important to the efficacy of 3-D kinematic protocols they do not allow the reliability of anatomical frame definition to be examined effectively as different (inter-session) dynamic data is being applied to the static anatomical reference trials obtained from each session. Therefore, despite the number of investigations utilizing 3-D analysis, there is currently a paucity of research investigating the true test-retest reliability in defining the segment anatomical co-ordinate system and the influence that differences in anatomical frame definition may have on the 3-D kinematic parameters measured during the stance phase of running.

The aim of this pilot investigation was therefore to assess the reliability of the anatomical frame definition when quantifying 3-D kinematics of the lower extremities during running.

4.3.2 Methods

4.3.2 (i) Participants

Ten participants (7 males and 3 females) volunteered to take part in this investigation (age 22.4 ± 2.05 years; height 1.79 ± 0.06 m; body mass 79.1 ± 8.2 kg). All were injury free at the time of data collection and provided written informed consent. Ethical approval for this project was obtained from the University of Central Lancashire School of Psychology ethics committee.

4.3.2 (ii) Procedure

Participants ran at $4.0 m.s^{-1}$±5% over a force platform (Kistler, Kistler Instruments Ltd) sampling at 1000 Hz, stance time was determined as the time over which 20 N or greater of vertical force was applied to the force platform. Velocity was controlled using infrared photocells Newtest 300 (Newtest, Oy Koulukatu). Each participant completed five trials.

4.3.2 (iii) 3-D Kinematics

Lower extremity kinematics were obtained using the marker set and protocol outlined in chapter 5. Two static calibration trials were captured with the participant standing in the anatomical position. The first static (test) was conducted prior to the running trials and the anatomical landmarks were removed. Following completion of the running trials the anatomical landmarks were re-positioned and the second static trial (retest) was obtained. Cluster markers used to define the technical tracking frame of each segment remained rigidly in place for the duration of the analysis and were not removed, allowing the test-retest reliability of the anatomical frame to be examined. The technical frame of the foot segment was defined using four retro-reflective markers glued rigidly onto the footwear (Saucony pro grid guide II, sizes 7-9 UK). The same model of footwear was used for all participants.

4.3.2 (iv) Data processing

Motion files from each participant were applied to both static trials. Kinematic parameters from static one (Test) and two (Retest) were quantified using Visual 3-D (C-Motion Inc, Germantown, USA) and filtered at 10 Hz using a zero-lag low pass Butterworth 4^{th} order filter. Lower extremity joint angles were created using an XYZ cardan sequence of rotations. All data were normalized to 100% of the stance phase then mean processed gait trial data was reported. 3-D kinematic measures from the hip, knee and ankle which were extracted for statistical analysis were 1) angle at footstrike, 2) angle at toe-off, 3) ROM from footstrike to toe-off during stance, 4) peak angle during stance, 5) relative ROM from footstrike to peak angle 6) velocity at footstrike, 7) velocity at toe-off and 8) peak velocity.

4.3.2 (v) Statistical Analyses

Descriptive statistics including means and standard deviations were calculated for each condition. Differences in stance phase kinematic parameters were examined using paired samples t-tests with significance accepted at the $p \leq 0.05$ level. The Shapiro-wilk statistic for each condition confirmed that the data were normally distributed. Intra-class correlations were utilized to compare test and retest sagittal, coronal and transverse plane waveforms of the hip, knee and ankle. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA).

4.3.3 Results

Tables 4.3-4.8 and figures 4.2 and 4.3 present the mean ± standard deviation waveforms and 3-D kinematic parameters obtained as a function of both test and retest static trials.

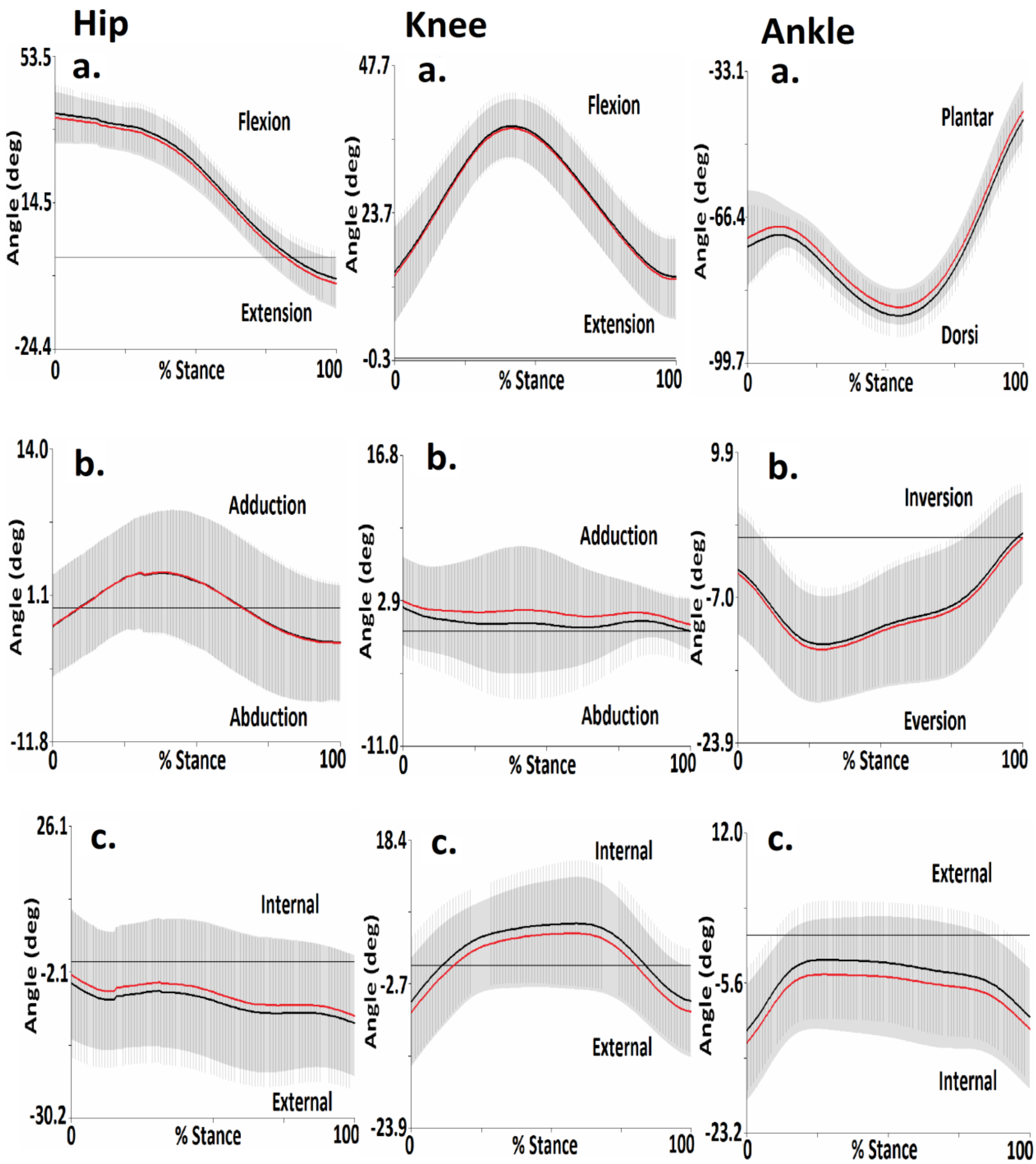

Figure 4.2: Mean and standard deviation hip, knee and ankle joint kinematics in the a. sagittal, b. coronal and c. transverse planes for Test (black line) and Retest (Red line), running (shaded area is 1 ±SD, Test = grey shade and Retest = horizontal).

Table 4.3: Hip joint kinematics (mean and standard deviation) from the stance limb as a function of Test and Retest anatomical co-ordinate axes (* = Significant main effect).

	Test	**Retest**	**Mean difference (°)**
Hip			
X (+ = flexion/ - = extension)			
Angle at Footstrike (°)	38.21 ± 3.96	39.11 ± 6.43	0.9
Angle at Toe-off (°)	-5.54 ± 6.77	-4.64 ± 6.69	0.9
Range of Motion (°)	43.77 ± 5.91	43.58 ± 6.06	0.19
Relative Range of Motion (°)	0.96 ± 0.97	0.94 ± 0.98	0.02
Peak Flexion (°)	38.73 ± 5.16	40.71 ± 5.12	1.98
Y (+=adduction/- =abduction)			
Angle at Footstrike (°)	-2.02 ± 3.96	-2.55 ± 4.84	0.53
Angle at Toe-off (°)	-3.82 ± 4.65	-4.40 ± 4.63	0.58
Range of Motion (°)	4.51 ± 2.17	5.01 ± 3.63	0.5
Relative Range of Motion (°)	5.34 ± 3.16	5.38 ± 3.19	0.02
Peak Adduction (°)	3.38 ± 4.90	2.94 ± 5.06	0.44
Z (+=internal /- =external)			
Angle at Footstrike (°)	-5.34 ± 11.36	-7.01 ± 11.71	1.67
Angle at Toe-off (°)	-13.42 ± 10.54	-13.58 ± 11.10	0.16
Range of Motion (°)	8.77 ± 5.91	8.18 ± 6.06	0.59
Relative Range of Motion (°)	9.53 ± 3.86	9.70 ± 3.78	0.17
Peak External rotation (°)	-13.99 ± 9.08	-15.16 ± 10.20	1.67

Table 4.4: Knee joint kinematics (mean and standard deviation) from the stance limb as a function of Test and Retest anatomical co-ordinate axes (* = Significant main effect).

	Test	**Retest**	**Mean difference (°)**
Knee			
X (+ = flexion/ - = extension)			
Angle at Footstrike (°)	13.88 ± 6.52	14.27 ± 6.72	0.39
Angle at Toe-off (°)	12.99 ± 5.32	13.45 ± 5.92	0.46
Range of Motion (°)	12.67 ± 2.63	2.70 ± 2.56	0.03
Relative Range of Motion (°)	2.70 ± 4.45	24.46 ± 4.44	0.24
Peak Flexion (°)	38.24 ± 3.56	38.87 ± 4.42	0.63
Y (+=adduction/- =abduction)			
Angle at Footstrike (°)	3.33 ± 4.07	2.92 ± 4.03	0.41
Angle at Toe-off (°)	0.89 ± 2.75	0.10 ± 2.91	0.79
Range of Motion (°)	3.58 ± 2.70	3.56 ± 2.70	0.02
Relative Range of Motion (°)	5.04 ± 2.87	5.83 ± 3.22	0.79
Peak Adduction (°)	-1.86 ± 4.11	-2.52 ± 4.40	0.66
Z (+=internal/- =external)			
Angle at Footstrike (°)	-5.08 ± 4.87	-2.89 ± 6.29	2.19
Angle at Toe-off (°)	-5.56 ± 6.77	-4.46 ±6.69	1.1
Range of Motion (°)	2.32 ± 1.50	2.35 ± 1.53	0.03
Relative Range of Motion (°)	12.97 ± 3.72	12.60 ± 3.82	0.37
Peak Internal Rotation (°)	8.46 ± 5.18	10.42 ± 5.96	1.96

Table 4.5: Ankle joint kinematics (mean and standard deviation) from the stance limb as a function of Test and Retest anatomical co-ordinate axes (* = Significant main effect).

	Test	**Retest**	**Mean difference (°)**
Ankle			
X (+ =plantar/- =dorsi)			
Angle at Footstrike (°)	-72.48 ± 11.10	-73.64 ± 10.34	1.16
Angle at Toe-off (°)	-43.44 ± 3.91	-45.16 ± 3.87	1.72
Range of Motion (°)	29.02 ± 12.67	28.48 ± 12.60	0.54
Relative Range of Motion (°)	15.35 ± 11.45	16.44 ± 11.53	1.09
Peak Dorsiflexion (°)	-87.35 ± 3.84	-89.99 ± 4.55	2.64
Y (+=inversion/ - =eversion)			
Angle at Footstrike (°)	-3.72 ± 7.41	-3.05 ± 7.70	0.67
Angle at Toe-off (°)	0.25 ± 4.97	1.13 ± 5.38	0.88
Range of Motion (°)	5.34 ± 2.22	5.43 ± 2.36	0.09
Relative Range of Motion (°)	9.51 ± 3.38	9.28 ± 3.39	0.23
Peak Eversion (°)	-13.24 ± 6.65	-12.33 ± 6.94	0.91
Z (- =internal/ + =external)			
Angle at Footstrike (°)	-12.13 ± 6.97	-9.91 ± 6.71	2.22
Angle at Toe-off (°)	-10.42 ± 7.17	-8.30 ± 7.26	2.12
Range of Motion (°)	2.08 ± 1.47	2.43 ± 2.36	0.35
Relative Range of Motion (°)	9.39 ± 3.57	9.66 ± 3.54	0.27
Peak Internal Rotation (°)	-2.75 ± 7.63	-0.22 ± 7.17	2.53

Comparisons between pre and post kinematic waveforms for the hip joint revealed strong correlations for the sagittal (R^2= 0.99), coronal (R^2=0.98) and transverse (R^2= 0.96) planes. For the knee joint strong correlations were observed in the sagittal (R^2= 0.99), coronal (R^2=0.96) and transverse (R^2= 0.96) planes. For the ankle joint strong correlations were observed in sagittal (R^2= 0.96), coronal (R^2=0.90) and transverse (R^2= 0.91) planes.

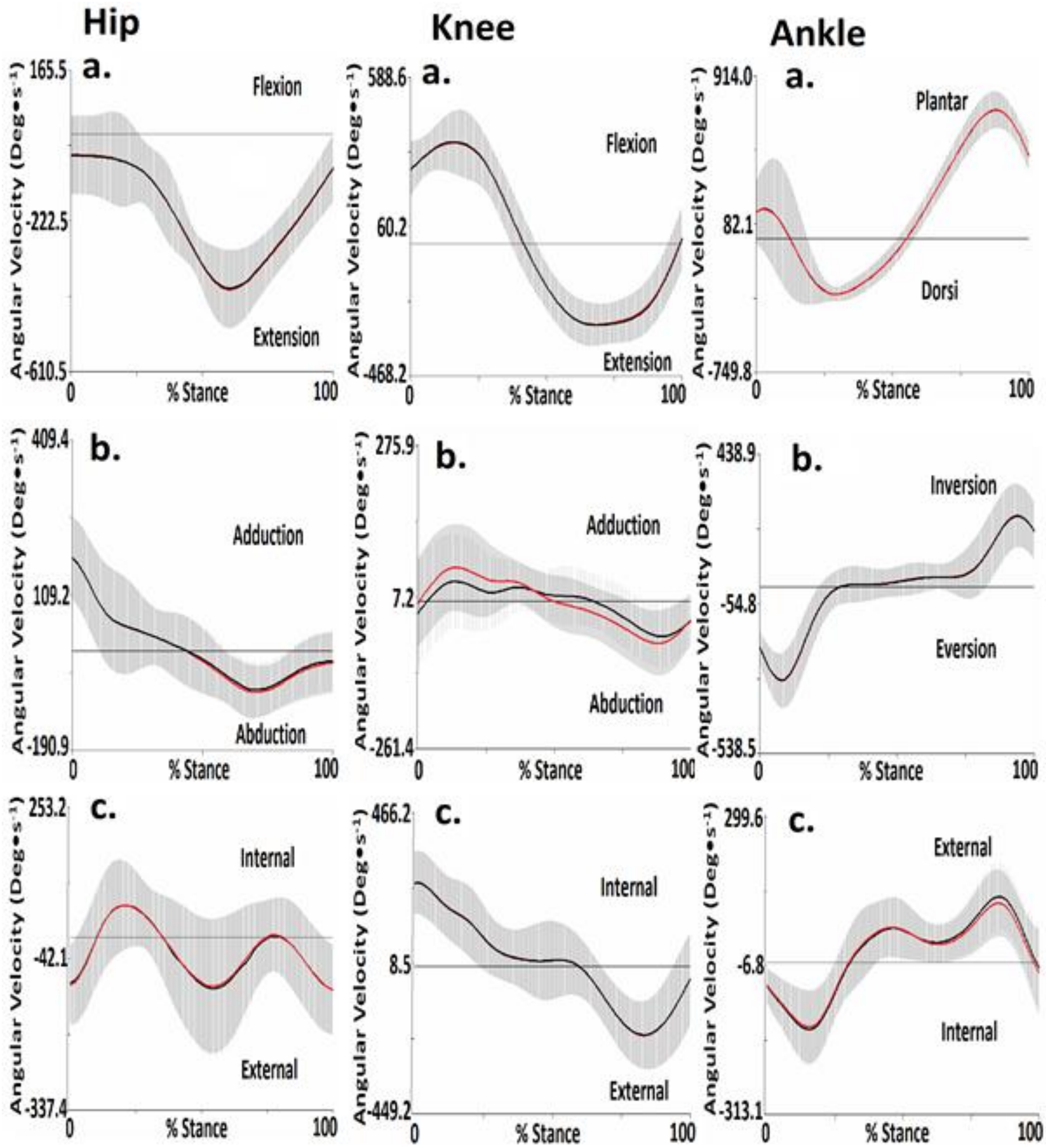

Figure 4.3: Mean and standard deviation hip, knee and ankle joint velocities in the a. sagittal, b. coronal and c. transverse planes for Test (black line) and Retest (red line), running (shaded area is 1 ±SD, Test = grey shade and Retest = horizontal).

Table 4.6: Hip joint velocities (mean and standard deviation) from the stance limb as a function of Test and Retest anatomical co-ordinate axes (* = Significant main effect).

	Test	**Retest**	**Mean difference (°.s^{-1})**
Hip			
X (+ = flexion/ - = extension)			
Velocity at FootStrike (°.s^{-1})	-54.03 ± 95.74	-55.75 ± 94.68	1.72
Velocity at Toe-Off (°.s^{-1})	-93.65 ± 76.21	-92.19 ± 79.21	1.46
Peak Extension Velocity (°.s-1)	-419.36 ± 94.91	-417.73 ± 94.25	1.63
Y (+=adduction/-=abduction)			
Velocity at FootStrike (°.s^{-1})	182.88 ± 66.48	183.37 ± 65.84	0.49
Velocity at Toe-Off (°.s^{-1})	-21.24 ±58.69	-18.43 ± 58.72	2.81
Peak Abduction Velocity (°.s^{-1})	-107.25 ± 36.60	-102.49 ± 38.37	5.26
Z (+=internal/- =external)			
Velocity at FootStrike (°.s^{-1})	-94.03 ± 67.55	-90.65 ± 76.32	3.38
Velocity at Toe-Off (°.s^{-1})	-102.24 ± 68.22	-101.20 ± 68.62	1.04
Peak Internal Rotation Velocity (°.s^{-1})	120.46 ± 42.87	120.60 ± 43.87	0.14

Table 4.7: Knee joint velocities (mean and standard deviation) from the stance limb as a function of Test and Retest anatomical co-ordinate axes (* = Significant main effect).

	Test	**Retest**	**Mean difference ($°.s^{-1}$)**
Knee			
X (+ = flexion/ - = extension)			
Velocity at FootStrike ($°.s^{-1}$)	265.89 ± 89.78	263.81 ± 83.88	2.08
Velocity at Toe-Off ($°.s^{-1}$)	20.05 ± 76.63	16.86 ± 76.64	3.19
Peak Flexion Velocity ($°.s^{-1}$)	397.68 ± 39.85	397.08 ± 61.33	0.6
Peak Extension Velocity ($°.s^{-1}$)	-320.42 ± 59.76	-322.36 ± 59.76	1.94
Y (+=adduction/-=abduction)			
Velocity at FootStrike ($°.s^{-1}$)	-13.67 ± 62.60	-21.57 ± 75.60	7.9
Velocity at Toe-Off ($°.s^{-1}$)	-34.25 ± 30.66	-36.44 ± 28.69	2.19
Peak Adduction Velocity ($°.s^{-1}$)	106.86 ± 39.85	101.46 ± 29.61	5.4
Peak Abduction Velocity ($°.s^{-1}$)	-104.20 ± 18.88	-103.07 ± 29.26	1.13
Z (+=internal/- =external)			
Velocity at FootStrike ($°.s^{-1}$)	253.64 ± 74.35	252.87 ± 74.08	0.23
Velocity at Toe-Off ($°.s^{-1}$)	-43.67 ± 123.92	-43.45 ± 123.90	0.22
Peak External Rotation Velocity ($°.s^{-1}$)	-255.83 ± 68.98	-254.56 ± 69.46	1.37

Table 4.8: Ankle joint velocities (mean and standard deviation) from the stance limb as a function of Test and Retest anatomical co-ordinate axes (* = Significant main effect).

	Test	**Retest**	**Mean difference (°.s^{-1})**
Ankle			
X (+ =plantar/- =dorsi)			
Velocity at FootStrike (°.s^{-1})	153.18 ± 163.31	153.56 ± 163.36	0.38
Velocity at Toe-Off (°.s^{-1})	466.83 ± 55.41	467.83 ± 56,06	1.0
Peak Plantarflexion Velocity (°.s^{-1})	739.35 ± 75.40	738.11 ± 75.74	1.24
Peak Dorsiflexion Velocity (°.s^{-1})	-366.96 ± 116.45	-366.91 ± 115.68	0.05
Y (+=inversion/ - =eversion)			
Velocity at FootStrike (°.s^{-1})	-195.08 ± 41.31	-194.45 ± 41.08	0.63
Velocity at Toe-Off (°.s^{-1})	180.20 ± 75.05	179.87 ± 71.69	0.33
Peak Inversion Velocity (°.s^{-1})	242.21 ± 66.95	240.51 ± 62.23	1.7
Peak Eversion Velocity (°.s^{-1})	-304.89 ± 63.17	-303.18 ± 61.95	1.71
Z (- =internal/ + =external)			
Velocity at FootStrike (°.s^{-1})	-46.14 ± 20.17	-47.69 ± 21.43	1.55
Velocity at Toe-Off (°.s^{-1})	-23.18 ± 68.83	-13.10 ± 69.36	10.08
Peak Internal Rotation Velocity (°.s^{-1})	164.33 ± 17.19	173.12 ± 20.15	8.79
Peak External Rotation Velocity (°.s^{-1})	-154.44 ± 31.96	-156.64 ±34.70	2.2

Comparisons between pre and post kinematic waveforms for the hip joint revealed strong correlations for the sagittal (R^2= 0.99), coronal (R^2=0.99) and transverse (R^2= 0.97) planes. For the knee joint strong correlations were observed in the sagittal (R^2=0.99), coronal (R^2=0.99) and transverse (R^2= 0.92) planes. For the ankle joint strong correlations were observed in sagittal (R^2= 0.92), coronal (R^2=0.90) and transverse (R^2= 0.87) planes.

4.3.4 Discussion

The aim of the current investigation was to determine the test-retest reliability of the segment anatomical reference frame definition. In the present study, running trials were analysed simultaneously using two different anatomical co-ordinate systems. This represents the first study investigating the test-retest reliability in defining the lower extremity segment anatomical co-ordinate system axes and their potential influence on 3-D kinematic parameters/waveforms during the stance phase of running.

The major finding from the current investigation is that the different anatomical reference frames obtained from the test and retest static trials had no significant ($p>0.05$) effect on 3-D kinematic parameters. This opposes the findings of Kabada et al. (1989) who observed that the angular deviations when examining reliability were much greater than those observed in the current study.

It is beyond the latitude of this study to specify acceptable levels of consistency for 3-D kinematic information. However in their review paper McGinley et al. (2009) propose that in most common clinical situations errors of 2° or less are highly likely to be considered acceptable and errors of between 2-5° are also likely to be regarded as reasonable. It is proposed that angular deviations in excess of 5° should be construed as excessive as they may be sufficient to mis-inform clinical analyses. Based on these recommendations it appears that the technique utilized in the current investigation is associated with low levels of error as the majority of test-retest angular deviations were found to be < 2°.

The intra class correlation analyses indicate that stance phase kinematic waveforms in the sagittal plane exhibited very good agreement ($R^2 \geq 0.92$) between test and retest defined co-ordinate axes. Furthermore, whilst coronal and transverse plane waveforms also exhibited good agreement the conformity ($R^2 \geq 0.87$) was lower than those observed in the sagittal plane. This concurs with the findings of Kabada et al. (1989) who noted that coronal and transverse plane angles were affected more pointedly than the sagittal plane profiles by differences in anatomical frame axes definition.

The lowest correlations between test and retest waveforms were observed for ankle joint parameters in all three anatomical planes. It is proposed that this relates to the fact that the anatomical co-ordinate system axes of the foot was defined by placing markers directly onto the shoe which has been identified as problematic. This is because it is more difficult to palpate non visible landmarks through the shoe. Furthermore, there is almost certainly movement of the foot within the shoe (Stacoff et al. 1992), thus it is questionable as to whether anatomical markers located on the shoe provide comparable results to those placed on the foot itself. Future studies may wish to re-examine the reliability of anatomical frame definition when placing markers directly onto the foot.

With the aim of increasing the efficacy and reliability of 3-D kinematic data, researchers have also developed methods of quantifying segmental axes of rotation that are independent of anatomical landmarks. The most common is the functional method of identifying segmental parameters which has been proposed as an effective way to reduce the proposed variability of anatomical definitions (Besier et al. 2003; Della Croce et al. 1999). However, the use of markerless technology to record 3-D kinematics is still a minority technique (Richards and Thewlis, 2008) and has been limited by the intricacy of obtaining precise 3-D kinematics using this approach (Corazza et al. 2006). Future, research may wish to replicate the current investigation using markerless anatomical frame definition to further examine the efficacy of this technique.

In conclusion; based on the results obtained from this pilot study it appears that the anatomical co-ordinate axes of the lower extremities can be defined reliably. This confirms the efficacy of the findings from this thesis in relation to joint kinematics.

4.4 Pilot study 4 – Optimal filtering techniques for kinematic data

4.4.1 Introduction

Errors associated with the measurement of kinematic data can result from soft tissue artefact, improper digitization of retro-reflective markers and electrical interference. These errors are typically referred to as noise and are an undesirable portion of any kinematic waveform. Noise is traditionally lower in amplitude and associated with a different frequency range than that of the true signal and can typically be removed using a low-pass filter; the objective of any filtering technique is not only to attenuate noise but also to leave the true signal unaffected. The aim of this pilot investigation was to determine the most appropriate cut-off frequency for the filtering of kinematic data throughout the project.

4.4.2 Methods

4.4.2 (i) Participants

Ten participants (eight males and two females) (Age 26.33 ± 5.37 years, height 1.76 ± 0.12m and body mass 75.5 ± 8.60kg) ran at 4.0m.s^{-1}±5%. All were injury free at the time of data collection and provided written informed consent. Ethical approval for this project was obtained from the University of Central Lancashire School of Psychology ethics committee.

4.4.2 (ii) Procedure

Participants performed five trials using the same protocol described in pilot study 4.3.

4.4.2 (iii) 3-D kinematics

Lower extremity kinematics were obtained using the marker set and protocol outlined in chapter 5.

4.4.2 (iv) Data Processing

Hip, knee and ankle joint kinematics were quantified using Visual 3-D (C-Motion Inc, Germantown, USA) processed using raw data and also filtered with a cut-off frequency of 1Hz, 3Hz, 5Hz, 7Hz, 10Hz, 15Hz, 20Hz, 25Hz using a zero-lag low pass Butterworth 4th order filter. Lower extremity joint angles were created using an XYZ cardan sequence of rotations. All data were normalized to 100% of the stance phase then mean processed gait trial data was reported. 3-D kinematic measures from the hip, knee and ankle which were extracted for statistical analysis were 1) angle at footstrike, 2) angle at toe-off, 3) ROM from footstrike to toe-off during stance, 4) peak angle during stance, 5) relative ROM from footstrike to peak angle. FFT analyses were conducted using custom software BioProc 7.0 (Robertson, 2005) on retroreflective marker information from the stance phase in accordance with Hamill and Selbie (2005) and Winter et al. (1974) to examine the signal frequency content of the marker data (Figure 4.13). All frequency domain analyses throughout this thesis were quantified using this software on stance phase data.

4.4.2 (v) Statistical Analyses

Descriptive statistics including means and standard deviations were calculated for each condition. The statistical differences of between the filtering techniques were examined using repeated measures ANOVA's with significance accepted at the ($p \leq 0.05$) level.

4.4.3 Results

Tables 4.9-4.11 and figures 4.4 – 4.11 present the mean ± standard deviation waveforms and 3-D kinematics from the stance phase, obtained as a function of cut-off frequency.

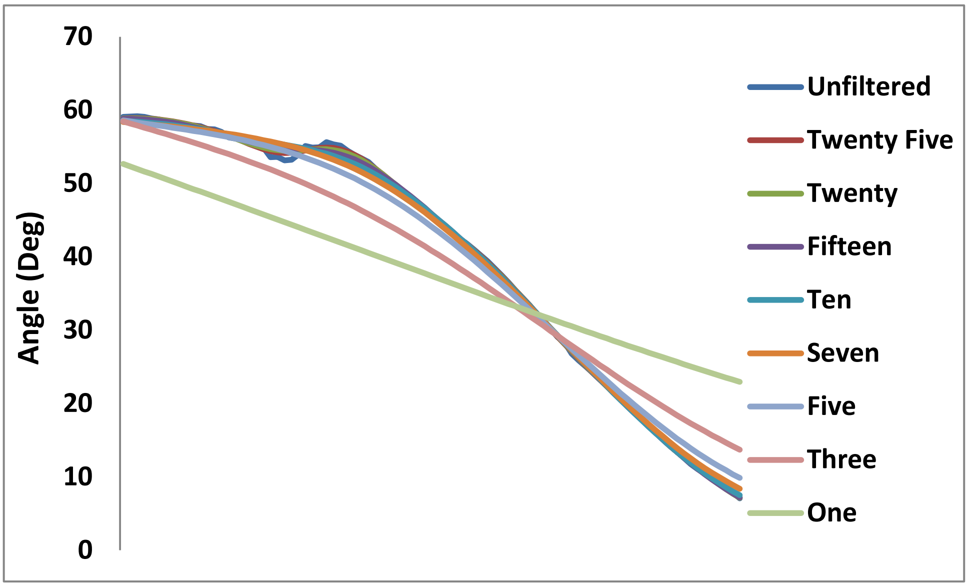

Figure 4.4: Representative stance phase hip joint kinematics in the sagittal plane as a function of cut-off frequency (Hz).

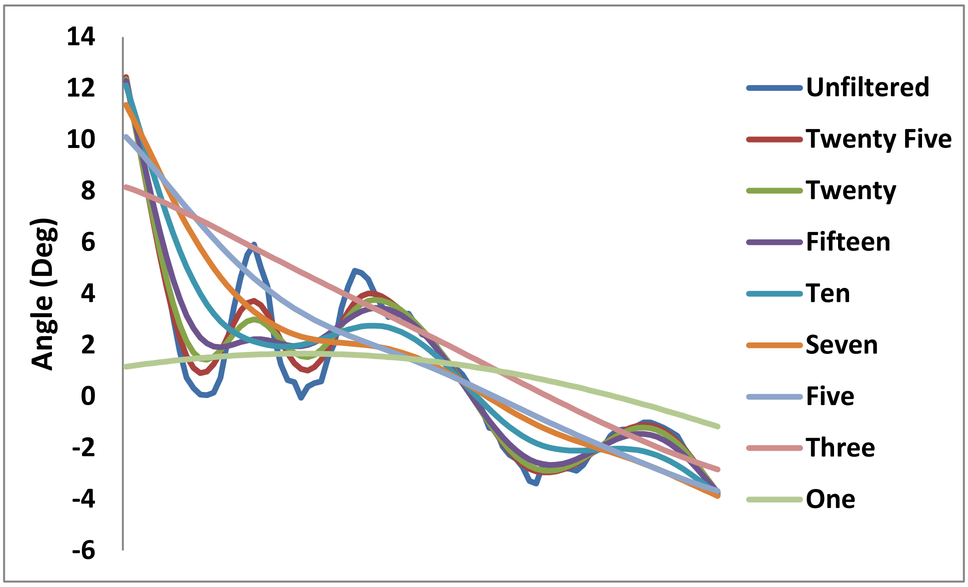

Figure 4.5: Representative stance phase hip joint kinematics in the coronal plane as a function of cut-off frequency (Hz).

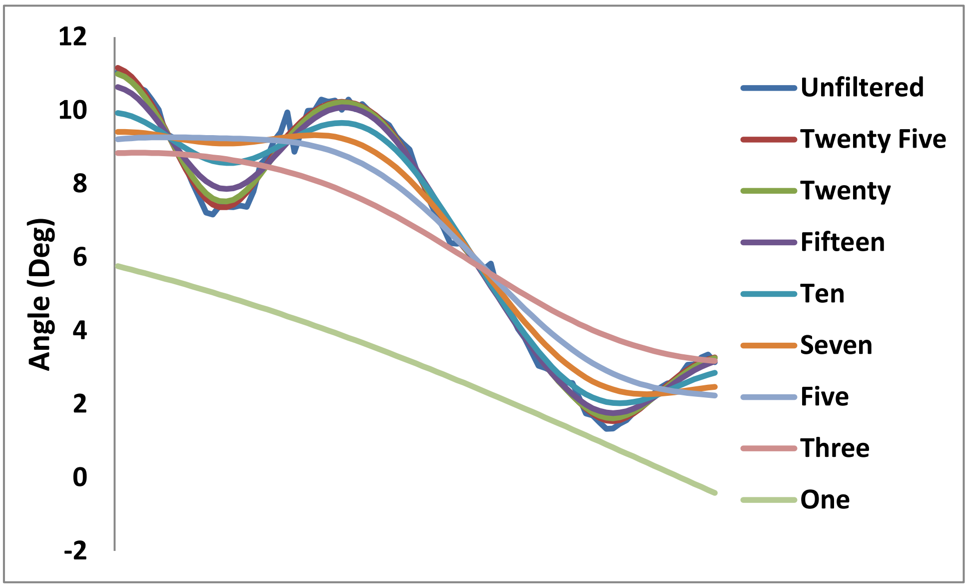

Figure 4.6: Representative stance phase hip joint kinematics in the transverse plane as a function of cut-off frequency (Hz).

In the sagittal plane significant main effects were observed for the magnitudes of rotation at footstrike $F_{(1.13, 10.16)} = 39.51$, $p \leq 0.01$, $\eta^2 = 0.81$ and toe-off $F_{(8, 72)} = 173.11$, $p \leq 0.01$, $\eta^2 = 0.95$. Furthermore, significant main effects were found for the magnitude of peak flexion $F_{(1.87, 16.83)} = 16.22$, $p \leq 0.01$, $\eta^2 = 0.54$, ROM $F_{(8, 72)} = 71.13$, $p \leq 0.01$, $\eta^2 = 0.89$ and relative ROM $F_{(1.12, 10.09)} = 7.37$, $p \leq 0.01$, $\eta^2 = 0.45$. In the coronal plane significant main effects were observed for the magnitude of peak adduction $F_{(8, 72)} = 20.21$, $p \leq 0.01$, $\eta^2 = 0.69$ and relative ROM $F_{(1.67, 14.95)} = 11.06$, $p \leq 0.01$, $\eta^2 = 0.55$. In the transverse plane significant main effects were observed for the magnitude of rotation at toe-off $F_{(8, 72)} = 10.21$, $p \leq 0.01$, $\eta^2 = 0.53$, peak external rotation $F_{(1.67, 10.50)} = 25.13$, $p \leq 0.01$, $\eta^2 = 0.74$ and relative ROM $F_{(2.32, 20.99)} = 8.02$, $p \leq 0.01$, $\eta^2 = 0.47$.

Table 4.9: Hip joint kinematics (means ± standard deviations) as a function of filtering cut-off frequency.

	Unfiltered	25	20	15	10	7	5	3	1
Hip									
X (+ = flexion/ - = extension)									
Angle at Footstrike (°)	44.05 ± 9.41	44.25 ± 9.52	44.22 ± 9.54	44.10 ±9.52	43.74 ± 9.42	43.63 ± 9.38	44.22 ± 14.45	42.45 ± 5.85	40.35 ± 12.53
Angle at Toe-off (°)	-6.08 ±13.92	-6.06 ±13.92	-6.05 ±13.92	-6.04 ±13.93	-5.94 ±13.95	-5.62 ±13.97	-4.70 ±12.06	2.87 ± 13.72	10.13 ± 12.57
Range of Motion (°)	50.14±7.40	50.31±7.58	50.28±7.57	50.15±7.54	49.68±7.42	49.24±7.24	49.02±7.20	48.83±7.52	30.23 ± 5.25
Relative Range of Motion (°)	3.93 ± 3.60	3.97±3.63	3.30 ±3.60	3.13± 3.54	2.79 ± 3.32	2.25 ±2.85	2.20 ± 2.90	0.30 ± 0.62	0.27 ± 0.85
Peak Flexion (°)	47.98±10.35	47.65±10.23	47.33±10.20	47.24±10.11	46.55±9.87	45.87±9.61	44.99±9.69	45.25±9.20	40.62±11.85
Y (+ =adduction/-=abduction)									
Angle at Footstrike (°)	2.97 ±6.26	3.06 ± 6.17	3.08 ± 6.19	3.05 ± 6.13	2.88 ± 6.07	2.79 ± 6.00	2.69 ± 6.80	3.64 ± 5.78	2.73 ± 6.52
Angle at Toe-off (°)	0.22 ± 6.46	0.21 ± 6.43	0.18 ±6.42	0.11 ± 6.39	-0.01 ± 6.31	-0.03 ± 6.23	0.40 ± 6.05	0.32 ± 6.01	0.49 ± 6.25
Range of Motion (°)	4.59 ± 2.58	4.59 ± 2.60	4.54 ± 2.59	4.46 ± 2.58	4.33 ± 2.51	4.22 ± 2.35	4.20 ± 2.40	3.83 ± 1.89	5.58 ± 3.50
Relative Range of Motion (°)	6.15 ± 3.55	5.70 ± 3.42	5.56 ± 3.32	5.32 ± 3.15	5.00 ± 2.87	4.63 ± 2.62	4.30 ± 2.71	1.92 ± 1.55	1.79 ± 2.67
Peak Adduction (°)	9.11 ± 5.21	8.76±5.33	8.63 ± 5.23	8.37 ± 5.30	7.89 ± 5.30	7.42 ±5.32	7.20 ± 4.90	5.56 ± 5.36	4.52 ± 5.96
Z (+ =internal /- =external)									
Angle at Footstrike (°)	4.28 ± 2.78	4.07 ± 7.67	4.04 ± 7.65	4.05 ± 7.62	4.18 ± 7.67	4.46 ± 7.77	4.26 ± 7.80	7.79 ± 7.66	7.15 ± 9.15
Angle at Toe-off (°)	-9.41 ±11.84	-9.42 ±11.83	-9.41± 11.79	-9.40 ±11.69	-9.37± 11.42	-9.43± 11.07	-9.27± 11.01	-9.12 ± 10.26	-4.15 ±8.37
Range of Motion (°)	14.82 ± 7.21	14.23 ± 6.96	14.19±14.08	14.08± 6.90	13.95 ± 6.55	14.04± 6.17	14.00 ± 6.12	12.90± 6.09	11.66 ± 6.87
Relative Range of Motion (°)	15.51± 6.43	15.12 ± 6.22	15.03 ± 6.27	14.85 ± 6.27	14.55 ± 6.08	14.44 ± 5.90	14.20 ± 6.07	13.92 ± 6.07	11.50 ± 7.13
Peak External rotation (°)	-11.23± 9.73	-11.05± 9.88	-10.98± 9.96	-10.80± 10.10	-9.99± 10.63	-9.60 ±10.01	-9.40± 9.60	-9.14 ± 10.25	-4.35 ±8.37

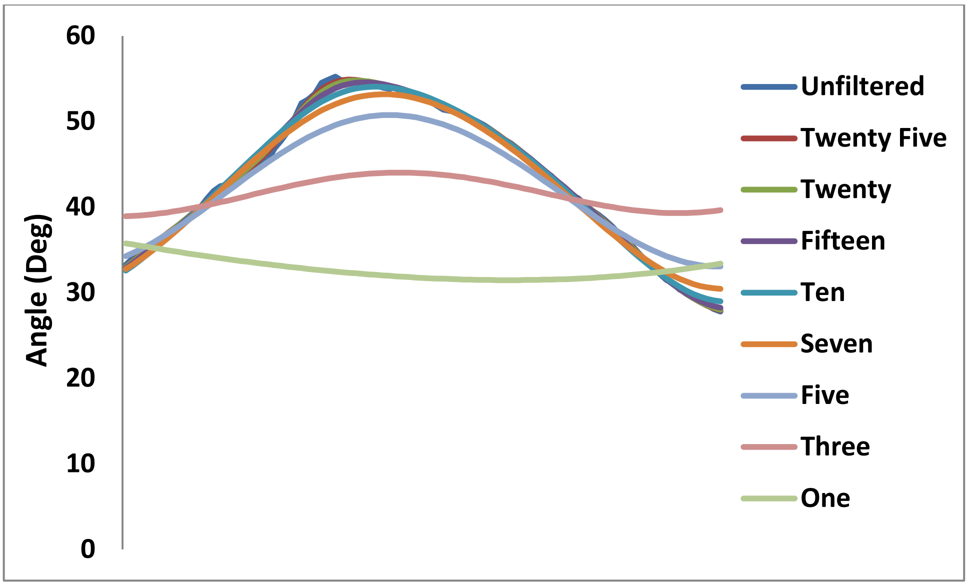

Figure 4.7: Representative stance phase knee joint kinematics in the sagittal plane as a function of cut-off frequency (Hz).

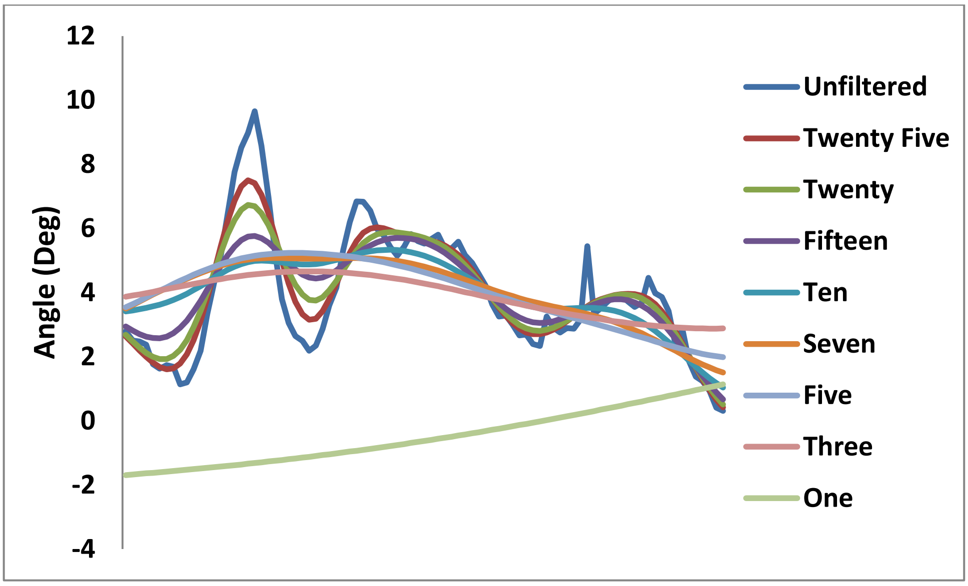

Figure 4.8: Representative stance phase knee joint kinematics in the coronal plane as a function of cut-off frequency (Hz).

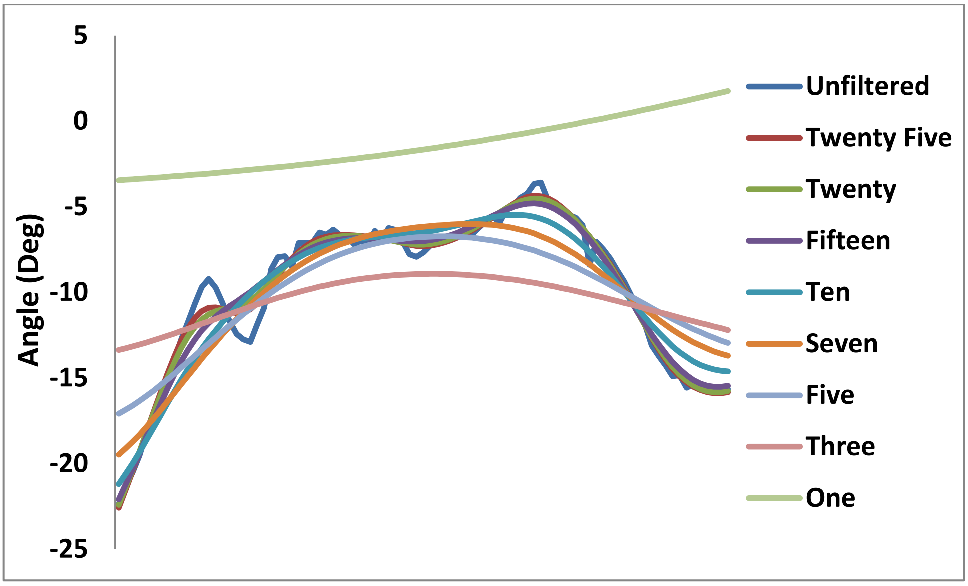

Figure 4.9: Representative stance phase knee joint kinematics in the transverse plane as a function of cut-off frequency (Hz).

In the sagittal plane significant main effects were observed for the magnitudes of rotation at footstrike $F_{(1.04, 9.34)}=5.05$, $p\leq0.05$, $\eta^2=0.36$ and toe-off $F_{(1.39, 12.46)} = 26.88$, $p\leq0.01$, $\eta^2=0.75$. Furthermore, significant main effects were found for the magnitude of peak flexion $F_{(1.16, 10.43)} = 30.06$, $p\leq0.01$, $\eta^2=0.80$, ROM $F_{(8, 72)} =71.13$, $p\leq0.01$, $\eta^2=0.89$, ROM $F_{(1.40, 12.61)} =10.02$, $p\leq0.01$, $\eta^2=0.53$ and relative ROM $F_{(1.04, 9.32)} =26.50$, $p\leq0.01$, $\eta^2=0.74$. In the coronal plane significant main effects were observed for the magnitude of rotation at toe-off $F_{(1.06, 9.57)} = 8.99$, $p\leq0.01$, $\eta^2=0.50$ and ROM $F_{(1.10, 9.87)} = 5.80$, $p\leq0.01$, $\eta^2=0.39$. In the transverse plane significant main effects were observed for the magnitude of rotation at toe-off $F_{(1.56, 13.99)} = 12.02$, $p\leq0.01$, $\eta^2=0.59$, peak internal rotation $F_{(1.35, 12.17)} = 9.89$, $p\leq0.01$, $\eta^2 =0.52$ and relative ROM $F_{(1.32, 11.86)} = 5.07$, $p\leq0.05$, $\eta^2=0.36$.

Table 4.10: Knee joint kinematics (means ± standard deviations) as a function of filtering cut-off frequency.

	Unfiltered	25	20	15	10	7	5	3	1
Knee									
X (+ = flexion/ - = extension)									
Angle at Footstrike (°)	21.49 ± 8.60	21.63 ± 8.79	21.60 ± 8.78	21.45 ± 8.72	21.06± 8.50	20.85 ± 8.24	21.31± 8.09	24.61 ± 7.94	28.14 ±14.04
Angle at Toe-off (°)	16.76 ±10.57	16.75±10.50	16.80±10.48	16.93±10.43	17.36±10.31	18.20±10.09	19.78±9.69	24.31±9.02	25.14±7.01
Range of Motion (°)	5.73±5.02	5.90±5.25	5.84±5.23	5.62±5.10	5.05±4.76	4.26±4.33	4.50±4.29	0.31±9.02	2.12±6.51
Relative Range of Motion (°)	24.13±11.03	23.51±11.06	23.49±11.02	23.57±10.97	23.76±10.99	23.47±11.07	21.60±11.29	12.99±10.47	3.66±6.62
Peak Flexion (°)	45.70±8.18	45.13±8.15	45.09±8.12	45.02±8.08	44.81±8.15	44.32±8.33	42.91±8.53	37.59±8.30	31.80±7.88
Y (+ =adduction/-=abduction)									
Angle at Footstrike (°)	1.01 ± 3.84	0.86± 3.81	0.84 ± 3.82	0.89 ± 3.83	1.12 ± 3.87	1.41 ± 3.93	1.77 ± 4.08	2.67 ± 4.66	4.03 ± 7.30
Angle at Toe-off (°)	-2.29±4.71	-2.24±4.81	-2.22±4.82	-2.14±4.82	-1.97±4.79	-1.83±4.78	-1.72±4.84	-1.19±5.08	-1.52±4.80
Range of Motion (°)	3.99±3.50	4.02±3.37	3.98±3.34	3.91±3.29	3.80±3.32	3.83±3.33	3.91±3.29	3.98±2.89	5.05±3.34
Relative Range of Motion (°)	6.46±3.60	5.20±4.02	5.04±3.98	4.84±3.92	4.61±3.85	4.51±3.82	5.88±4.55	4.39±3.14	8.31±2.97
Peak Adduction (°)	-3.81±4.68	-4.33±5.56	-4.20±5.56	3.95±5.61	-3.10±5.63	3.50±5.60	-0.91±6.04	-1.62±5.45	2.44±3.03
Z (+ =internal /- =external)									
Angle at Footstrike (°)	-14.69±10.69	-14.51±10.55	-14.55 ±10.56	-14.67±10.53	-14.67±10.39	14.25±10.23	-13.63±9.93	11.99±8.83	-14.57±10.35
Angle at Toe-off (°)	-9.83±11.15	-9.82±11.16	-9.74±11.13	-9.56±11.04	-9.05±10.76	-8.34±10.42	-7.61±10.04	-6.79±9.35	-3.23±12.31
Range of Motion (°)	6.57±4.89	6.46±4.67	6.46±4.28	6.45±5.12	6.16±5.41	5.68±5.34	6.92±5.30	5.14±4.66	10.82±9.98
Relative Range of Motion (°)	18.60±6.59	17.24±5.99	17.02±5.91	16.71±5.68	15.97±4.82	14.78±3.84	13.18±3.32	9.23±3.29	12.70±10.28
Peak Internal rotation (°)	3.91±8.22	2.73±8.26	2.47±8.31	2.04±8.43	1.31±8.63	0.53±8.61	-0.45±8.55	-2.76±8.60	-1.87±12.71

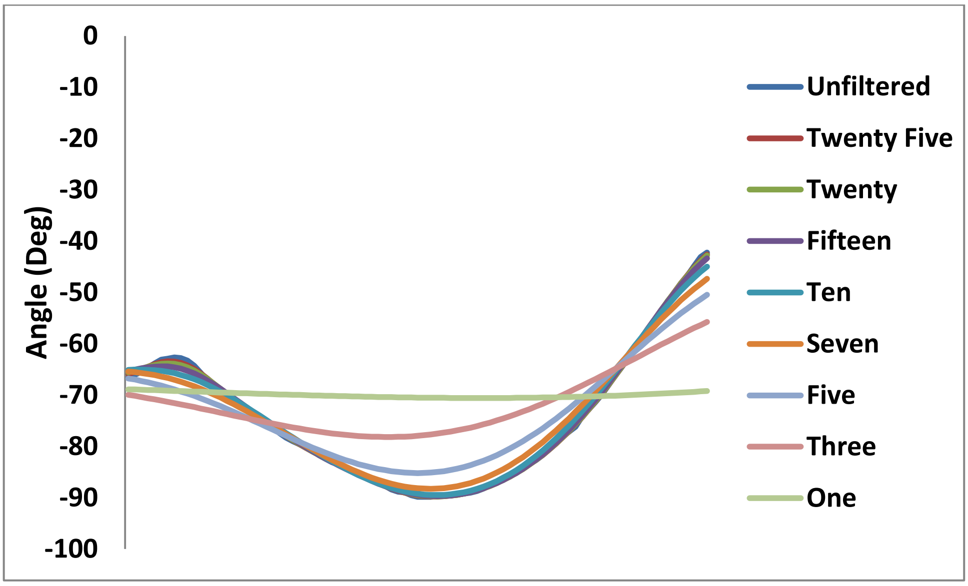

Figure 4.10: Representative stance phase ankle joint kinematics in the sagittal plane as a function of cut-off frequency (Hz).

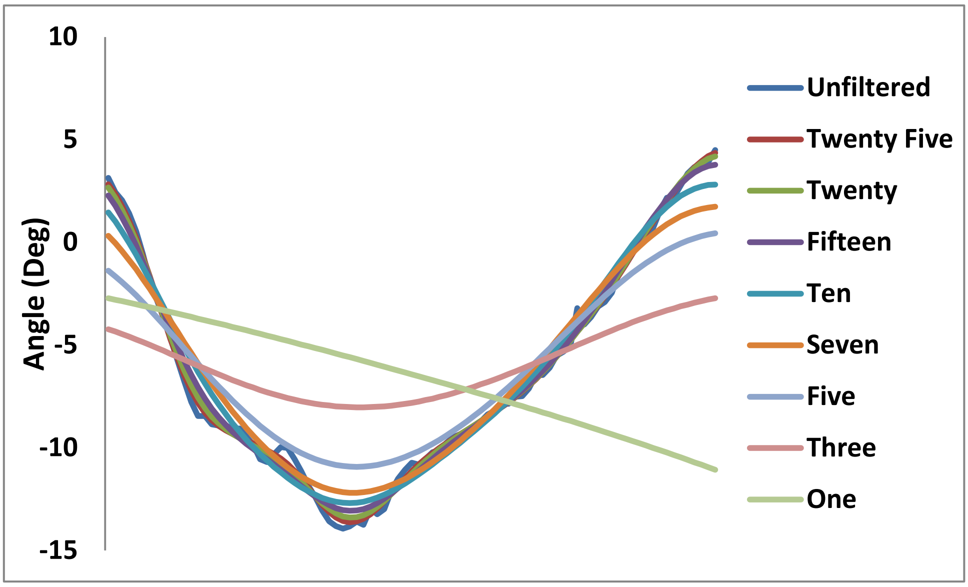

Figure 4.11: Representative stance phase ankle joint kinematics in the coronal plane as a function of cut-off frequency (Hz).

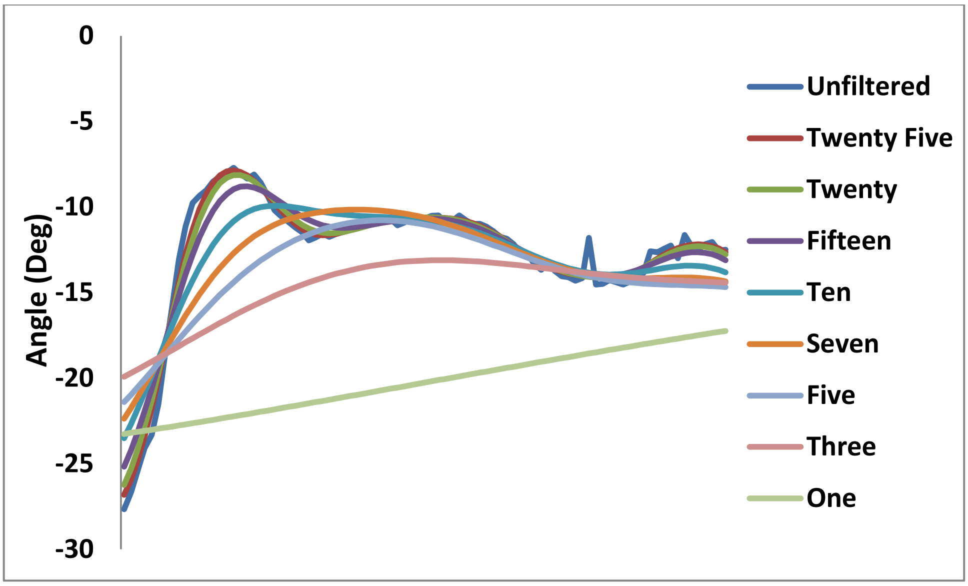

Figure 4.12: Representative stance phase ankle joint kinematics in the transverse plane as a function of cut-off frequency (Hz).

In the sagittal plane significant main effects were observed for the magnitudes of rotation at toe-off $F_{(1.06, 9.57)} = 8.99$, $p \leq 0.01$, $\eta^2=0.50$ and ROM $F_{(1.10, 9.87)} =9.87$, $p \leq 0.05$, $\eta^2=0.53$. In the coronal plane significant main effects were observed for the magnitude of rotation at toe-off $F_{(1.10, 9.90)} = 17.51$, $p \leq 0.01$, $\eta^2=0.66$ and peak eversion $F_{(1.40, 12.32)} = 7.31$, $p \leq 0.01$, $\eta^2=0.45$. Furthermore, a significant main effect was observed for the magnitude of relative ROM $F_{(1.60, 14.34)} = 7.61$, $p \leq 0.01$, $\eta^2=0.46$. In the transverse plane significant main effects were observed for the magnitude of rotation at footstrike $F_{(1.30, 11.72)} = 5.20$, $p \leq 0.05$, $\eta^2=0.42$, peak rotation $F_{(1.19, 10.23)} = 10.95$, $p \leq 0.01$, $\eta^2 =0.55$ and relative ROM $F_{(1.56, 13.16)} = 9.62$, $p \leq 0.01$, $\eta^2=0.52$.

Table 4.11: Ankle joint kinematics (means ± standard deviations) as a function of filtering cut-off frequency.

	Unfiltered	**25**	**20**	**15**	**10**	**7**	**5**	**3**	**1**
Ankle									
X (+ = Plantar/ - = dorsi)									
Angle at Footstrike (°)	-67.10±30.45	-66.80±30.38	-66.59±30.35	-66.20±30.30	-65.39±30.20	-64.66±30.14	-64.53±30.18	-65.79±29.81	-68.95±36.59
Angle at Toe-off (°)	-35.87±26.26	-35.96±26.21	-36.04±26.23	-36.28±26.33	-37.09±26.66	-38.52±27.06	-40.66±27.40	-45.32±27.38	-59.91±30.82
Range of Motion (°)	32.15±8.50	31.76±8.46	30.50±8.42	30.87±8.29	28.16±7.81	27.78±7.29	25.05±6.74	20.85±6.30	27.75±18.09
Relative Range of Motion (°)	11.25±5.04	11.34±5.04	11.48±5.04	11.79±4.96	12.43±4.59	12.33±4.03	10.15±3.77	3.97±2.73	7.33±11.80
Peak Dorsi-Flexion (°)	-78.35±29.64	-78.11±29.64	-78.07±29.70	-77.99±29.68	-77.82±29.58	-76.99±29.38	-74.67±29.06	-69.75±28.72	-76.78±32.42
Y (+ =inversion/-= eversion)									
Angle at Footstrike (°)	-4.78±4.37	-4.73±4.33	-4.67±4.28	-4.59±4.17	-4.77±4.00	-5.38±3.90	-6.11±3.81	-7.08±3.92	-8.62±10.75
Angle at Toe-off (°)	2.59±5.65	2.47±5.61	2.38±5.59	2.15±5.65	1.55±5.40	0.78±5.12	-0.09±4.82	-2.32±5.02	-11.44±7.72
Range of Motion (°)	8.60±4.28	8.42±4.07	8.26±3.96	7.94±3.85	7.49±3.87	7.19±3.87	6.91±3.66	5.37±2.60	3.89±7.75
Relative Range of Motion (°)	13.20±4.83	12.47±4.83	12.95±4.82	12.71±4.82	11.01±4.17	10.12±3.35	8.94±2.33	4.98±1.98	7.70±7.77
Peak eversion (°)	-17.80±5.16	-17.40±519	-17.22±5.23	-16.86±5.24	-16.18±4.94	-15.43±4.42	-14.31±3.80	-11.69±3.21	-15.93±9.14
Z (- =internal /+ =internal)									
Angle at Footstrike (°)	-16.55±6.65	-16.44±6.39	-16.32±6.24	-16.00±5.94	-15.33±5.50	-14.68±5.23	-14.03±4.84	-14.03±4.98	-13.07±4.80
Angle at Toe-off (°)	-11.73±5.11	-11.71±5.09	-11.77±5.08	-11.92±5.06	-12.24±4.98	-12.48±4.74	-12.56±4.36	-12.39±4.78	-8.25±8.46
Range of Motion (°)	6.62±4.56	6.29±4.33	6.15±4.09	5.78±3.66	4.85±2.90	4.11±2.58	3.39±2.27	3.09±1.70	9.91±10.88
Relative Range of Motion (°)	12.74±4.08	11.78±3.62	11.41±3.43	10.97±3.09	9.58±2.81	8.61±2.75	1.38±1.69	5.28±1.62	7.59±10.46
Peak External rotation (°)	-3.81±4.68	-4.66±4.32	-4.91±4.22	-5.60±4.21	-5.76±4.26	-6.07±4.36	-15.42±5.21	-7.78±4.37	-5.45±8.87

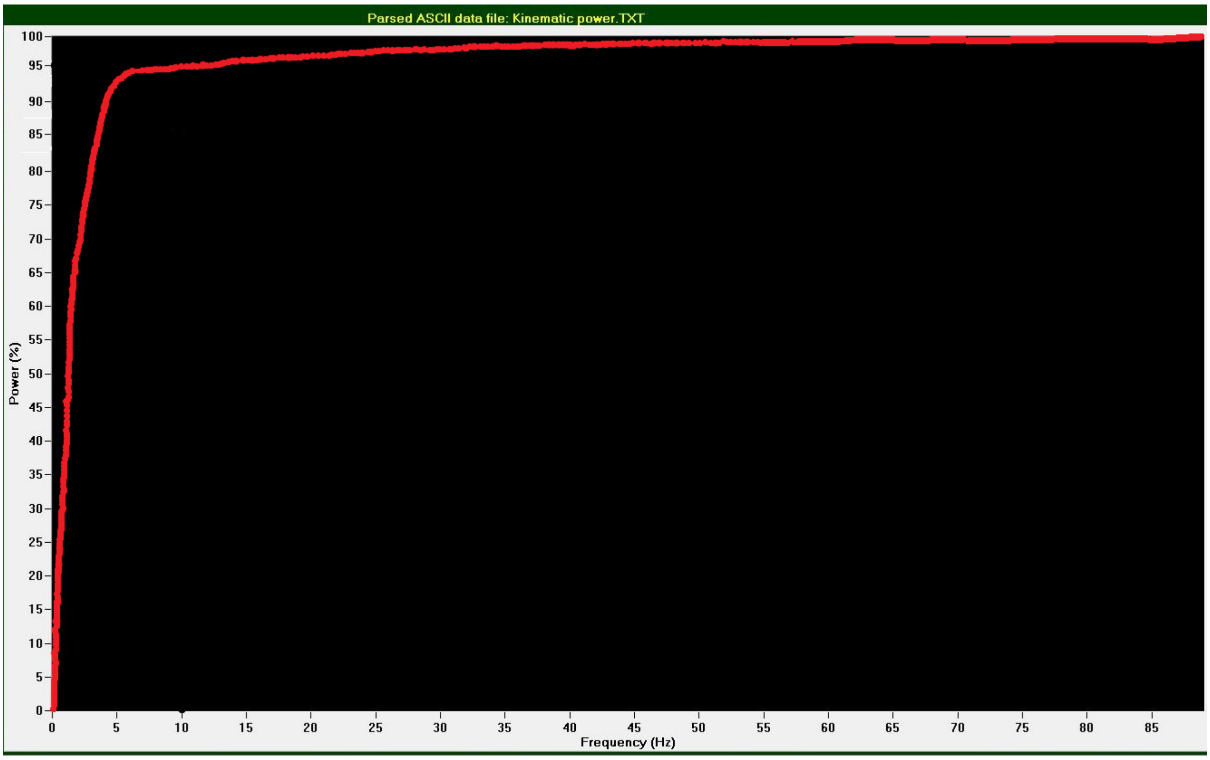

Figure 4.13: Representative FFT power analysis of marker information.

4.4.4 Discussion

It is clear from this pilot investigation that different cut-off frequencies can significantly affect the lower extremity kinematic outcomes in all three planes of rotation. Therefore selecting the appropriate cut-off frequency is of critical importance in order to achieve empirically meaningful findings in the main studies. Observation of the lower extremity angle time curves evidences both over and under smoothing of kinematic data. The 10Hz cut-off frequency is the highest cut-off that appears qualitatively to have no evidence of noise and is substantiated by the FFT analysis as being the frequency at which 95% of the signal power is maintained. As such the 10Hz cut-off frequency appears to be the optimal method for this project and will be used throughout.

4.5 Pilot study 5 - Determining the most appropriate cut-off frequency for filtering of tibial acceleration signal.

4.5.1 Introduction

It is recognised that tissue artefact interferes with acceleration signal recording of the underlying bone (Light et al. 1980). Direct attachment of the device to the bone using Steinmann/kirscher pins is considered the most effective method of measuring skeletal transients during gait, although this technique is not used extensively due to its invasiveness. Therefore, the less invasive skin mounting technique is typically used and the component of the signal associated with tissue resonance is removed using a low-pass filter. The aim of this pilot investigation was to determine the most appropriate cut-off frequency for the acceleration signal during running.

4.5.2 Methods

4.5.2 (i) Participants

Twelve participants (eight male and four female) Age 27.31years ± 3.43, height 1.79 ± 0.14m and mass 77.7 ± 6.86kg volunteered for this investigation, all were injury free and provided written consent. Ethical approval for this project was obtained from the University of Central Lancashire School of Psychology ethics committee.

4.5.2 (ii) Procedure

Participants performed five trials using the same protocol described in pilot study 4.3.

4.5.2 (iii) Tibial acceleration

Tibial accelerations were obtained using the protocol outlined in chapter 5.

4.5.2 (iv) Data processing

The acceleration signal was filtered using a Butterworth zero lag 4th order low-pass filter at 20, 40, 60, 80, 100, 120, 140 and 160Hz. Peak positive axial tibial acceleration was defined as the highest positive acceleration peak measured during the stance phase. A FFT was also run on tibial acceleration information from the stance phase in accordance with Lafortune and Hennig, (1995) to examine the signal frequency content.

4.5.2 (v) Statistical Analysis

Differences in peak tibial shock were compared using a repeated measures ANOVA with significance at $p \leq 0.05$. Post-hoc analysis was conducted using a Bonferroni correction to control for type I error.

4.5.3 Results

Table 4.12 and figure 4.14 present the mean ± standard deviation peak tibial acceleration magnitudes and tibial acceleration signals as a function of each cut-off frequency.

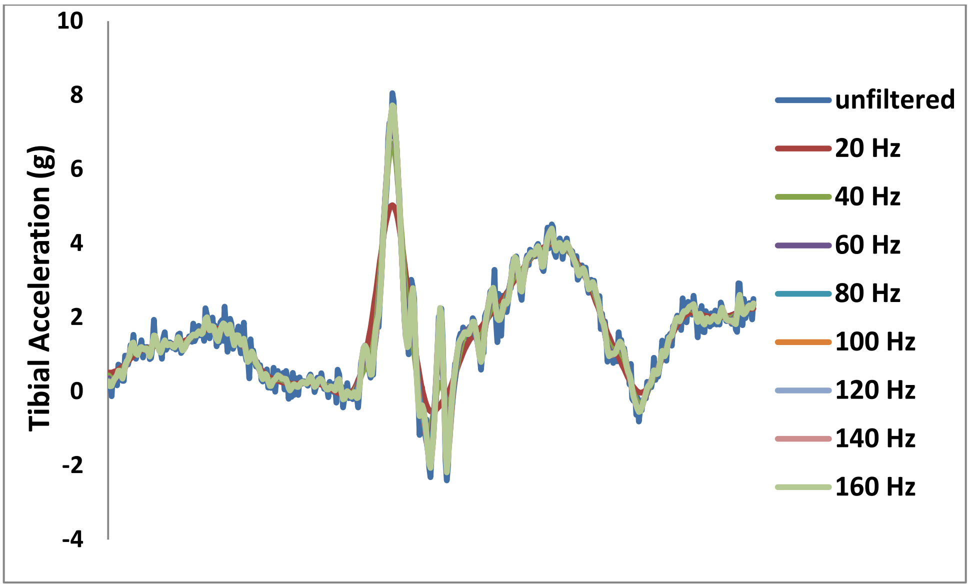

Figure 4.14: Representative tibial acceleration signals as a function of cut-off frequency.

Table 4.12: Tibial shock (mean and standard deviation) magnitude as a function of cut-off frequency (* = significant main effect.)

	Peak Impact Shock (g)
Unfiltered	7.61 $\pm$ 3.51
20 Hz	3.57 $\pm$ 0.92
40 Hz	5.95 $\pm$ 2.30
60 Hz	6.73 $\pm$ 2.93
80 Hz	7.07 $\pm$ 3.20
100Hz	7.24 $\pm$ 3.33
120 Hz	7.32 $\pm$ 3.41
140 Hz	7.37 $\pm$ 3.50
160 Hz	7.39 $\pm$ 3.46
	*

The results indicate that a significant main effect $F_{(1.08, 16.19)} = 44.91$, $p \leq 0.01$, $\eta^2 = 0.75$ exists for the magnitude of peak tibial shock as a function of cut-off frequency. Post-hoc analysis revealed that each cut-off frequency differed significantly $p \leq 0.05$ from the others.

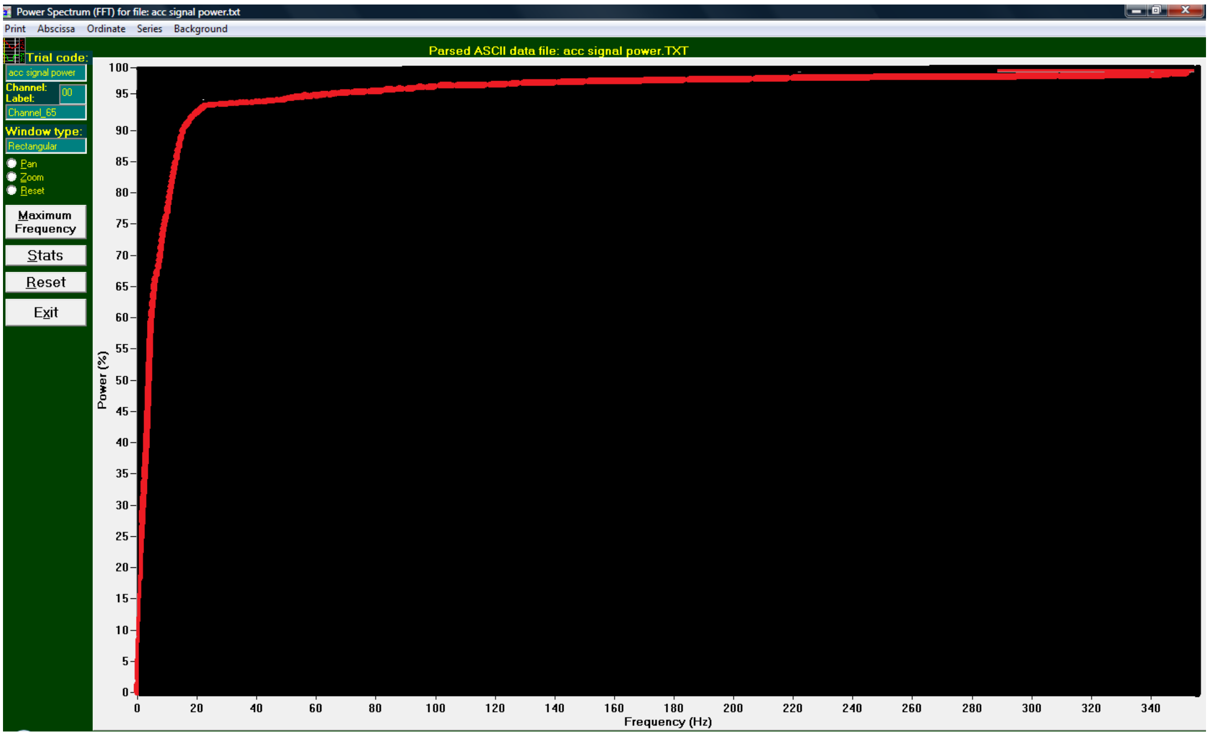

Figure 4.15: Representative FFT power analysis of tibial acceleration information.

4.5.4 Discussion

The results of the current pilot investigation indicate that alterations in cut-off frequency significantly influenced the magnitude of peak tibial shock. Therefore, selecting the most appropriate cut-off frequency is important when investigating tibial shock magnitude between conditions. Hennig and Lafortune, (1991) recommend a cut-off frequency of 60Hz based on their observation that 95% of the signal power being below this frequency. The results of this pilot investigation correspond with this finding, and as such it was determined that this cut-off frequency is the most appropriate for tibial acceleration.

4.6 Pilot Study 6 – The influence of different cardan sequences on lower extremity joint kinematics.

4.6.1 Introduction

Cardan/Euler rotations of a rigid segment with respect to another are obtained using an ordered sequence of rotations (Schache et al. 2001). Rotations are considered to occur about the axis of the segment co-ordinate system. For example during an XYZ cardan sequence of rotations, the segment is rotated about the X axes by an angle α, it is then rotated about a rotated Y' by an angle β and then finally rotated about a twice rotated Z'' axes by an angle γ.

For a given motion, different cardan sequences can influence the angular calculations (Cole et al. 1993). The International Society of Biomechanics (ISB) recommends that joint angles for the lower limbs should be calculated using an XYZ sequence of rotations (Cole et al. 1993). However, it has been proposed that the XYZ sequence when applied to rotations outside the sagittal plane may not be the most appropriate method of quantifying coronal and transverse plane kinematics.

In addition to the commonly used Cardan/euler method, helical angles can also be used to describe the position of one reference system with respect to another (Woltring et al. 1985). Using this technique, a position vector and an orientation vector are defined and movement from a reference position is described in terms of rotation along a single projected axis. This method is considered to be stable over any conceivable joint motion, yet it is utilized infrequently as angular motion using this technique may not correspond with an anatomical representation that is clinically meaningful (Hamill and Selbie, 2004).

A selected number of investigations have examined the influence of segmental kinematic calculations on the representation of angular profiles during gait (Tupling and Pierrynowski, 1987; Woltring, 1991; Kardura et al. 2000; Schache et al. 2001; Thewlis et al. 2008; Lees et al. 2010). Despite this, most the appropriate method for the determination of lower extremity joint kinematics during running remains unknown. This pilot study therefore investigates the influence of the Helical method and six available cardan sequences on lower extremity joint planar crosstalk and kinematic parameters in the sagittal, coronal, and transverse planes.

4.6.2 Method

4.6.2 (i) Participants

Twelve male participants volunteered to take part in this investigation (age = 19.25 ± 1.55 years; height = 1.77 ± 0.52m; mass = 78.4 ± 9.0 kg). All were injury free at the time of data collection and provided written informed consent. Ethical approval for this project was obtained from the School of Psychology ethics committee.

4.6.2 (ii) Procedure

Participants performed five trials using the same protocol described in pilot study 4.3.

4.6.2 (iii) 3-D kinematics

Lower extremity kinematics were obtained using the marker set and protocol outlined in chapter 5.

4.6.2 (iv) Data processing

Lower extremity kinematic parameters of peak angles and ranges of motion were quantified using Visual 3-D (C-Motion Inc, Germantown, USA) and filtered at 10 Hz using a zero-lag low pass Butterworth 4th order filter. Angles were created using the helical method and about XYZ, ZXY, XZY, YXZ, YZX, and YXZ rotation cardan sequences.

4.6.2 (v) Statistical Analyses

Descriptive statistics including means and standard deviations were calculated for each condition. Differences in stance phase peak angles and ranges of motion were examined using repeated measures ANOVAs with significance accepted at the $p \leq 0.05$ level. Appropriate post-hoc analyses were conducted using a Bonferroni correction to control for type I error. Intra-class correlations were also utilized to compare sagittal, coronal, and transverse plane waveforms using the seven different methods. Furthermore, sagittal plane angles from all three joints were also correlated with the associated coronal and transverse plane waveforms in order to identify evidence of planar crosstalk. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc., Chicago, IL, USA).

4.6.3 Results

Tables 4.13-4.18 and figure 4.16 present the mean ± standard deviation kinematic waveforms and 3-D kinematic parameters obtained as a function of the helical and six available cardan sequences.

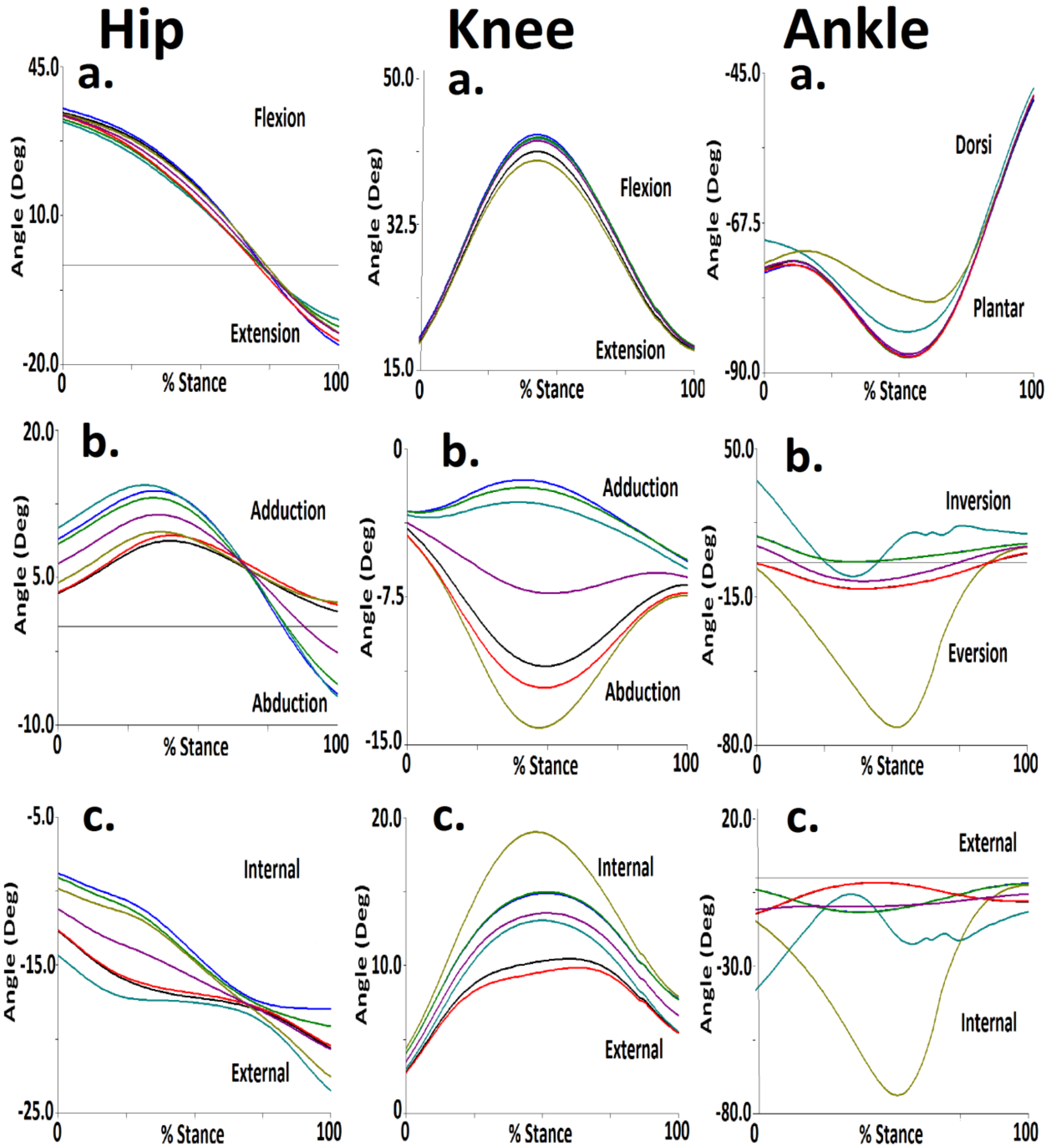

Figure 4.16: Hip, knee and ankle joint kinematics in the a. sagittal, b. coronal and c. transverse planes as a function of cardan sequence (XYZ= black, XZY= red, YXZ=yellow, YZX=blue, ZXY= cyan, ZYX=green and Helical =purple).

Table 4.13: Mean (and standard deviation) hip ROM for each rotation as a function of cardan sequence (* = significant main effect).

	X	**Y**	**Z**
XYZ (°)	40.70±26.06	7.88±2.27	5.56±5.07
XZY (°)	36.63±29.71	7.07±2.09	5.48±4.99
YXZ (°)	37.23±30.09	8.73±2.11	11.05±9.27
YZX (°)	40.46±30.09	11.17±2.78	10.49±11.47
ZXY (°)	33.59±26.31	10.28±2.40	9.61±1.110
ZYX (°)	34.78±23.88	9.38±2.27	9.18±8.51
Helical (°)	43.27±22.90	8.49±8.01	8.02±6.45
Main Effect			

Table 4.14: Mean (and standard deviation) peak hip angle for each rotation as a function of cardan sequence (* = significant main effect).

	X	**Y**	**Z**
XYZ	28.96±13.45	9.08±6.57	-21.38±16.32
XZY	22.79±16.23	10.60±7.06	-19.55±14.04
YXZ	25.62±14.55	13.23±9.77	-13.60±9.28
YZX	25.89±14.35	14.83±9.23	-5.93±13.60
ZXY	25.37±13.88	15.67±9.58	-25.09±22.95
ZYX	24.03±15.22	13.99±8.56	-6.68±11.70
Helical	29.91±14.20	10.72±6.13	-19.49±14.96
Main Effect			*

It was observed that a peak hip angle main effect was found for the transverse plane $F_{(5, 50)} = 17.86$, $p \leq 0.01$, $\eta^2 = 0.64$. Post-hoc analyses revealed that peak values using the YXZ and ZXY sequences were significantly greater than the others. Comparisons between hip angles using the seven different methods revealed very strong correlations for the sagittal plane (R^2=0.99) and moderate-strong correlations for the transverse (R^2=0.75) plane. However, comparisons between the methods in the coronal plane revealed only moderate correlations (R^2=0.55). When coronal and sagittal plane angles were correlated, very low correlations were observed when using the helical (R^2= 0.10) XYZ (R^2= 0.025), XZY (R^2= 0.094), YXZ (R^2=0.04), YZX (R^2= 0.03), and ZYX (R^2= 0.03) techniques indicating little crosstalk. However, when the ZXY (R^2= 0.31) sequence were used there was evidence of crosstalk. When transverse and sagittal plane angles were correlated, very low correlations were observed when using the helical (R^2= 0.06) XYZ (R^2= 0.04), XZY (R^2= 0.04), YXZ (R^2= 0.07), YZX (R^2= 0.07) and ZYX (R^2= 0.04) techniques indicating little crosstalk. However, when the ZXY (R^2= 0.15) sequence was used there was evidence of planar crosstalk.

Table 4.15: Mean (and standard deviation) knee ROM for each rotation as a function of cardan sequence (* = significant main effect).

	X	Y	Z
XYZ	3.32±3.89	1.85±2.51	2.31±2.72
XZY	4.16±2.88	1.98±2.72	2.33±2.69
YXZ	4.22±2.95	3.16±4.33	3.21±2.40
YZX	4.29±2.81	1.80±2.70	3.53±2.27
ZXY	3.75±4.09	1.43±4.29	1.81±3.66
ZYX	4.09±2.79	1.70±2.61	3.45±2.16
Helical	6.17±4.97	2.26±2.03	4.38±2.83
Main Effect			

Table 4.16: Mean (and standard deviation) peak knee angle for each rotation as a function of cardan sequence (* = significant main effect).

	X	Y	Z
XYZ	45.73±13.91	-3.09±4.57	10.79±14.57
XZY	49.40±12.11	-13.03±8.86	10.21±13.27
YXZ	43.55±12.75	-33.10±28.10	31.14±26.37
YZX	47.77±12.46	-1.79±5.13	15.87±13.64
ZXY	44.94±9.35	-9.47±10.52	11.92±17.72
ZYX	47.32±12.55	-2.27±3.56	15.99±13.82
Helical	42.54±5.45	-8.25±3.47	15.92±14.01
Main Effect		*	*

It was observed that peak angle main effects were found for the coronal $F_{(5, 50)} = 5.27$, $p \leq 0.05$, $\eta^2=0.35$ and transverse planes $F_{(5, 50)} = 5.60$, $p \leq 0.05$, $\eta^2= 0.36$. Post-hoc analyses revealed that peak values using the YXZ sequence were significantly greater than the others. Comparisons between knee angles using the seven different methods revealed very strong correlations for the sagittal plane ($R^2=0.96$) and moderate-strong correlations for the transverse ($R^2=0.79$) plane. However, comparisons between the methods in the coronal plane revealed only moderate correlations ($R^2=0.55$). When coronal and sagittal plane angles were correlated, very low correlations were observed when using the helical ($R^2= 0.02$) XYZ ($R^2= 0.01$), XZY ($R^2= 0.02$), YXZ ($R^2= 0.08$), YZX ($R^2= 0.03$), ZXY ($R^2= 0.03$) and ZYX ($R^2= 0.01$) techniques indicating little crosstalk. However, when the YXZ ($R^2= 0.14$) sequence was used there was evidence of crosstalk. When transverse and sagittal plane angles were correlated, very low correlations were observed when using the helical ($R^2= 0.01$), XYZ ($R^2= 0.01$), XZY ($R^2= 0.04$), YZX ($R^2= 0.02$), ZXY ($R^2=0.06$) and ZYX ($R^2= 0.04$) techniques indicating little crosstalk. However, when the YXZ ($R^2= 0.18$) sequence was used there was evidence of planar crosstalk.

Table 4.17: Mean (and standard deviation) ankle ROM for each rotation as a function of cardan sequence (* = significant main effect).

	X	Y	Z
XYZ	28.30±10.76	5.72±3.77	4.42±3.31
XZY	28.11±11.09	7.25±2.74	4.89±4.63
YXZ	26.85±11.36	18.23±11.34	14.37±10.04
YZX	27.69±11.29	5.85±5.72	3.72±2.02
ZXY	25.89±8.59	26.41±17.64	28.48±12.39
ZYX	27.86±11.90	5.83±5.67	3.69±2.13
Helical	25.07±8.88	5.26±2.33	5.30±2.45
Main Effect		*	*

Table 4.18: Mean (and standard deviation) peak ankle angle for each rotation as a function of cardan sequence (* = significant main effect).

	X	Y	Z
XYZ	-87.14±2.42	-9.46±4.21	-2.66±3.84
XZY	-87.37±2.38	-9.57±4.22	-0.60±4.89
YXZ	-85.86±2.92	-67.65±19.17	-71.06±16.64
YZX	-87.34±2.42	1.96±4.46	-10.27±3.58
ZXY	-85.42±1.76	2.25±18.21	2.54±14.46
ZYX	-87.16±2.38	1.99±4.39	-10.33±3.59
Helical	-88.26±3.21	-5.66±5.77	-11.39±3.16
Main Effect		*	*

The results indicate that significant ROM main effects were observed for the coronal $F_{(1.59, 15.90)} = 14.23$, $p \leq 0.01$, $\eta^2 = 0.58$ and transverse plane $F_{(2.01, 20.14)} = 25.29$, $p \leq 0.01$, $\eta^2 = 0.72$. Furthermore, it was also observed that peak angle main effects were found for the coronal $F_{(2.09, 18.79)} = 78.94$, $p \leq 0.01$, $\eta^2 = 0.90$ and transverse planes $F_{(1.98, 17.85)} = 82.13$, $p \leq 0.01$, $\eta^2 = 0.90$. Post-hoc analyses revealed that ROM and peak values using the YXZ sequence were significantly greater than the others. Comparisons between ankle angles using the seven different methods revealed very strong correlations for the sagittal plane $R^2=0.99$ and moderate-strong correlations for the transverse $R^2=0.75$ plane. However, comparisons between methods in the coronal plane revealed only moderate correlations $R^2=0.55$. When coronal and sagittal plane angles were correlated, very low correlations were observed when using the helical ($R^2 = 0.019$) XYZ ($R^2 = 0.01$), XZY ($R^2 = 0.015$), YZX ($R^2 = 0.011$), and ZYX ($R^2 = 0.011$) techniques indicating little crosstalk. However, when the YXZ ($R^2 = 0.18$) and ZXY ($R^2 = 0.15$) sequences were used there was evidence of crosstalk. When transverse and sagittal plane angles were correlated, very low correlations were observed when using the helical ($R^2 = 0.002$) XYZ ($R^2 = 0.01$), XZY ($R^2 = 0.01$), YZX ($R^2 = 0.013$), ZXY ($R^2 = 0.04$), and ZYX ($R^2 = 0.014$) techniques indicating little crosstalk. However, when the YXZ ($R^2 = 0.15$) sequence was used there was evidence of planar crosstalk.

4.6.4 Discussion

Euler/Cardan angles are used extensively within the fields of clinical and sport biomechanics. To date, the effect of altering the sequence of rotations has yet to be fully investigated with respect to the lower extremities. This pilot investigation aimed to determine the optimal cardan sequence for the representation of lower extremity kinematics.

The results indicate that altering the sequence of rotations when observing kinematics in the sagittal plane has no significant effect on lower extremity joint angular parameters. This is unsurprising given the dominance of sagittal plane motion during running gait (Novacheck, 1998). This concurs with the majority of literature with regard to sequence-dependent angles (Areblad et al. 1990; Thewlis et al. 2008) as the coronal and transverse plane movements are small in comparison and thus the potential for planar crosstalk is minimal. It leads to the conclusion that selecting the appropriate sequence of rotations is not an issue when investigating kinematics in the sagittal plane.

However, in the coronal and transverse planes, significant main effects were observed in terms of both the ROM and the peak angle observed during the stance phase. The results of this pilot study with respect to the lower extremity joint found that the ZXY and YXZ sequences significantly affected lower extremity joint kinematics, producing extremely large values for both ROM and peak angles. Furthermore, when coronal and transverse plane profiles were correlated with the sagittal plane, the strongest correlations were observed for the YXZ and ZXY rotation sequences indicating that they are most susceptible to planar crosstalk. This concurs with the observations of Lees et al. (2010), who observed that these sequences were associated with the greatest degree of error. The key implication of this finding is that the YXZ and ZXY sequences differed in both magnitude and profile, and the error associated with these sequences is such that the kinematic estimates are anatomically unrealistic. Thus, it leads to the conclusion that these sequences cannot be utilized to accurately interpret lower extremity kinematics outside of the sagittal plane.

With respect to the helical axis technique, its lack of sequence dependency and ability to attenuate gimbal lock have led researchers to suggest that it may serve as an alternative to the Cardanic method (Hamill and Selbie, 2004). However, the susceptibility of this method to measurement error and sensitivity to the magnitude of rotation (Woltring et al. 1985), in conjunction with the inability to define a meaningful anatomical reference frame, suggest that the representation of segmental motion may be compromised. Thus, it leads to the conclusion that the limited utilization of this technique may be justified.

It is interesting to note that the two combinations that were found to be associated with the greatest amount of crosstalk (YXZ and ZXY) each had X second in the order of rotations. This was the case even when the principal axis under investigation is placed first. However, when the coronal and transverse plane profiles are observed it is apparent that peak angles occur at or around maximum flexion/dorsiflexion. Thus, it appears to support the existence of planar crosstalk, and concurs with the findings of Blankevoort et al. (1988), Kadaba et al. (1990), Thewlis et al. (2008), and Lees et al. (2010). However, when X is placed last in the order of rotations it has little effect on the magnitude of the angular values, and the coronal and transverse plane joint profiles appear to be independent to the movement in the sagittal plane. These results appear to oppose those reported by Areblad et al. (1990), who reported that altering the sequence of rotations has only a small influence on the angular calculations.

It is clear from the results of this pilot investigation that different computational methods can yield different angular kinematic patterns. Observation of the angular profiles and statistical data suggests that using the XYZ sequence to calculate coronal and transverse plane kinematics appears to cause minimal crosstalk from the sagittal plane. Based on these results, it appears that at the current time the ISB recommendations are the most appropriate for the representation of lower extremity kinematics during the stance phase of running, and as such it will be utilized throughout this thesis.

4.7 Pilot study 7: The variability in 3-D running kinematics caused by different force measuring devices: Determination of the optimal device.

4.7.1 Introduction

The analysis of running kinematics is commonly undertaken by having participants make contact with a force measuring device. A number of analyses simultaneously examine running kinematics in conjunction with kinetics. In running studies where both kinetic and kinematic information is required, problems may arise if participants have to modify their gait patterns in order to ensure contact with the device (Challis, 2001). Such deliberate modification of the natural gait pattern is referred to as targeting. Whilst previous analyses have examined the influence of targeting on GRF parameters, there is currently a paucity of information regarding the influence of targeting on 3-D lower extremity kinematics and how different force measuring devices may affect the extent to which targeting takes place. The aim of the current pilot investigation was to examine the influence of different force measuring transducers on targeting.

4.7.2 Methods

4.7.2 (i) Participants

Fifteen male participants volunteered to take part in this investigation. All were injury free at the time of data collection and provided written informed consent. The mean characteristics of the participants were; age 24.2 ± 5.4 years, height 1.74 ± 0.07m and body mass 72.4 ± 6.6 kg. Ethical approval for this project was obtained from the University of Central Lancashire School of Psychology ethics committee.

4.7.2 (ii) Procedure

Participants ran at 4.0m.s^{-1} in four different conditions 1). over an embedded piezoelectric force plate (Kistler, Kistler Instruments Ltd., Alton, Hampshire), 2). over an RS scan mat overlaying the force platform, 3). over a Tek Scan mat overlaying the force platform and 4). uninhibited running through the testing area without concern for striking a force measuring transducer. During the practice trials, the starting position of the run and the position of the fourth footfall were recorded allowing the starting position to be repeatable in that participants struck the force transducer on their fourth footstrike.

Running velocity was quantified using infrared timing gates Newtest 300 (Newtest, Oy Koulukatu, Finland), a maximum deviation of ±5% from the set velocity was allowed. Participants completed a minimum of five successful trials in each condition. A successful trial was defined as one within the specified velocity range, where all tracking clusters were in view of the cameras. Because force information was not available for each condition footstrike and toe-off was determined using the Dingwell et al. (2001) technique as described in chapter 5. The order in which participants ran in each condition was randomized.

In addition participants were asked to rate their subjective comfort in striking each of the force measuring devices using a 10 point likert scale with 10 being totally comfortable and zero being totally uncomfortable.

4.7.2 (iii) Data processing

Lower extremity stance phase 3-D kinematic parameters from each of the four conditions were quantified using Visual 3-D (C-Motion Inc, Germantown, USA) and filtered at 10 Hz using a zero-lag low pass Butterworth 4th order filter. Angles were created using an XYZ, sequence of rotations.

4.7.2 (iv) Statistical Analysis

Descriptive statistics including means and standard deviations were calculated for each condition. Differences between the parameters were examined using repeated measures ANOVA's with significance accepted at the $p \leq 0.05$ level. Appropriate post-hoc analyses were conducted using a Bonferroni correction to control for type I error. Effect sizes were calculated using a η^2. If the sphericity assumption was violated then the degrees of freedom were adjusted using the Greenhouse Geisser correction. The Shapiro-Wilk statistic for each condition confirmed that all data were normally distributed. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA).

4.7.3 Results

Tables 4.19-4.21 and figure 4.17 present the mean ± standard deviation kinematic waveforms and 3-D kinematic parameters obtained as a function of the four running conditions.

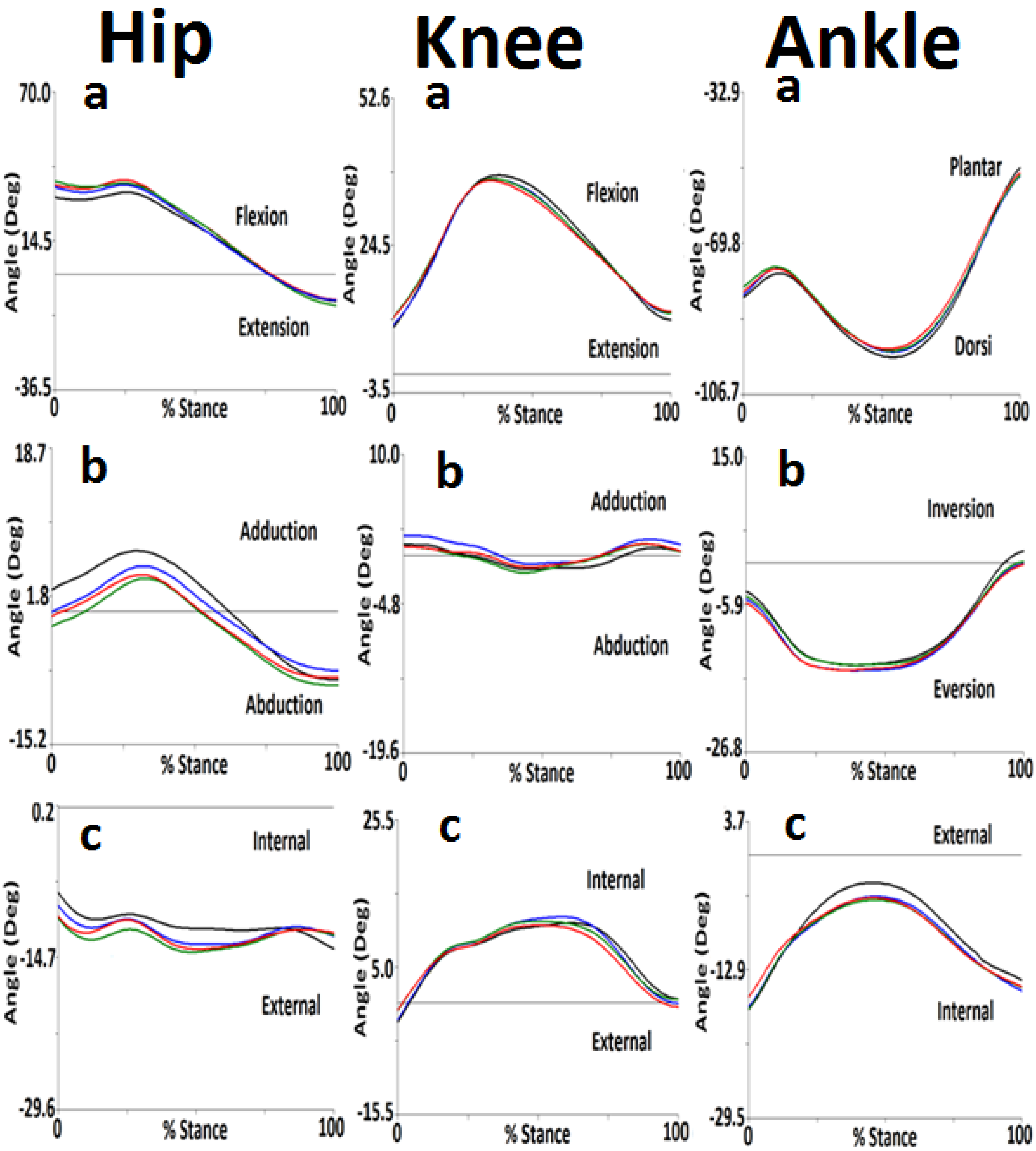

Figure 4.17: Hip, knee and ankle joint kinematics in the a. sagittal, b. coronal and c. transverse planes as a function of cardan sequence (Black = uninhibited, Blue = force platform, Green = RS scan and Red = Tek scan).

Table 4.19: Hip joint kinematics as a function of running condition (* = significant main effect).

	Uninhibited	Force platform	RS Scan	Tek Scan	
Hip					
X (+ = flexion/ - = extension)					
Angle at Footstrike (°)	36.50 ± 9.18	39.02 ± 8.31	41.45 ± 9.02	40.19 ± 9.12	*
Angle at Toe-off (°)	-7.41 ± 10.34	-6.82 ± 5.14	-8.73 ± 7.50	-7.07 ± 8.14	
Range of Motion (°)	43.92 ± 8.02	45.95 ± 9.17	49.01 ± 8.23	47.91 ± 7.44	
Relative Range of Motion (°)	2.61 ± 1.30	1.96 ± 1.14	1.06 ± 1.01	2.02 ± 1.21	
Peak Flexion (°)	39.08 ± 8.75	40.87 ± 8.39	41.45 ± 8.44	42.20 ± 9.79	
Y (+ =adduction/-=abduction)					
Angle at Footstrike (°)	2.51 ± 7.05	1.30 ± 5.45	-0.17 ± 5.53	-2.00 ± 6.09	*
Angle at Toe-off (°)	-7.76 ± 2.75	-8.18 ± 2.99	6.82 ± 3.15	-7.45 ± 3.31	
Range of Motion (°)	10.32 ± 6.14	9.42 ± 3.68	6.92 ± 3.04	5.55 ± 4.12	
Relative Range of Motion (°)	4.69 ± 3.15	4.35 ± 2.78	4.51 ± 2.30	6.63 ± 2.66	
Peak Adduction (°)	7.50 ± 4.83	5.40 ± 3.99	4.40 ± 3.97	4.62 ± 4.30	*
Z (+ =internal /- =external)					
Angle at Footstrike (°)	-8.56 ± 4.72	-9.75 ± 4.29	-11.02 ± 4.15	10.68 ± 3.29	
Angle at Toe-off (°)	-14.48 ± 5.62	-12.57 ± 3.96	-12.44 ± 4.43	-12.27 ± 4.89	
Range of Motion (°)	6.21 ± 4.44	2.95 ± 2.01	1.57 ± 3.22	1.85 ± 3.19	
Relative Range of Motion (°)	7.64 ± 5.74	6.05 ± 4.08	6.01 ± 4.31	5.57 ± 3.93	
Peak External rotation (°)	-16.24 ± 4.30	-15.80 ± 3.17	-17.07 ± 3.19	-16.24 ± 3.93	

A significant main effect was observed $F_{(3, 30)}$ =3.93, p≤0.05, η^2=0.32 for the extent of hip flexion at footstrike. Post-hoc analysis revealed that hip flexion in the RS scan mat condition was significantly (p=0.048) greater than when running in the uninhibited condition. In the coronal plane a significant main effect was found $F_{(1.59,\ 15.99)}$ =7.82, p≤0.01, η^2=0.44 at footstrike. Post-hoc analyses revealed that adduction at footstike was significantly (p=0.009) greater in the uninhibited condition in comparison to the Tek scan mat. Finally, a significant main effect was also observed $F_{(3,\ 30)}$ = 7.28, p≤0.05, η^2=0.41 for the magnitude of peak adduction. Post-hoc analysis revealed that peak adduction was significantly greater in the uninhibited condition in comparison to the RS scan.

Table 4.20: Knee joint kinematics as a function of running condition (* = significant main effect).

	Uninhibited	**Force platform**	**RS Scan**	**Tek Scan**	
Knee					
X (+ = flexion/ - = extension)					
Angle at Footstrike (°)	9.15 ± 5.85	9.77 ± 5.95	10.91 ± 6.55	10.82 ± 6.12	
Angle at Toe-off (°)	10.34 ± 5.92	11.44 ± 5.33	11.65 ± 4.53	11.94 ± 4.48	
Range of Motion (°)	1.84 ± 2.59	2.52 ± 2.19	1.81 ± 4.10	1.99 ± 1.91	
Relative Range of Motion (°)	29.33 ± 4.57	28.06 ± 4.37	26.68 ± 3.07	26.46 ± 4.07	
Peak Flexion (°)	38.48 ± 6.43	37.83 ± 7.09	37.59 ± 5.59	37.28 ± 6.15	
Y (+ =adduction/-=abduction)					
Angle at Footstrike (°)	0.96 ± 5.38	1.63 ± 3.89	0.99 ± 4.53	0.88 ± 4.87	
Angle at Toe-off (°)	0.33 ± 5.47	0.88 ± 4.50	0.42 ± 4.52	0.41 ± 4.87	
Range of Motion (°)	1.09 ± 1.16	1.22 ± 1.09	0.53 ± 1.19	1.33 ± 0.92	
Relative Range of Motion (°)	4.37 ± 3.05	4.62 ± 3.22	4.43 ± 2.74	4.02 ± 2.31	
Peak Abduction (°)	-3.40 ± 6.17	-2.98 ± 5.39	-3.43 ± 5.54	-3.14 ± 5.48	
Z (+ =internal /- =external)					
Angle at Footstrike (°)	-2.50 ± 3.63	-2.04 ± 4.51	-1.05 ± 5.64	-1.00 ± 3.82	
Angle at Toe-off (°)	0.40 ± 3.57	0.08 ± 4.31	0.44 ± 4.03	-0.55 ± 3.02	
Range of Motion (°)	2.89 ± 2.99	1.92 ± 1.78	2.45 ± 1.99	2.76 ± 1.62	
Relative Range of Motion (°)	14.48 ± 5.74	14.76 ± 5.78	13.42 ± 6.38	12.74 ± 6.22	
Peak Internal Rotation (°)	11.98 ± 5.62	12.73 ± 5.39	12.37 ± 5.66	11.73 ± 5.36	

No significant ($p>0.05$) main effects were observed for knee joint kinematics.

Table 4.21: Ankle joint kinematics as a function of running condition (* = significant main effect).

	Uninhibited	Force platform	RS Scan	Tek Scan	
Ankle					
X **(+ = plantar/ - = dorsi)**					
Angle at Footstrike (°)	-82.92 ± 5.69	-82.35 ± 6.82	-80.36 ± 6.90	-81.59 ± 7.08	*
Angle at Toe-off (°)	-51.16 ± 4.95	-52.89 ± 7.35	-53.33 ± 6.45	-52.49 ± 5.06	
Range of Motion (°)	30.76 ± 3.79	29.46 ± 8.93	27.03 ± 9.61	29.10 ± 7.28	
Relative Range of Motion (°)	14.99 ± 4.33	14.14 ± 3.51	16.09 ± 4.86	14.57 ± 4.41	
Peak Dorsi-flexion (°)	-97.90 ± 4.99	-96.48 ± 4.95	-96.45 ± 4.81	-96.05 ± 5.33	
Y **(+ =inversion/-=eversion)**					
Angle at Footstrike (°)	-4.07 ± 4.51	-5.15 ± 4.07	-4.88 ± 4.52	-5.65 ± 5.01	
Angle at Toe-off (°)	1.45 ± 6.13	-0.11 ± 6.88	-0.07 ± 6.86	-0.28 ± 6.61	
Range of Motion (°)	5.88 ± 3.07	5.72 ± 2.57	5.47 ± 3.14	5.51 ± 2.57	
Relative Range of Motion (°)	11.35 ± 3.69	10.91 ± 4.16	10.62 ± 4.68	10.22 ± 4.50	
Peak Eversion (°)	-15.42 ± 6.36	-15.66 ± 6.67	-15.51 ± 7.07	-15.87 ± 6.61	
Z **(+ =external /- =internal)**					
Angle at Footstrike (°)	-17.16 ± 5.18	-17.21 ± 5.34	-17.17 ± 4.90	-15.87 ± 4.88	
Angle at Toe-off (°)	-14.88 ± 4.98	-14.06 ± 4.44	-14.91 ± 4.42	-14.77 ± 4.60	
Range of Motion (°)	5.64 ± 3.09	4.35 ± 2.91	3.78 ± 2.56	4.19 ± 2.20	
Relative Range of Motion (°)	14.39 ± 4.37	12.76 ± 3.96	12.45 ± 3.41	11.58 ± 4.25	
Peak External rotation (°)	-2.93 ± 4.06	-3.99 ± 3.93	-4.71 ± 3.99	-4.29 ± 3.95	*

In the sagittal plane a significant main effect was observed $F_{(1.32,\ 12.20)} = 6.10$, $p \leq 0.01$, $\eta^2=0.40$ at footstrike. Post-hoc analyses revealed that when striking the RS scan mat the ankle was significantly (p=0.03) more plantarflexed than in the uninhibited condition. In the transverse plane significant main effects were observed for the magnitude of peak internal rotation $F_{(1.28,\ 12.80)} = 5.96$, $p \leq 0.01$, $\eta^2=0.37$. Post-hoc analyses revealed that peak rotation in the uninhibited condition was significantly greater than when striking the three force measuring devices. In addition a significant transverse plane main effect was also observed for ROM $F_{(3,\ 30)} = 4.82$, $p \leq 0.05$, $\eta^2=0.33$. Post-hoc analysis revealed that ROM was significantly (p=0.009) greater in the uninhibited condition in comparison to the RS scan mat.

Finally a significant main effect was also observed for relative ROM $F_{(3, 30)} = 7.95$, $p \leq 0.01$, $\eta^2=0.44$. Post-hoc analysis revealed that relative ROM in the uninhibited condition was significantly ($p=0.014$) greater in the uninhibited condition in comparison to the Tek scan.

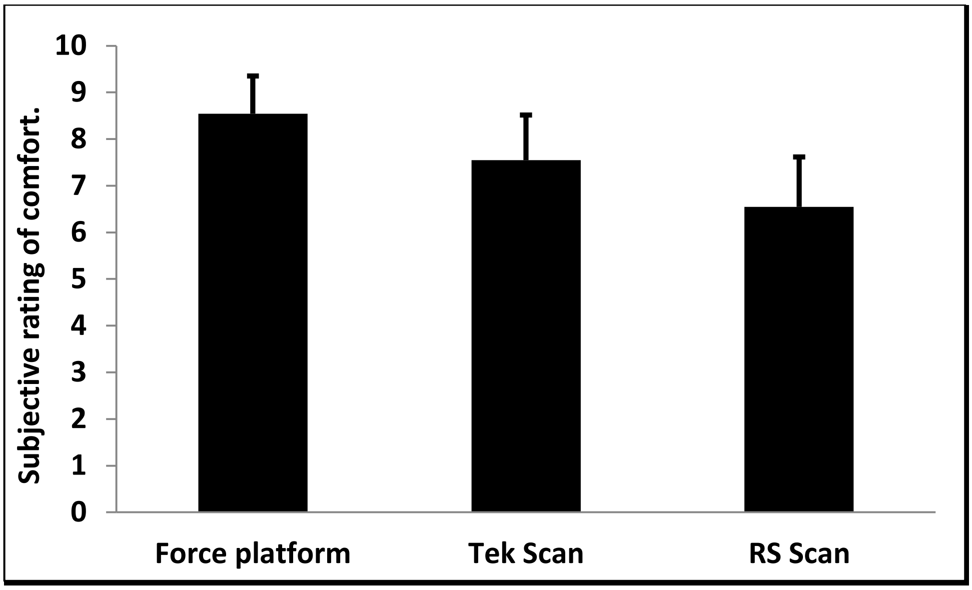

Figure 4.18: Subjective ratings of running comfort when striking the three different force measuring devices.

A significant main effect $F_{(1.27,\ 12.27)} = 20.63$, $p \leq 0.01$, $\eta^2=0.67$ was also observed for the magnitude of subjectively rated comfort. Post-hoc analyses revealed that participants rated the force platform running condition significantly more comfortable than either the RS scan ($p=0.0002$) or Tek scan ($p=0.001$). No significant differences were observed between the RS scan and Tek scan conditions.

4.7.4 Discussion

The aim of the current pilot investigation was to examine the influence of different force measuring transducers on targeting. This pilot investigation represents the first to document the effects of these different transducers on 3-D kinematics in comparison to running uninhibited.

That significant increases in hip flexion at footstrike were observed when running in the RS scan mat suggests that in this condition, participants increased their stride length in order to make contact with the RS scan mat. It is hypothesized that this finding is attributable to the fact that the RS scan is almost 1cm thick in comparison to the embedded force platform and the Tek scan which is less than 0.5cm. Striking an object of this nature is likely to have influenced the running mechanics to a due to the perceptual influence of a raised object.

The increases in ankle plantarflexion observed at footstrike in the RS scan running condition may also relate to the visuo-perceptual influence of the raised RS scan condition. In relation to the embedded force platform and uninhibited conditions, participants may alter their sagittal plane ankle position at footstrike in order to produce a secure strike with the surface of the device. This concurs with the observations of Challis, (2001) who found alterations in ankle position at footstrike when stride length was altered.

Furthermore, participants rated subjectively that the force platform facilitated a significantly more natural/comfortable running gait pattern in comparison to the RS scan and Tek scan conditions. This finding whilst subjective does appear to relate to the kinematic observations, in particular those at the hip and ankle.

In conclusion findings from this investigation have a number of key implications. Firstly from the aims of this pilot study it suggests that the force platform is the optimal force measuring device (from those examined in the current investigation) when obtaining simultaneous kinetics and 3-D kinematics. Secondly this pilot investigation does call in to question the efficacy of previous analyses which have used devices such as the RS scan/Tek scan mats overlying the laboratory surface to collect information during running.

4.8 Pilot study 8: The reliability of three hip joint centre identification techniques for the quantification of hip and knee kinematics during running

4.8.1 Introduction

Locating the centre of the hip is required to calculate hip joint rotations and moments in gait analysis (Cappozzo et al. 1975; Kirkwood et al. 1999; Stagni et al. 2000). A number of techniques currently exist which may include anatomical (Bell et al. (1989), functional (Cappozzo, 1984; Leardini et al. 1999) and projection (Weinhandl and O'Connor, 2010) methods, all of which may influence the resultant hip and knee joint profiles (Stagni et al. 2000). Although validity of each method has been reported to justify their utilization, there is currently a lack of consensus regarding the most appropriate technique for running analyses. Furthermore, whilst investigations have been conducted whereby the reliability in determining the anatomical position of the hip joint centre is examined, there is a paucity of information regarding the influence that the different techniques have upon the reliability of the resultant kinematic waveforms and discrete variables. Therefore the aim of the current investigation was to compare the reliability of the three (Bell, projection and functional) different hip joint centre estimation techniques using both discrete variable and waveform analyses.

4.8.2 Methods

4.8.2 (i) Participants

Fifteen male participants volunteered to take part in this investigation (age 25.1 ± 1.95 years; height 1.77 ± 0.07m; body mass 76.1 ± 6.9 kg). All were injury free at the time of data collection and provided written informed consent. Ethical approval for this project was obtained from the University of Central Lancashire School of Psychology ethics committee.

4.8.2 (ii) Procedure

Participants completed five trials running at $4.0 m.s^{-1}$±5% over a force platform. The stance phase was defined as the time over which 20 N or greater of vertical force was applied to the force platform.

To examine the reliability of hip joint centre estimation via the projection and Bell techniques two static calibration trials were captured with the participant standing in the anatomical position. The first static (test) was conducted prior to the running trials, to define the anatomical reference frames of the thigh, tibia and pelvic segments, retro-reflective markers were positioned on the medial and lateral epicondyles of the femur, greater trochanter of the right leg, left and right iliac crests, medial and lateral malleoli and left and right ASIS and PSIS; following which the anatomical markers (with the exception the femoral epicondyles and medial and lateral malleoli) were removed. These markers were left on throughout and attached with strong tape to ensure that the distal end of the thigh and the proximal/distal ends of the tibia remained consistent. Allowing the reliability of the hip joint centre to be examined. Tracking clusters were also positioned, on the pelvis, thigh and tibial segments. Following completion of the running trials the anatomical landmarks were re-positioned and the second static trial (retest) was obtained. Cluster markers used to define the technical tracking frame of the thigh, tibial and pelvic segments remained rigidly in place for the duration of the analysis and were not removed, allowing the test-retest reliability of the hip joint centre location to be examined.

The Bell technique is based on the positions of the ASIS markers using the following regression equation: Hip joint centre= (0.36* Distance between ASIS markers medial to the ASIS marker, distance, 0.19* Distance between ASIS markers posterior to the ASIS marker, 0.3* Distance between ASIS markers inferior to the ASIS marker). The projection technique places the hip joint centre at one-quarter of the distance from the ipsilateral to the contralateral greater trochanter (Weinhandl and O'Connor, 2010).

To define the functional hip joint centre participants performed five sequential flexion-extension and abduction-adduction movements of the right hip at a self-selected velocity followed by a cycle of full hip circumduction. Flexion-extension and abduction-adduction ranges of movement were in the order of 45 and 40°, respectively (Besier et al. 2003). To examine the repeatability of this technique, this procedure was repeated before and after the collection of the running data (before the collection of the second static trial).

4.8.2 (iii) Data processing

Lower extremity (Hip and knee joint) 3-D kinematic parameters from test and retest hip joint centre locations were quantified using Visual 3-D (C-Motion Inc, Germantown, USA) and filtered at 10 Hz using a zero-lag low pass Butterworth 4^{th} order filter. Angles were created using an XYZ, sequence of rotations.

4.8.2 (iv) Statistical analysis

Descriptive statistics including means and standard deviations were calculated between test and retest each of the three hip joint centre techniques. Differences in stance phase kinematic parameters were examined using paired samples t-tests with significance accepted at the $p \leq 0.05$ level. The Shapiro-wilk statistic for each condition confirmed that the data were normally distributed. Intra-class correlations were also utilized to compare test and retest sagittal, coronal and transverse plane waveforms of the hip and knee for each hip joint centre location. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA).

4.8.3 Results

Tables 4.22 and 4.23 and figures 4.19-4.21 present the mean ± standard deviation angles and waveforms obtained as a function of test-retest for each hip joint centre location technique.

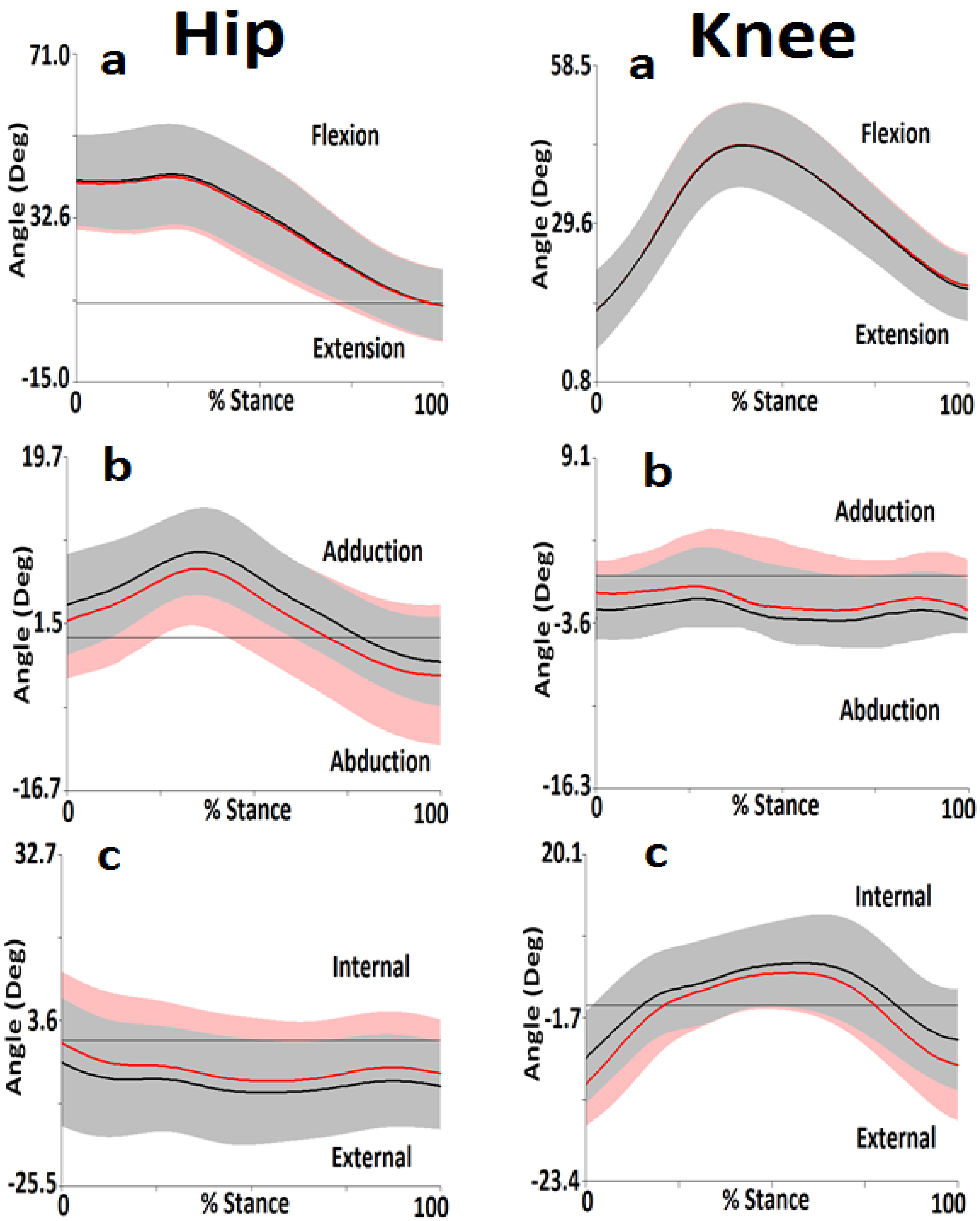

Figure 4.19: 3-D kinematics of the hip and knee in the a. sagittal, b. coronal and c. transverse plane using the Bell technique (Black = test and Red = Retest).

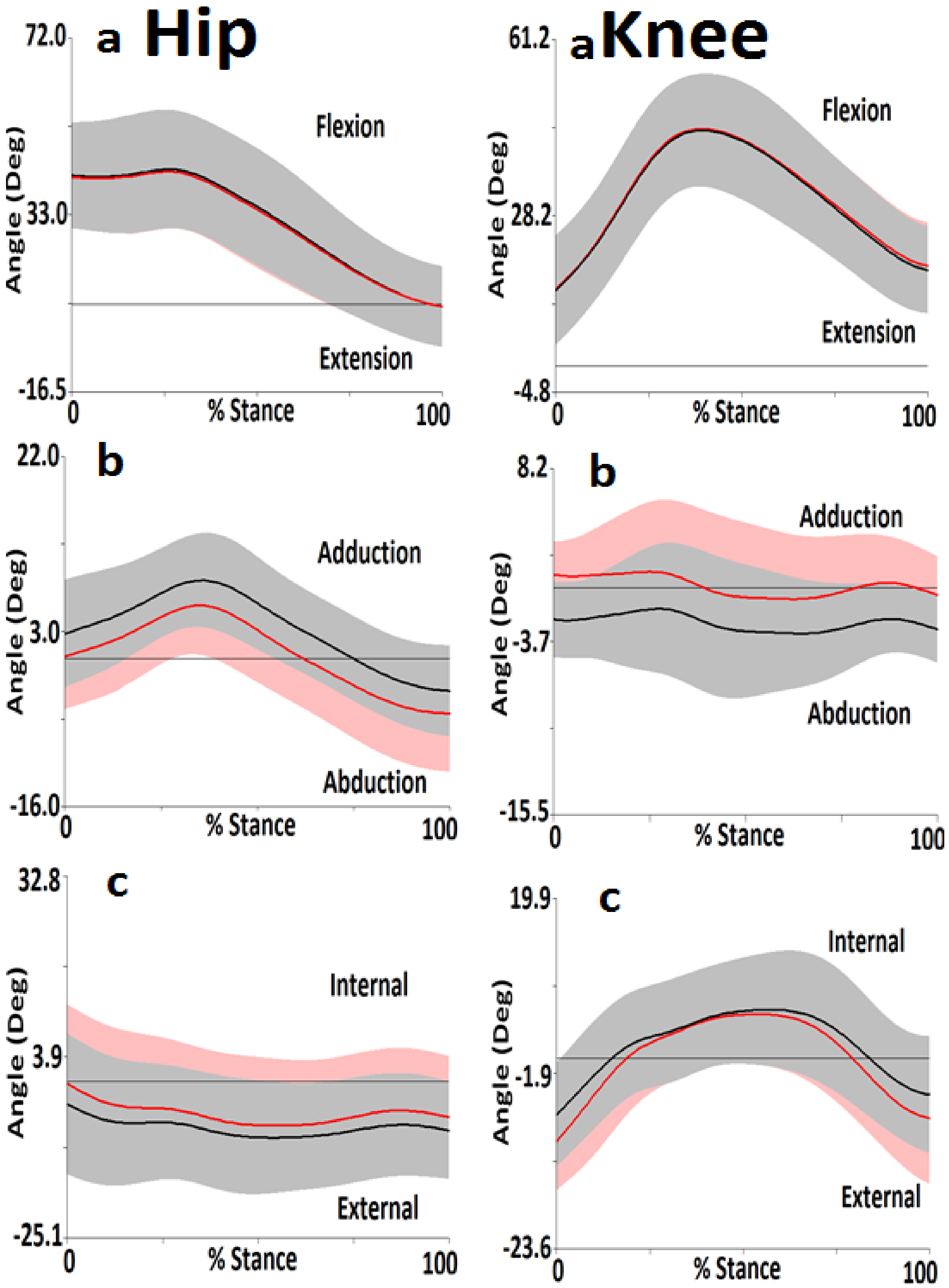

Figure 4.20 3-D kinematics of the hip and knee in the a. sagittal, b. coronal and c. transverse plane using the projection technique (Black = test and Red = Retest).

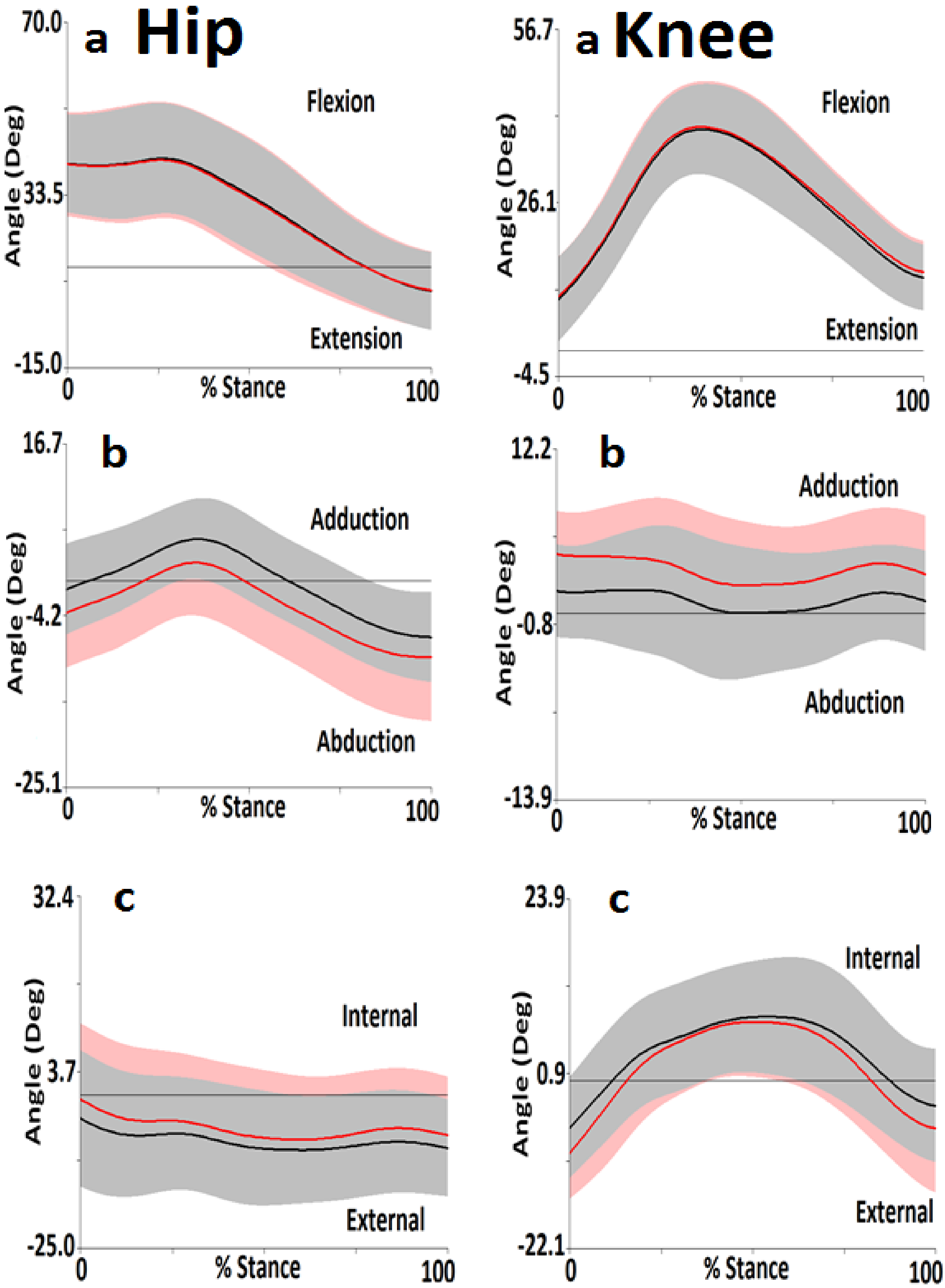

Figure 4.21 3-D kinematics of the hip and knee in the a. sagittal, b. coronal and c. transverse plane using the functional technique (Black = test and Red = Retest).

Table 4.22: Hip joint kinematics from test – retest as a function of hip joint centre location (* = significantly different from test).

Hip	Bell		Functional		Projection	
X (+ =flexion/ - =extension)	**Test**	**Retest**	**Test**	**Retest**	**Test**	**Retest**
Angle at Footstrike (°)	40.91 ± 12.39	40.35 ± 11.85	35.59± 13.43	35.53 ± 12.21	40.94 ± 13.56	41.71 ± 12.30
Angle at Toe-off (°)	-1.30 ± 9.19	-2.10 ± 9.22	-6.97 ± 10.42	-7.37 ± 10.51	-1.23 ± 9.63	-0.79 ± 8.30
Range of Motion (°)	42.43 ± 8.96	43.65 ± 9.05	42.60 ± 8.79	42.90 ± 8.85	42.11 ± 8.69	42.59 ± 8.61
Relative Range of Motion (°)	2.75 ± 2.01	2.60 ± 2.06	2.51 ± 1.96	2.28 ± 2.00	2.74 ± 2.12	2.55 ± 2.15
Peak Flexion (°)	43.63 ± 13.53	42.92 ± 12.99	38.09 ± 14.77	37.81 ± 14.53	43.70 ± 14.79	44.27 ± 13.71
Y (+ =adduction/ - =abduction)						
Angle at Footstrike (°)	3.74 ± 5.45	2.64 ± 5.83	-0.43 ± 5.44	-2.33 ± 6.69	3.43 ± 5.99	1.57 ± 5.81
Angle at Toe-off (°)	-1.80 ± 5.81	-3.33 ± 7.90	-5.48 ± 6.67	-7.75 ± 8.66	-2.21 ± 6.27	-4.73 ± 7.16 *
Range of Motion (°)	6.30 ± 3.34	6.81 ± 3.86	6.35 ± 3.39	6.82 ± 3.29	6.18 ± 3.96	6.76 ± 3.76
Relative Range of Motion (°)	6.29 ± 2.54	6.11 ± 2.71	6.65 ± 2.73	6.52 ± 2.83	6.26 ± 2.35	5.96 ± 2.30
Peak Adduction (°)	10.05 ± 5.51	8.79 ± 6.39	6.21 ± 5.61	4.19 ± 7.10 *	9.68 ± 6.06	7.56 ± 6.20 *
Z (+ =internal/ - =external)						
Angle at Footstrike (°)	1.33 ± 14.74	2.06 ± 14.29	-1.41 ± 14.60	1.86 ± 14.14	-1.19 ± 14.73	2.05 ± 14.36
Angle at Toe-off (°)	-5.21 ± 9.05	-3.91 ± 9.89	-7.34 ± 9.36	-4.66 ± 10.03	-6.69 ± 9.10	-3.93 ± 10.12
Range of Motion (°)	6.69 ± 5.77	6.77 ± 5.93	6.89 ± 5.67	7.02 ± 5.87	6.75 ± 5.85	6.84 ± 5.96
Relative Range of Motion (°)	6.59 ± 5.64	8.89 ± 5.58	8.67 ± 5.49	9.03 ± 5.48	3.25 ± 5.54	9.39 ± 8.97
Peak external Rotation (°)	-9.94 ± 10.16	-8.84 ± 10.61	-10.08 ± 10.23	-7.17 ± 10.61	-2.64 ± 6.34	-5.11 ± 7.18 *

The comparative results for the hip joint indicate that for the functional technique significant differences $t_{(14)} = 2.29$, $p \le 0.05$ were observed between test and retest for the magnitude of peak coronal plane abduction. In addition, for the projection technique significant differences were also observed in the coronal plane for the magnitudes of rotation at toe-off $t (14) = 2.11$, $p \le 0.05$, peak adduction $t_{(14)} = 2.72$, $p \le 0.05$ and relative ROM $t_{(14)} = 2.62$, $p \le 0.05$. No significant differences were observed for the Bell et al. technique.

Comparisons between test and retest hip joint kinematic waveforms for the Bell technique revealed strong correlations for the sagittal ($R^2 = 0.984$), coronal ($R^2 = 0.994$) and transverse ($R^2 = 0.90$) planes. For the functional method strong correlations were observed between test and retest waveforms in the sagittal ($R^2 = 0.982$), coronal ($R^2 = 0.968$) and transverse ($R^2 = 0.894$) planes. Finally, for the projection technique strong correlations were also observed for the sagittal ($R^2 = 0.974$), coronal ($R^2 = 0.99$) and transverse ($R^2 = 0.894$) planes.

Table 4.23: Knee joint kinematics from test – retest as a function of hip joint centre location (* = significantly different from test).

Knee	Bell		Functional		Projection	
X (+ =flexion/ - =extension)	**Test**	**Retest**	**Test**	**Retest**	**Test**	**Retest**
Angle at Footstrike (°)	14.55 ± 7.04	14.32 ± 6.14	9.97 ± 7.38	9.91 ± 6.62	14.90 ± 10.07	15.94 ± 8.84
Angle at Toe-off (°)	18.35 ± 5.06	17.94 ± 5.22	13.55 ± 4.91	13.51 ± 5.08	18.70 ± 7.86	19.55 ± 7.65
Range of Motion (°)	6.43 ± 3.22	6.39 ± 3.03	6.39 ± 3.21	6.40 ± 3.04	6.43 ± 3.21	6.39 ± 3.03
Relative Range of Motion (°)	30.42 ± 3.45	30.21 ± 3.42	30.42 ± 3.43	30.16 ± 2.23	30.39 ± 3.44	30.16 ± 3.59
Peak Flexion (°)	44.97 ± 7.59	44.53 ± 6.98	40.13 ± 7.96	40.07 ± 7.46	45.30 ± 10.55	46.10 ± 9.74
Y (+ =adduction/ - =abduction)						
Angle at Footstrike (°)	-2.78 ± 3.74	-1.11 ± 3.96	1.47 ± 3.52	3.59 ± 3.52 *	-2.05 ± 2.75	0.74 ± 2.48 *
Angle at Toe-off (°)	-3.96 ± 3.81	-2.42 ± 4.37	0.18 ± 4.09	2.16 ± 4.63 *	-3.26 ± 2.54	-0.63 ± 2.58
Range of Motion (°)	2.63 ± 2.09	2.66 ± 1.99	2.70 ± 2.13	2.69 ± 2.05	2.63 ± 2.06	2.66 ± 2.01
Relative Range of Motion (°)	3.85 ± 2.98	3.49 ± 2.83	4.24 ± 3.26	4.04 ± 3.35	3.80 ± 3.00	3.63 ± 2.97
Peak Abduction (°)	-6.63 ± 4.28	-5.01 ± 4.88	2.44 ± 4.52	0.41 ± 4.65 *	-5.86 ± 3.67	-1.89 ± 3.71 *
Z (+ =internal/ - =external)						
Angle at Footstrike (°)	-8.21 ± 6.24	-10.64 ± 5.42	-7.02 ± 7.08	-9.79 ± 5.76	-7.89 ± 6.94	-10.56 ± 5.84
Angle at Toe-off (°)	-6.69 ± 7.41	-8.39 ± 7.37	-4.33 ± 8.30	-7.05 ± 8.27	-5.56 ± 8.08	-8.07 ± 8.03
Range of Motion (°)	4.27 ± 1.89	4.53 ± 1.96	4.74 ± 2.33	4.99 ± 2.48	4.37 ± 1.98	4.72 ± 2.12
Relative Range of Motion (°)	13.79 ± 5.20	14.48 ± 5.13	15.33 ± 5.76	16.36 ± 5.69	14.07 ± 5.48	15.29 ± 5.27
Peak Internal Rotation (°)	6.05 ± 6.33	4.83 ± 5.43	8.33 ± 7.99	6.56 ± 7.55	6.18 ± 7.39	4.70 ± 6.43

The comparative results for the knee joint indicate that for the functional technique significant differences were observed between test and retest for the magnitude of rotation at footstrike $t_{(14)} = 2.91$, $p \leq 0.01$, toe-off $t_{(14)} = 2.90$, $p \leq 0.05$ and peak adduction $t\ (14) = 2.49$, $p \leq 0.05$. In addition, for the projection technique significant differences were also observed in the coronal plane for the magnitudes of rotation at toe-off $t_{(14)} = 4.10$, $p \leq 0.01$, peak adduction $t_{(14)} = 3.31$, $p \leq 0.01$ and relative ROM $t_{(14)} = 2.62$, $p \leq 0.05$. No significant differences were observed for the Bell et al. technique.

Comparisons between test and retest knee joint kinematic waveforms for the Bell technique revealed strong correlations for the sagittal ($R^2 = 0.999$), coronal ($R^2 = 0.942$) and transverse ($R^2 = 0.971$) planes. For the functional method strong correlations were observed between test and retest waveforms in the sagittal ($R^2 = 0.997$), coronal ($R^2 = 0.892$) and transverse ($R^2 = 0.96$) planes. Finally, for the projection technique strong correlations were also observed for the sagittal ($R^2 = 0.99$), coronal ($R^2 = 0.838$) and transverse ($R^2 = 0.861$) planes.

4.8.4 Discussion

The aim of the current pilot investigation was to determine the reliability of the three principal methods of defining the hip joint centre. In the present study, running trials were analysed simultaneously using three different anatomical co-ordinate systems. This represents the first study investigating the test-retest reliability in defining the hip joint centre and their potential influence on 3-D kinematic parameters during the stance phase of running.

In the sagittal plane no differences were observed between test and retest angular parameters for any of the three hip joint centre locations techniques. In addition to this the highest intra-class correlations were observed for the sagittal plane waveforms indicating a high level of reliability across all techniques in the sagittal plane. However in the coronal and transverse planes significant differences were observed between test and retest angular parameters using both the functional and projection techniques. Significantly, however the Bell et al. (1989) technique was associated with no significant test-retest differences in any of the three planes of rotation.

The waveform analysis corresponds with the discrete variable examination in that the Bell et al. (1989) technique was associated with the highest intra-class correlations between test-retest waveforms in comparison to the functional and projection techniques. This finding counters the hypotheses of Besier et al. (2003) who postulate that techniques for joint centre location using anatomical markers have greater propensity for error from marker re-application. There are several potential mechanisms for this observation; the examiner performing the analyses had extensive experience in locating anatomical positions around the pelvis, thus facilitating the increased reliability in defining the hip joint centre using the Bell et al. (1989) technique. Furthermore, all participants in this examination were trained distance runners with minimal body fat, which facilitate examiners ability to locate anatomical landmarks. Finally, it is hypothesized that this finding relates to the difficulty in re-producing the same range of movements required to locate the hip joint centre using the functional method thus potentially compromising its reliability in comparison to the Bell et al. (1989) technique. That the projection technique was associated with the lowest levels of reliability is perhaps unsurprising despite the observations of Weinhandl and O'Connor (2010). The head of the greater trochanter is broad in comparison to other anatomical locations and difficulty in defining this landmarks as a mere point, was documented by Della Croce et al. (1999) who reported test-retest errors of up to 12.2mm in the anteroposterior direction in locating the greater trochanter marker. This potential mislocation between test-retest static trials would influence the results kinematic waveforms, particularly in the coronal and transverse planes.

Therefore the results of this pilot investigation suggest that the Bell et al. (1989) technique for the estimation of the hip joint centre appears to be the most appropriate technique and will as such be utilized throughout this thesis.

4.9 Pilot study 9 Shock attenuating properties of lab and treadmill surfaces

4.9.1 Introduction

In order to obtain an understanding of the stiffness characteristics of the tested surfaces a pilot investigation was undertaken using a Berlin artificial athlete.

4.9.2 Methods

The Berlin Athlete (Labosport UK Unit 3 Heanor Gate Road, Derbyshire) apparatus involved dropping a 20 kg mass and measuring the peak force developed (Figure 4.22). The peak force developed, is compared to the force developed on concrete, and an expression for shock attenuation is provided as the % greater than or less than concrete. In addition to shock attenuation, measures of surface vertical deformation were also simultaneously obtained as a function of the total displacement of the falling mass after impact minus the deformation of the spring.

4.9.3 Results

Table 4.24 presents the attenuation and deformation properties of both the treadmill belt and laboratory floor (See also appendix D).

Table 4.24: Deformation and shock attenuation measures from each surface

	Treadmill	**Lab Floor**
Vertical Deformation (mm)	2.99 ± 0.19	0.27 ± 0.05
Shock attenuation (% Concrete)	52.33 ± 2.31	19.00 ± 3.46

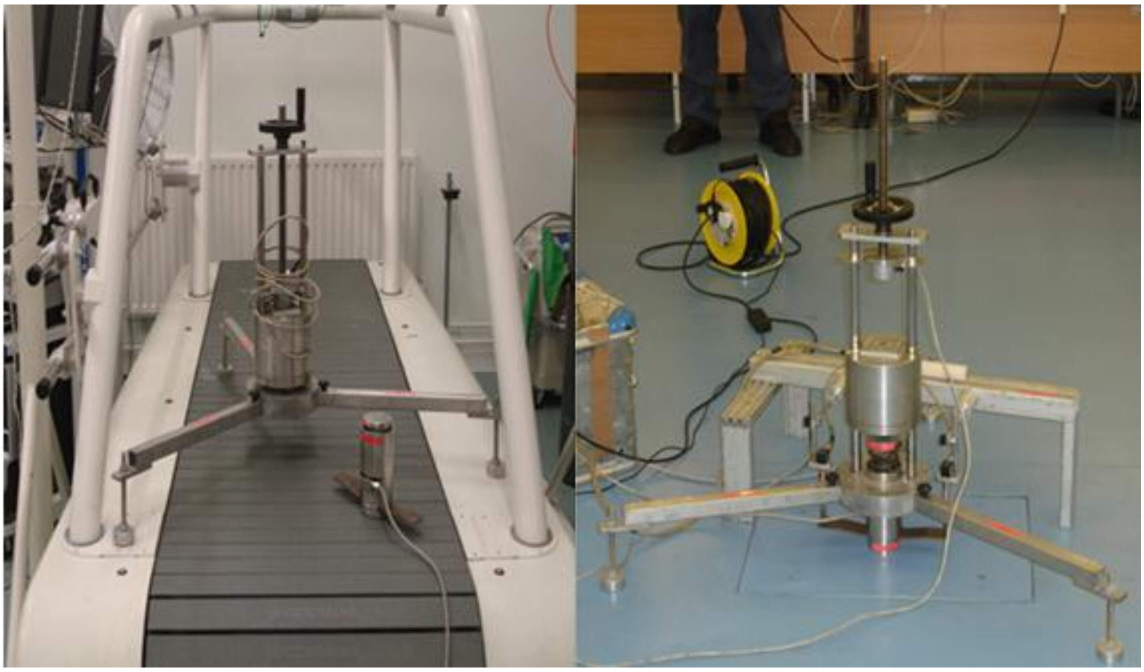

Figure 4.22: Berlin artificial athlete positioned on both treadmill and biomechanics laboratory floor.

4.9.4 Discussion

The results indicate that the Woodway treadmill belt offered greater surface compliance and mechanical shock attenuating properties in comparison to the laboratory floor.

4.10 Pilot Study 10 – Mechanical shock attenuating properties of footwear

4.10.1 Introduction

In order to determine the mechanical shock attenuating properties of the footwear examined during this thesis; a pilot study was conducted again using the Berlin artificial athlete.

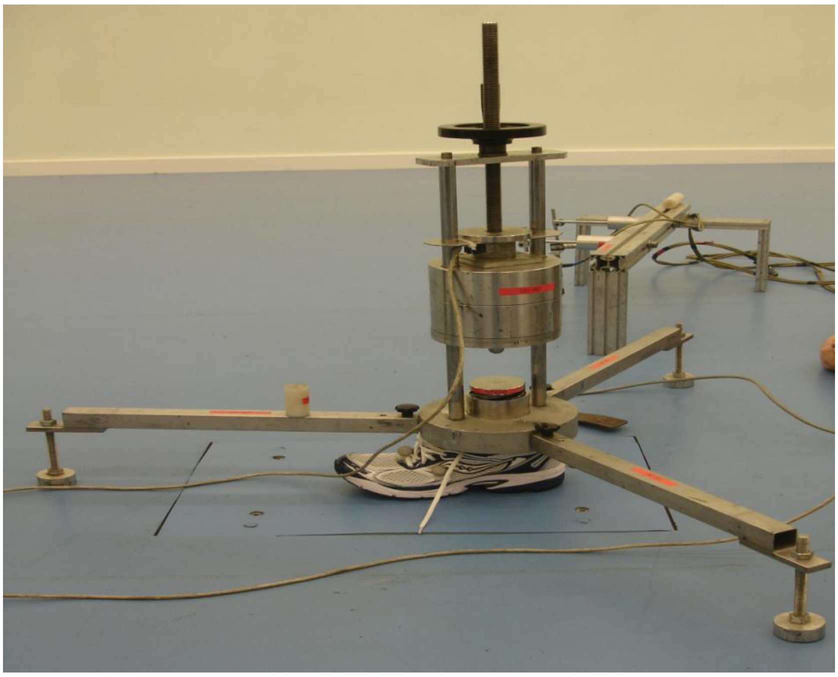

Figure 4.23: Mechanical testing of footwear shock attenuation properties using the Berlin Artificial Athlete.

4.10.2 Method

The analysis was conducted in an identical manner to that described above, whereby a 20kg mass was dropped a distance of 55mm onto the test foot which was placed inside the shoe. The shock attenuating properties of the experimental footwear were examined using a Berlin artificial athlete (Labosport UK Unit 3 Heanor Gate Road, Derbyshire) in agreement with ASTM F 1614 (ASTM 2006). In accordance with Aguinaldo and Mahar (2003), the test foot was dropped onto to the extreme rearfoot area of the footwear positioned on top of a piezoelectric force platform (Kistler Ltd; Model 9281CA, Kistler Instruments Ltd., Alton, Hampshire) sampling at 1000 Hz (Figure 4.23). In order to examine the mechanical shock attenuating properties of running barefoot, the test foot was placed directly on top of the force platform rather than into the rearfoot section of one of the shoes in accordance with Lake (2002); this condition was labelled as 'no footwear' for this pilot investigation. Vertical GRF parameters: impact peak, time to impact peak, average loading rate and instantaneous loading rate were averaged from five impacts in the same manner as described in chapters 6 and 7.

4.10.2 (i) Tested footwear

The shoes examined in this pilot investigation, and throughout the remained of the thesis include a Saucony pro grid guide II (Figure 4.24) and a Nike Free 3.0 (Figure 4.25). The Saucony pro grid guide II was selected as being the bestselling conventional shoe on the website pro.directrunning.com at the time that data collection for this thesis commenced. Conventional footwear is described as modern footwear that combines midsole cushioning with a structured element designed to reduce rearfoot eversion.

Figure 4.24: Saucony pro grid guide II.

The Nike Free 3.0 was selected firstly because the Free 3.0 range were the bestselling barefoot inspired footwear model on the website pro.directrunning.com at the commencement of data collection, and secondly because the Nike Free 3.0 represented the most minimal footwear in this range (The numbering system indicates the cushioning of the shoe and follows a scale ranging from 0 (barefoot) to 10 (normal running shoe), i.e. Free 3.0 being the least and Free 7.0 being the most cushioned). The Nike Free 3.0 aims to simulate barefoot running through a flexible outsole construction and feature a relatively wide and soft heel (Nigg, 2009).

Figure 4.25: Nike Free 3.0 footwear.

4.10.3 Results

Table 4.25 presents the kinetic parameters observed as a function of footwear.

Table 4.25: Mean and standard deviations of impact drop tests for each condition

	Impact Peak (k.N)	Time to Peak (ms)	Average Loading Rate (k.N.s^{-1})	Instantaneous Loading rate (K.N.s^{-1})
Saucony	1.73 ± 0.06	16.0 ± 0.006	109.34 ± 3.80	219.31 ± 5.10
Nike Free	1.71 ± 0.02	9.4 ± 0.7	149.253 ± 7.34	308.43 ± 7.91
No footwear	3.90 ± 0.04	7.2 ± 0.70	557.34 ± 72.12.	1105.10 ± 81.90

4.10.4 Discussion

The results obtained from the current pilot investigation suggest that in relation to both of the experimental footwear the no footwear condition was associated with noticeably higher impact parameters. In addition the results from the drop tests further indicate that the Saucony shoes offer more advanced mechanical shock attenuating properties, in comparison to the Nike Free shoes.

5. Differences in kinetics and 3-D kinematics between treadmill and overground running: Implications for footwear design

Publications

1. Sinclair J, Hobbs SJ, Edmundson CJ and Brooks D (2011). Evaluation of kinematic methods of identifying foot strike and toe-off during running, International Journal of Sports Science and Engineering, 5, 3, 188-192.
2. Sinclair, J, Richards J, Taylor PJ, Edmundson, CJ and Hobbs SJ (2012). 3-D kinematics of treadmill and overground running, Sports Biomechanics, 12, 1-12, I First.
3. Sinclair J, Edmundson CJ, Brooks D and Hobbs SJ (2011). Tibial Transients measured during treadmill and overground running. Journal of Sports Sciences, S12
4. Sinclair, J, Hobbs SJ, Protheroe L and Greenhalgh A (2012). Determination of gait events using an externally mounted shank accelerometer, Journal of Applied Biomechanics. (In press).

Conference presentations

5. Sinclair J, Edmundson CJ, Brooks D and Hobbs SJ (2011). Tibial Transients measured during treadmill and overground running. Presented at the BASES annual conference, University of Essex, 6th-8th September

6. Sinclair J, Edmundson, CJ, Brooks and Hobbs SJ (2010). Skeletal transients during overground and treadmill locomotion, Science and Technology, Annual research conference. University of Central Lancashire.

7. Sinclair J, Edmundson, CJ, Brooks and Hobbs SJ (2011). Biomechanical comparison of overground and treadmill running. European Conference of Sports Sciences, ECSS, Liverpool, John Moores University

8. Sinclair J, Hobbs, SJ, Protheroe L and Greenhalgh A (2010). Utilization of an externally mounted accelerometer to determine gait events. Presented at the Bases annual conference, University of Aberystwyth, 31st March-1st April.

5.1 Pilot study 1 – determination of gait events using kinematic data.

5.1.1 Introduction

Given that the treadmill used throughout this chapter did not feature an integrated force platform an alternative method of determining footstrike and toe-off events had to be established.

Several kinematic methods are available for gait event determination, yet comparisons of their accuracy in defining stance phase events have, yet to be reported. Mickelborough et al. (2001) developed a method of determining gait events during walking. Heel-strike was associated with the second of the W shaped minima of the foot's vertical velocity curve in the Z (vertical) axis whilst toe-off was determined as the minimum position of the toe-markers in the Z axis. O'Connor et al. (2007) developed the foot velocity algorithm, whereby heel-strike was associated with the first trough of the foot segment velocity in the Z (vertical) axis and toe-off was associated with the peak foot segment velocity in the Z (vertical axis). Alton et al. (1998) used the minimum position of the lateral malleolus in the Z axis in order to determine footstrike. Toe-off was defined using the same method as Mickelborough et al. (2001) via the position of the metatarsal markers in the Z axis. Similarly, Zeni et al. (2008) proposed two methods of identifying gait events. The first used the difference in displacement of the peaks and troughs of sacral and foot markers in the sagittal plane. The second method is a velocity based technique. The velocity of the heel marker in the sagittal plane changes from positive to a negative direction at each heel strike. The frame at which backward movement of the foot is initiated is termed heel-strike. At the initiation of swing phase the velocity of the heel or toe markers alters from negative to positive and is thus labelled toe-off.

Hreljac and Stergiou, (2000) utilized shank and foot motion in the sagittal plane. They determined foot strike as the time that coincided with the minimum sagittal plane foot angular velocity, and toe-off as the local minimum of the shank angular velocity. Schache et al. (2001) utilized the vertical velocity and displacement of the foot markers to identify gait events for overground and treadmill running. Heel strike was deemed to be the time of the downward spike of the vertical velocity of the 1st metatarsal and the plateau in the displacement of the lateral malleoli marker in the Z axis. Toe-off was deemed to be the onset of the rise in vertical displacement and velocity of the 1st metatarsal marker. Finally, Dingwell et al. (2001) provided a kinematic method designed specifically for treadmill locomotion. Foot strike was deemed to be the first time when peak knee extension occurred and toe-off was determined as the second occurrence of peak knee extension.

The overall objective of this pilot investigation was to illustrate the most accurate means of predicting heel strike and toe-off during running, by contrasting the computationally predicted events to those detected using force data.

5.1.2 Methods

5.1.2 (i) Participants

Eleven male participants volunteered to take part in this investigation (age 19.3 ± 1.56 years; Height 1.77 ± 0.52 m; Mass 78.4 ± 9.0 kg). The study was approved by the School of Psychology ethical committee, and all participants provided written informed consent.

5.1.2 (ii) Procedure

Participants completed five trials, running at 4.0 $m.s^{-1}$ along a 20 m runway striking the centre of a force platform (Kistler, Kistler Instruments Ltd; Model 9281CA), sampling at 1000 Hz. Timing gates Newtest 300 (Newtest, Oy Koulukatu) were used to monitor running velocity, a maximum deviation of ±5% from the specified velocity was allowed.

Plots of vertical force were produced from which heel strike and toe-off events were identified, specifically heel strike was quantified as the first instance at which the vertical component of the GRF was greater than 20N; toe-off was determined to be the first instance in which the vertical GRF fell below 20N.

5.1.2 (iii) Data processing

Average and absolute errors were quantified as the average (net discrepancy indicative of the magnitude of the difference which is influenced by the direction of the time difference between methods) and absolute (the absolute discrepancy indicative of the magnitude of the difference irrespective of direction) discrepancy between the two methods for identifying the event times. A positive value represented an event defined after the event established from the force data and a negative value represented an event defined prior to the force plate event. The difference in the time of occurrence in milliseconds was then quantified.

5.1.2 (iv) Statistical analysis

The statistical differences of both average (the net discrepancy indicative of the magnitude of the difference which is influenced by the direction of the time difference between methods) and absolute (the absolute discrepancy indicative of the magnitude of the difference irrespective of direction) errors between the kinematic methods were examined using repeated measures ANOVA's with significance accepted at the ($p \leq 0.05$) level. Post hoc analyses were conducted using a Bonferroni correction to control for type I error.

5.1.3 Results

Tables 5.1 and 5.2 present the accuracy in defining footstrike and toe-off events of the eight kinematic techniques in comparison to the force platform.

For heel-strike a significant main effect was found for both absolute $F_{(7, 63)} = 33.72$, $p \leq 0.01$, $\eta^2 = 0.79$ and average error $F_{(7, 63)} = 42.20$, $p \leq 0.01$, $\eta^2 = 0.82$. Post-hoc analysis revealed that the Alton et al. (1998), O'Connor et al. (2007) and Dingwell et al. (2001) algorithms were associated with significantly $p \leq 0.05$ lower average and absolute errors. For toe-off a significant main effect for both absolute $F_{(7, 63)} = 4.51$, $p \leq 0.05$, $\eta^2 = 0.33$ and average error $F_{(7, 63)} = 4.35$, $p \leq 0.05$, $\eta^2 = 0.33$ was found. Post-hoc analysis indicated that the Dingwell et al. (2001) and Schache et al. (2001) algorithms were associated with significantly $p \leq 0.05$ lower average and absolute errors.

Table 5.1: Average and absolute Error (ms) of heel-strike determination methods (means, standard deviation, minimum maximum and 95% confidence intervals).

		Schache	Alton	Dingwell	Mickelborough	Hreljac	Zeni A	Zeni B	O'Connor
Average error (ms)	mean	-40.17	15.84	-28.43	295.02	-57.34	-47.87	2.82	12.82
	std.dev	47.38	11.36	17.35	126.63	42.71	21.90	72.82	15.55
	max	21.66	40.00	17.30	472.30	16.70	-11.00	176.90	50.00
	min	-104.29	3.00	-45.00	66.40	-113.20	-92.00	-58.30	-1.40
	95% C.I	-74.0/-6.27	7.72/23.96	-40.8/-16.02	204.43/385.61	-87.89/-26.79	-63.54/-32.20	-49.27/-54.91	1.63/23.95
Absolute error (ms)	mean	45.83	15.84	31.89	295.02	62.82	47.87	53.40	14.22
	std.dev	41.28	11.36	8.31	126.63	33.05	21.90	46.29	18.83
	max	104.29	40.00	45.00	472.30	113.20	92.00	176.90	62.50
	min	3.33	3.00	17.30	66.40	10.70	11.00	10.20	1.40
	95% C.I	16.30/75.36	7.72/23.96	25.95/37.83	204.43/385.61	39.18/86.46	32.30/63.54	20.29/86.51	0.75/27.69

Table 5.2: Average and absolute Error (ms) of toe-off determination methods (means, standard deviation, minimum maximum and 95% confidence intervals).

		Schache	Alton	Dingwell	Mickelborough	Hreljac	Zeni A	Zeni B	O'Connor
Average error (ms)	mean	-45.77	-80.59	10.99	-80.59	-97.32	2.23	45.46	-123.47
	std.dev	25.48	71.86	14.19	71.86	82.62	118.55	146.46	124.66
	max	20	87.1	43.5	87.1	10.0	112.4	153.3	46.7
	min	-80.70	-170.2	-8.3	-170.2	-260	-265.0	-332.0	-276.3
	95% C.I	-63.99/-27.54	-131.99/-29.19	0.84/21.14	-131.99/-29.19	-156.42/-38.22	-82.57/87.03	-59.31/150.23	-212.65/-34.29
Absolute error (ms)	mean	49.77	98.01	13.27	98.01	100.74	89.93	123.98	135.90
	std.dev	14.99	41.31	11.82	41.31	77.93	71.22	81.66	109.37
	max	80.7	170.2	43.5	170.2	260.0	265.0	332.0	280.12
	min	20.0	43.3	3.1	43.3	7.1	3.30	39.01	0.40
	95% C.I	39.05/60.49	68.46/127.56	4.81/21.79	68.46/127.56	44.99/156.48	38.98/140.88	65.56/182.40	57.65/214.13

5.1.4 Discussion

The aim of this investigation was to identify the most appropriate algorithms for the determination of heel-strike and toe-off using kinematic techniques during overground running. A reliable algorithm must be both reliable and accurate allowing the gait cycle to be separated into phases of stance and swing.

The results suggest that heel-strike and toe-off during running are most accurately determined using algorithms from different manuscripts. Heel-strike was most accurately determined using the Alton et al. (1998), O'Connor et al. (2007) and Dingwell et al. (2001) algorithms, which use the position of the lateral malleolus marker, foot velocity algorithm and the first incidence of peak knee extension. Toe-off was most appropriately determined via the Dingwell et al. (2001) and Schache et al. (2001) knee extension and 1st metatarsal velocity methods. The mean errors for event detection appear to correspond to those reported by other studies, with the exception of the Mickelborough et al. (2000) method which was confounded by repeatability issues. That is, the vertical velocity of the foot markers often exhibited multiple maxima and/or minima causing gait events to be located incorrectly. This is common when applying algorithms designed for walking to running data.

In conclusion the Alton et al. (1998), Dingwell et al. (2001), O'Connor et al. (2007) and Schache et al. (2001) event detection methods represent simple and robust methods for determining heelstrike and toe-off events during running that do not require 3-D analysis to employ. An argument is therefore presented for the utilization of these algorithms when force data is unavailable.

5.2 Pilot study 2 – determination of gait events using an externally mounted tibial accelerometer.

5.2.1 Introduction

Whilst previous investigations have used accelerometers to define the stance phase of running (Boyer and Nigg, 2004; Nigg et al. 2004), they have not described the mechanism by which they did so. A method was developed for the determination of heel-strike and toe-off events using the signal from an accelerometer mounted to the shank, measuring accelerations along the longitudinal axis of the tibia (Figure 5.1). The rapid change in the axial tibial acceleration signal as a result of footstrike was regarded as a highly precise and repeatable measurement of heel strike accuracy. Therefore heel-strike was determined as the onset of the peak tibial shock using a threshold of zero which was employed and had to be crossed by a minimum of 20 frames in order to be implemented to prevent false detection. Toe-off was determined using target pattern recognition with a 2% tolerance as the first plateau in the descent phase of the second peak of the axial tibial acceleration time-curve. Toe-off was employed at this point based on the information gathered using a combination of tibial acceleration and 3-D kinematic information whereby toe-off was found to occur immediately prior to max plantar flexion as the foot rolls onto the forefoot i.e. metatarsals which facilitated the development of plateau phase of the tibially mounted accelerometer signal.

The aim of this pilot study was to determine whether specific gait events could be accurately and consistently identified using the method explained, applied to data collected from an externally mounted accelerometer attached to the distal tibia by contrasting the predicted events to those detected using force data.

5.2.2 Methods

5.2.2 (i) Participants

Sixteen participants consisting of eleven males and five females (Age 29.38 ± 5.68 years; Height 1.73 ± 4.87 m; body mass 67.83 ± 10.65 kg), took part in this investigation. The study was approved by the School of Psychology ethical committee, and all participants provided written informed consent.

5.2.2 (ii) Tibial acceleration

Tibial accelerations were obtained using the overground protocol outlined in chapter 5.

5.2.2 (iii) Data Processing

The accuracy of the accelerometer method was determined by comparing events in the acceleration signal to a gold standard method, in which event detection is identified through vertical GRF data. A threshold of 20 N for the vertical GRF component was chosen to determine the time onset of both heel-strike (above 20 N) and toe-off (below 20 N). The difference in the time of the occurrence of both heel-strike and toe-off events identified using the accelerometer method and the force platform method were calculated.

Average and absolute errors were quantified as the average (the net discrepancy indicative of the magnitude of the difference which is influenced by the direction of the time difference between methods) and absolute (the absolute discrepancy indicative of the magnitude of the difference irrespective of direction) discrepancy between the two methods for identifying the event times. A negative value represented the occurrence of the accelerometer signal event prior to the vertical GRF event, a positive value was recorded when the vertical GRF event occurred first.

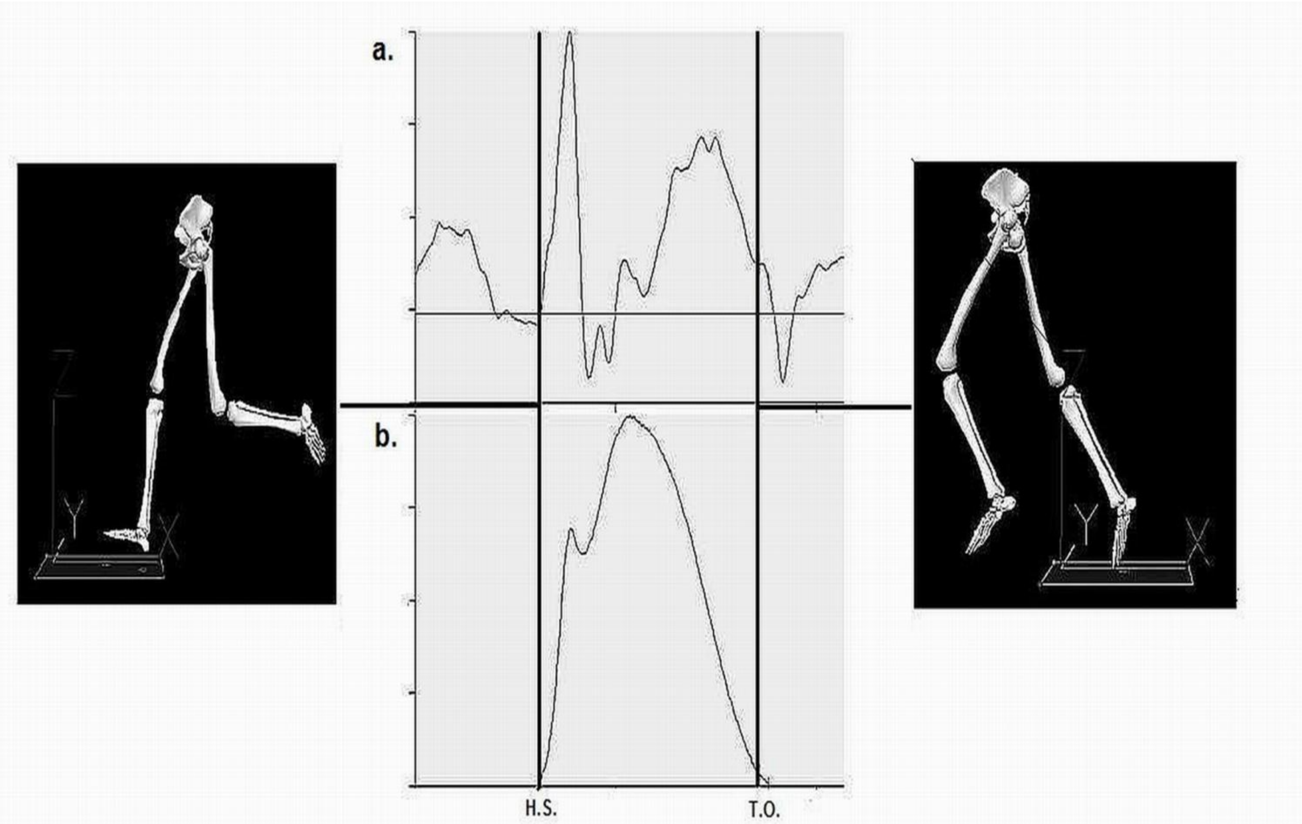

Figure 5.1: The axial acceleration signal (a), and the vertical component of the ground reaction force (b), indicating the timing of footstrike (H.S.) and toe off (T.O.) as determined by the method derived from the accelerometer.

5.2.2 (iv) Statistical Analysis

Heel strike and toe-off events determined via the force platform and accelerometer techniques were implemented using Visual 3-D software (C-Motion, Germantown, USA). Descriptive statistics (means, minimums and maximums) and 95% confidence intervals were calculated for the timings of the events using SPSS 19.0 (SPSS Inc, Chicago, USA).

5.2.3 Results

Table 5.3 presents the accuracy in defining footstrike and toe-off events using the tibially mounted accelerometer in comparison to the force platform.

Table 5.3 Error (ms) of heel-strike and toe-off determination (means, standard deviation minimum maximum and 95% confidence intervals).

	Heel-Strike	**Toe-Off**
Average error (ms)		
Mean	1.68	-3.59
Standard deviation	8.27	1.01
Max	16.6	1.5
Min	-19	-8.56
95% C.I	-2.94 – 6.24	-5.4– -1.78
Absolute error (ms)		
Mean	5.46	5.00
Standard deviation	7.51	9.75
Max	21.2	0.89
Min	0.4	9.22
95% C.I	1.89 – 9.03	3.49 – 8.53

5.2.4 Discussion

Attachment of the accelerometer to the distal tibia during overground running provided a distinguishable and repeatable pattern, presenting a clearly discernible and reliable pattern of heel strike and toe-off with mean and absolute errors at least comparable to those obtained using kinematic information. As such the results of this pilot study suggest that gait events can be reliably and accurately detected using a shank mounted accelerometer provided that due care is taken when mounting the device as the signal obtained from the accelerometer can be influenced by the security of the mounting and centripetal acceleration due to angular motion of the shank in the sagittal plane (Lafortune and Hennig, 1991; Greenhalgh and Sinclair, 2012b).

5.3 General discussion on identification of gait events in the absence of force information

The results of pilot investigations 1 and 2 in this chapter suggest that gait events can be defined accurately in the absence of force information. However, these pilot investigations were conducted using overground analyses thus the findings cannot be generalized to the treadmill. Therefore an informed decision must be made regarding the technique utilized to determine gait events when examining the 3-D kinematics of treadmill running. Whilst Fellin et al. (2010b) examined the effectiveness of kinematic gait events during treadmill running using an integrated force platform; they acknowledged the limitations associated with a low powered treadmill belt which is in contrast to the current study where a high powered treadmill is used.

It was found that during overground running; footstrike could be accurately identified using kinematic techniques that centre around the vertical position/velocity of the foot and foot markers. However, given the potential influence of the high powered treadmill belt which increases the horizontal component of foot motion in comparison to overground (Zeni et al. 2008) it was determined that these techniques were unsuitable for the current study (see appendix C).

The accelerometer technique also appears to be effective for overground analyses; however its effectiveness when identifying gait events during treadmill running is not yet known. Greenhalgh and Sinclair, (2012b) documented that the profile of the tibial acceleration signal was influenced by the angular velocity of the tibia in the sagittal plane. Given that the motion of the tibia may differ between overground and treadmill locomotion and that the determination of footstrike using the accelerometer depends on a precise threshold crossing it was determined that this technique needs further investigation before being adopted for treadmill running.

It was determined therefore that the most appropriate technique for the identification of both footstrike and toe-off events is the Dingwell et al. (1998) kinematic method. This method was found to be one of the most accurate of the kinematic methods of identifying footstrike and is not likely to be influenced by the horizontal motion of the treadmill belt. This method was also found to be the most accurate of the kinematic methods of identifying toe-off. This technique was designed specifically for treadmill locomotion, with footstrike deemed to be the first time when peak knee extension occurs and toe-off determined as the second occurrence of peak knee extension.

5.4 Pilot study 3 – validation of treadmill belt velocity

Throughout the course of the project the treadmill was used with a 0% gradient to remain consistent with overground running. This was quantified using a spirit level prior to the commencement of each investigation. In addition to this it was also necessary to determine that the actual treadmill velocity corresponded with the displayed data. The true velocity of the motorized belt was determined by placing a retro-reflective marker on the surface and recording the time taken to complete 10 revolutions. As the camera system is capable of sampling at very high frequencies it represents both an accurate and practically significant method that could be used for a range of belt velocities (Groot et al. 2006).

This method was used to time required for ten revolutions (6.25m) of the belt. Belt velocity was calculated as a function of the distance covered divided by the time taken ($m.s^{-1}$) then converted to $km.h^{-1}$ by multiplying by 3.6. This approach was used with and without a runner, giving a value that can be compared to the displayed velocity.

Table 5.4 and figure 5.2 present the displayed and measured treadmill belt velocities which provide an indication of the accuracy of the treadmill velocity.

Table 5.4: Comparison between displayed and measured velocities of the treadmill

Displayed belt Velocity (km.h^{-1})	Actual belt Velocity(km.h^{-1})
5	4.968
7.5	7.516
10	10.08
12.5	12.422
14.4	14.004
15	15.084

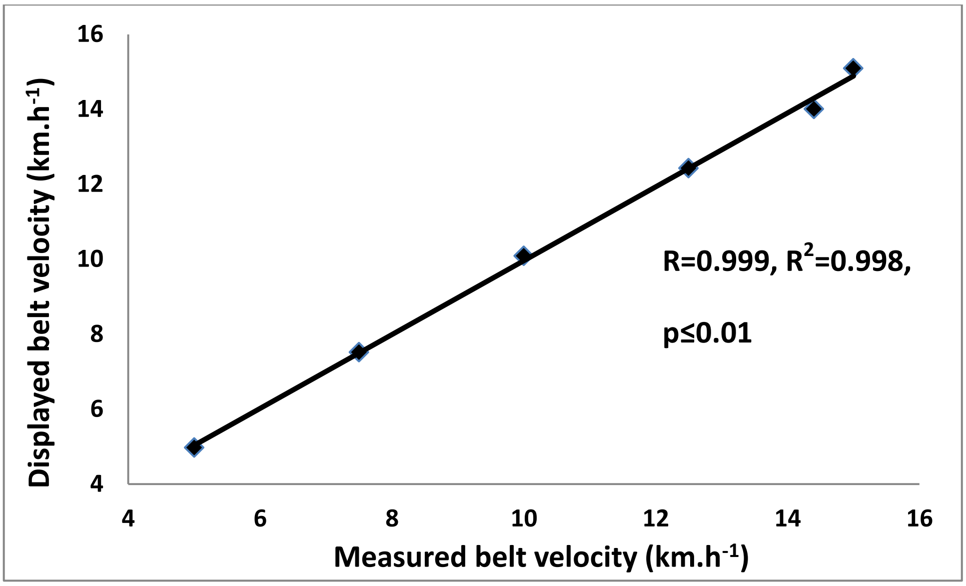

Figure 5.2: Displayed and measured belt velocities in km.h^{-1}.

The table above shows that the presented errors were never more than 1%, thus it is concluded that the velocity of locomotion is acceptable during treadmill running.

5.5 Main study

5.5.1 Introduction

The treadmill is now a common mode of exercise, and is becoming more and more popular (Corey 2005). Since the early 1980's the sport of running has changed dramatically, with a significant increase in the number of treadmill runners (Milgrom et al. 2003). Runners' World suggests that, 40 million people in the U.S alone run using treadmills. Treadmills were traditionally used in clinical and laboratory research, but are now used extensively in both fitness suites and homes. Treadmills allow ambulation at number of velocities whilst indoors in a safe controlled environment.

A number of studies investigating the mechanics of human movement have been conducted using the treadmill. The treadmill presents an environment where variables such as velocity and gradient can be standardized and reproduced consistently (Schache et al. 2001). Furthermore, the treadmill allows a larger number of steps to be captured and ensures that continuous movement kinematics are obtained. Thus the treadmill may facilitate a more repeatable pattern of movement in comparison to the short discontinuous trials associated with overground analyses (Fellin et al. 2010a), although this is advantageous it must be demonstrated that the treadmill does not alter the mechanics of the examined movements in comparison to overground motion (Brand and Crowninshield, 1984). There remains debate regarding the assumption that treadmill running approximates overground running. A number of investigations have been conducted examining the biomechanical differences between the two conditions (Nigg et al. 1995; Schache et al. 2001; Fellin et al. 2010a; Riley et al. 2008; Frishberg, 1983); the results however are often conflicting.

Using a theoretical literature review Van Ingen Schenau, (1980) proposed that the mechanics of overground and treadmill locomotion are similar provided that velocity is maintained. A number of studies have examined the kinematic differences between overground and treadmill walking. Lee and Hidler, (2007) established that peak flexion and extension measures of the lower extremities did not differ between the two conditions. Alton et al. (1998), Matsas et al. (2000) and Riley et al. (2007) found comparable sagittal plane knee kinematics during overground and treadmill locomotion. Strathy et al. (1983) found that knee joint angular kinematics in the coronal and transverse planes did not differ significantly between the two conditions. Alton et al. (1998) and Riley et al. (2007) reported significantly greater hip ROM and flexion angles during treadmill locomotion.

The kinematics of running have also been compared between overground and treadmill locomotion. Frishberg, (1983), Gamble et al. (1988) and Schache et al. (2001) observed that overground running was associated with increased hip flexion at initial contact, whilst Schache et al. (2001) found no alterations in transverse plane hip motion between the two conditions. There is currently a paucity of comprehensive comparisons regarding the 3-D kinematics of the lower extremities during treadmill and overground running during the stance phase. Riley et al. (2008) examined the differences in hip, knee and ankle joint kinematics from both treadmill and overground motion. However they examined only maximum and minimum angles of the full gait cycle, therefore as the majority of these occurred during the swing phase; angles during the stance phase were not compared. Similarly Fellin et al. (2010a) investigated lower extremity motion during both treadmill and overground locomotion; their examination utilized a trend symmetry design which is an effective method of comparing the similarities between kinematic curves, but it does not examine the differences in lower extremity angulation between the two conditions. Furthermore, investigations that have been conducted to date, have been restricted to discrete kinematic parameters and have thus failed to consider the ROM and ROM from footstrike to peak angle during stance.

As stated previously distance runners are known to be susceptible to overuse injuries (Taunton et al. 2003). During gait, cyclical loading at footstrike produces transient shockwaves which propagate through the musculoskeletal system. Research suggests that a relationship exists between the magnitude/frequency of these transient waves and the development of overuse injuries (Folman et al, 1986; Collins and Whittle 1989; Voloshin and Wosk, 1982). Kinematic factors may also have a significant influence on the incidence of overuse injuries given that they may modify the typical alignment of the lower extremities during the stance phase (Eddington et al. 1990). Runners have the option of conducting their training on the treadmill or overground. Unfortunately, in contrast to overground training, there is no epidemiological information available for treadmill running (Milgrom et al. 2003).

An integrated approach that simultaneously measures the kinetics and kinematics of running is needed to evaluate and compare running on level ground and the treadmill. Clinically this knowledge is important to gain an understanding of the susceptibility of treadmill runners to overuse injuries. If significant variations in axial impact shock and stance phase kinematics exist between treadmill and overground locomotion, there may be implications regarding the relative susceptibility of runners to lower-extremity overuse injuries. Furthermore, this may have significant implications for the design and prescription of running shoes.

There has yet to be specific shoes developed for treadmill running, which, to date, have been based on biomechanical data obtained from overground motion, thus runners are forced to wear the same shoes as they do outdoors. Which as previously stated may have implications regarding their applicability to treadmill locomotion. The purpose of this study is therefore to determine whether running shoes specifically designed for treadmill locomotion are necessary.

The aims of the current investigation were twofold 1. to assess the extent to which the stance phase kinetics and kinematics of overground and treadmill locomotion are similar during running. 2. determine whether different footwear designs may be necessary for treadmill locomotion. Specifically the tibial acceleration and 3-D angular kinematics of the lower extremity joints were observed during overground running and compared to the corresponding data from the treadmill.

5.5.2 Methods

5.5.2 (i) Participants

Eleven males and one female who were free from musculoskeletal injury volunteered to take part in this study. Participants were active recreational runners engaging in training at least 3 times per week whilst completing a minimum of 25 km per week and had previous experience of treadmill running. The mean characteristics of the participants were; age 22.5 ± 4.2 years, height 1.71 ± 0.06m and body mass 75.4 ± 8.4 kg. An a priori power analysis was conducted using the Hopkins method based on a moderate effect size and a power measure of 80%, which suggested that 12 subjects were adequate for the design. The study was approved by the School of Psychology ethical committee, and all participants provided written informed consent.

5.5.2 (ii) Procedure

5.5.2 (iii) 3-D Kinematics

All kinematic data were captured at 250 Hz via an eight camera motion analysis system (Qualisys Medical, Goteburg, Sweden). Two separate camera systems were used to collect each mode of running. Calibration of the Qualysis^{TM} systems was performed before each data collection session. Only calibrations which produced average residuals of less than 0.85 mm for each camera for a 750.5 mm wand length and points above 4000 in all cameras were accepted prior to data collection. The order in which participants performed in each condition was counterbalanced.

The marker set used for the study was based on the CAST technique (Cappozzo et al. 1995). A static trial was conducted with the participant in the anatomical position allowing the positions of the anatomical markers to be referenced in relation to the tracking clusters, following which they were removed. Markers used for tracking remained in place for the duration of the treadmill and overground analyses.

Retro-reflective markers were attached to the 1st and 5th metatarsal heads, medial and lateral malleoli, medial and lateral epicondyle of the femur, greater trochanter of the right leg, iliac crest, anterior superior iliac spines and posterior superior iliac spines with tracking clusters positioned on the shank and thigh. Hip joint centre was determined based on the Bell, et al. (1989) equations via on the positions of the PSIS and ASIS markers. Each rigid cluster comprised four 19mm spherical reflective markers mounted to a thin sheath of lightweight carbon fibre with length to width ratios of 2.05:1 and 1.5:1 for the femur and tibia respectively, in accordance with Cappozzo et al. (1997) recommendations.

5.5.2 (iv) Tibial Accelerations

A tri-axial (Biometrics ACL 300, Units 25-26 Nine Mile Point Ind. Est. Cwmfelinfach, Gwent United Kingdom) accelerometer sampling at 1000Hz was utilized to measure axial accelerations at the tibia. The device was mounted on a piece of lightweight carbon-fibre material (figure 5.3). The combined weight of the accelerometer and mounting instrument was 9g. The voltage sensitivity of the signal was set to 100mV/g, allowing adequate sensitivity with a measurement range of ± 100 g.

The device was attached securely to the distal anterio-medial aspect of the tibia 8 cm above the medial malleolus in alignment with its longitudinal axis in accordance with the Sinclair et al. (2010) recommendations. This location was selected to decrease the influence that rotational motion about the ankle can have on the acceleration magnitude (Lafortune and Hennig, 1991). Precautions were taken to ensure mounting consistency across subjects. The Strong adhesive tape was placed over the device and the leg lower to avoid overestimating the peak positive acceleration due to tissue artefact. The device was attached as close to the tibia as possible, the skin on overlying the bone itself was stretched thus ensuring a more rigid coupling between accelerometer and tibia. The accelerometer analogue signal was recorded by a Biometrics DataLog system (Biometrics Units 25-26 Nine Mile Point Ind. Est. Cwmfelinfach, Gwent United Kingdom) securely fastened to the participant via a back pack. This allowed the participants to be as free moving as possible during data acquisition.

Figure 5.3: Positioning of the accelerometer on the distal anterio-medial aspect of the tibia.

5.5.2 (v) Overground

In the overground condition participants ran at 4.0 $m.s^{-1}$ in one direction across a 22 m long biomechanics laboratory floor (Altrosports 6 mm, Altro Ltd, Letchworth Garden City, Hertfordshire). Running velocity was monitored using infrared timing gates Newtest 300 (Newtest, Oulu Finland); a maximum deviation of ± 5% from the set velocity was allowed. Runners completed five successful trials. A successful trial was defined as one within the specified velocity range, where all tracking clusters were in view of the cameras and with no evidence of gait modification due to the experimental conditions.

5.5.2 (vi) Treadmill

A Woodway™ (ELG, Steinackerstrasse D-79576 Weil Rhein-Germany) high power slatted treadmill maintained at a gradient of 0% was used throughout. Participants were given a five minute habitation period, in which participants ran at the determined velocity, following which the treadmill was stopped for 30's, and participants dismounted the treadmill before mounting the treadmill for data analysis in accordance with the Alton et al. (1998) recommendation. When participants indicated that they were ready to begin, the treadmill was started and the velocity of the belt was gradually increased until the speed matched that of overground locomotion (4.0$m.s^{-1}$). Five trials were recorded.

Given that the treadmill did not feature an integrated force platform, heel strike and toe-off events during both treadmill and overground running were determined using kinematic data based on the Dingwell et al. (2001) method. Footstrike was deemed to be the first occurrence of peak knee extension and toe-off was determined as the second occurrence of the peak knee extension. This technique was selected in accordance with Fellin et al. (2010ab) due the high powered treadmill belt and its potential influence on both 3-D kinematic techniques which utilize the vertical position of the foot and toe markers. Furthermore it was selected in favour of the accelerometer technique as the tibial acceleration signal has been shown to be sensitive to the angular motion of the tibia (Greenhalgh and Sinclair 2012b) who noted that the correction required for angular information would influence the determination of specific gait events using the accelerometer. As the tibial segment may move differently during treadmill locomotion in comparison to overground, the accuracy of the accelerometer technique is therefore unknown during treadmill locomotion whereas the Dingwell et al. (2001) method was developed for treadmill locomotion.

5.5.2 (vii) Data Processing

Trials were processed in Qualisys Track Manager in order to identify anatomical and tracking markers then exported as C3D files. Kinematic parameters were quantified using Visual 3-D (C-Motion, Germantown, USA) after marker data was filtered using a low pass Butterworth 4^{th} order zero-lag filter at a cut off frequency of 10 Hz which was selected as being the frequency at which 95% of the signal power was below. 3-D kinematics of the hip, knee and ankle joints were calculated using an XYZ cardan sequence of rotations. All data were normalized to 100% of the stance phase then processed gait trials were averaged. 3-D kinematic measures from the hip, knee and ankle which were extracted for statistical analysis were 1) angle at footstrike, 2) angle at toe-off, 3) ROM from footstrike to toe-off during stance, 4) peak angle during stance and 5) relative ROM from footstrike to peak angle. In addition to this the eversion/tibial internal tibial internal rotation (EV/TIR) ratio was quantified in accordance with De Leo et al. (2004) as the relative eversion ROM / the relative tibial internal rotation ROM. Finally, in order to obtain a measure of COM progression during the stance phase the amount of pelvic segment movement in the anterior-posterior direction was also quantified.

A FFT analysis of the acceleration signal revealed that more than 95% of the signal power was below 60 Hz. Therefore, the acceleration signal was filtered using a 60Hz Butterworth 4^{th} order zero-lag filter in accordance with the Hennig and Lafortune, (1991) guidelines, to prevent any resonance effects on the acceleration signal. Peak tibial acceleration was defined as the highest positive acceleration peak measured during the stance phase. To analyze data in the frequency domain, a FFT function was performed and median power frequency content of the acceleration signals were calculated in accordance with (Lafortune and Hennig, 1995).

5.5.2 *(viii) Shoes*

The shoes utilized during this study consisted of a Saucony Pro Grid Guide II; they differed in size only (sizes 6, 7 and 9 in men's shoe UK sizes).

5.5.2 (ix) Statistical analysis

Descriptive statistics (mean ± standard deviation) were calculated for the outcome measures. To compare differences in 3-D kinematic and tibial acceleration parameters paired t-tests were utilized with significance accepted at $p \leq 0.05$. The Shapiro-Wilk statistic for each condition confirmed that the data were normally distributed. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA).

5.5.3 Results

Tables 5.5-5.9 and figures 5.4 and 5.5 present information regarding the kinetic and kinematic differences between treadmill and overground locomotion.

5.5.3 (i) Kinetic and temporal parameters

Table 5.5: Kinetic and temporal variables (means and standard deviations) as a function of condition (* = Significant main effect).

	Overground	**Treadmill**	
Peak impact shock (g)	8.59 ± 4.52	5.97 ± 2.78	*
Median power frequency (Hz)	13.81 ± 5.08	12.14 ± 5.08	
Stance phase pelvic COM progression (m)	0.91 ± 0.14	0.032 ± 0.02	*
Stance time (ms)	220.52 ± 21.45	214.16 ± 30.21	

The results indicate that the magnitude of the peak axial impact shock peak was significantly higher during the overground condition $t_{(11)} = 2.40$, $p \leq 0.05$ in comparison to the treadmill. Furthermore the of progression COM position in the X (anterior-posterior) direction was found to be significantly greater $t_{(11)} = 21.0$, $p \leq 0.01$ in the overground condition in comparison to the treadmill.

5.5.3 (ii) 3-D kinematic parameters

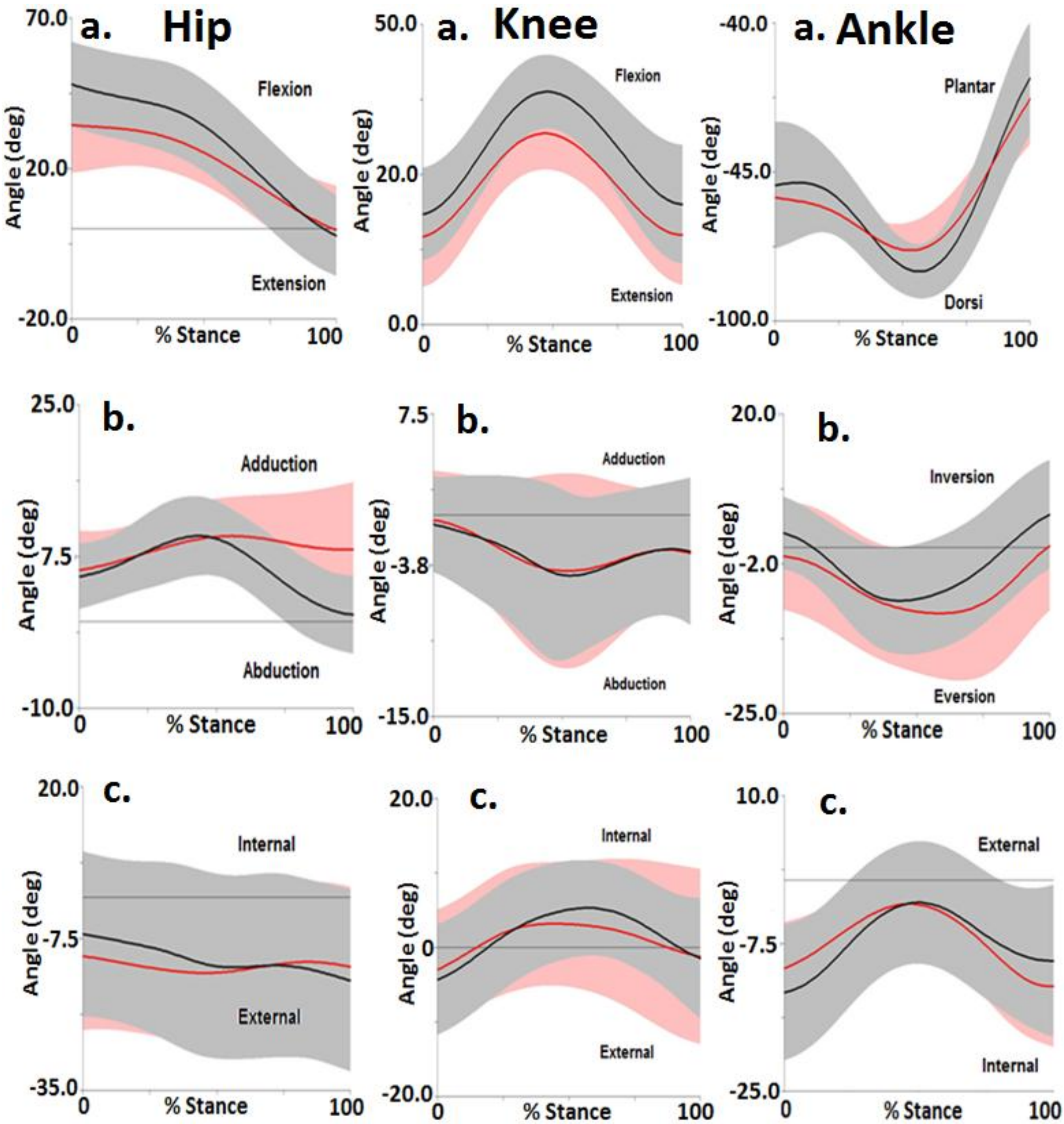

Figure 5.4: Mean and standard deviation hip, knee and ankle joint kinematics in the a. sagittal, b. coronal and c. transverse planes for overground (black line) and treadmill (red line), running (shaded area is 1 ±SD, treadmill=grey shade and overground = red).

Table 5.6: Hip joint kinematics (mean and standard deviation) as a function of condition (* = Significant main effect).

	Overground	**Treadmill**	
X (+=flexion/-=extension)			
Angle at Footstrike (°)	47.10 ± 13.45	35.11 ± 12.65	*
Angle at Toe-off (°)	-2.24 ± 23.78	1.45 ± 14.69	
Range of Motion (°)	49.61 ± 10.42	32.63 ± 10.46	*
Relative Range of Motion (°)	2.27 ± 2.72	1.45 ± 2.0	
Peak Flexion (°)	49.31 ± 8.6	36.61 ± 7.90	
Y (+=adduction/-=abduction)			
Angle at Footstrike (°)	5.71 ± 3.36	6.08 ± 4.46	
Angle at Toe-off (°)	1.53 ± 4.30	6.09 ± 7.59	
Range of Motion (°)	4.01 ± 5.38	0.27 ± 7.63	
Relative Range of Motion (°)	5.45 ± 2.78	5.61 ± 7.38	
Peak Adduction (°)	11.16 ± 3.84	11.69 ± 5.65	
Z (+=internal/- =external)			
Angle at Footstrike (°)	-5.68 ± 15.32	-11.25 ± 11.68	*
Angle at Toe-off (°)	-13.36 ± 17.55	-10.72 ± 17.96	
Range of Motion (°)	8.40 ± 5.38	0.41 ± 13.51	*
Relative Range of Motion (°)	10.29 ± 4.85	8.93 ± 6.76	
Peak External Rotation (°)	-15.97 ± 16.53	-20.08 ± 13.74	

The hip flexion angle at footstrike was found to be significantly $t_{(11)} = 5.48$, $p \leq 0.01$ greater the overground condition. Furthermore, it was also observed that the hip exhibited a significantly greater $t_{(11)} = 5.87$, $p \leq 0.01$ overall ROM in the overground condition. In the transverse plane the external rotation magnitude at footstrike was found to be significantly larger in the treadmill condition $t_{(11)} = 2.43$, $p \leq 0.05$. The results also demonstrate that the hip exhibited significantly greater $t_{(11)} = 2.99$, $p \leq 0.05$ transverse plane ROM in the overground condition in comparison to the treadmill.

Table 5.7: Knee joint kinematics (mean and standard deviation) as a function of condition (* = Significant main effect).

	Overground	**Treadmill**	
X (+=flexion/-=extension)			
Angle at Footstrike (°)	17.96±7.01	19.08±6.31	
Angle at Toe-off (°)	19.05±7.85	16.94±7.72	
Range of Motion (°)	1.09±8.71	2.11±8.05	
Relative Range of Motion (°)	21.52±7.49	15.37±6.22	
Peak Flexion (°)	39.48±5.20	34.46±5.69	*
Y (+=adduction/-=abduction)			
Angle at Footstrike (°)	-0.44±3.59	1.67±5.58	
Angle at Toe-off (°)	-2.57±5.23	-1.88±5.76	
Range of Motion (°)	2.12±8.05	3.54±4.57	
Relative Range of Motion (°)	4.97±2.66	4.63±4.18	
Peak Abduction (°)	-5.41±5.42	-2.96±8.97	
Z (+=internal/- =external)			
Angle at Footstrike (°)	-4.54±7.15	-3.61±7.78	
Angle at Toe-off (°)	-1.34±6.95	-0.45±10.70	
Range of Motion (°)	3.20±4.01	3.16±9.94	
Relative Range of Motion (°)	11.16±4.97	10.02±7.02	
Peak Internal Rotation (°)	6.62±6.03	6.41±8.03	

In the sagittal plane the results indicate that peak knee flexion in the overground condition was shown to be significantly greater $t_{(11)} = 2.89$, $p \leq 0.05$ than the treadmill.

Table 5.8: Ankle joint kinematics (mean and standard deviation) as a function of condition (* = Significant main effect).

	Overground	**Treadmill**	
X (+=plantar/-=dorsi)			
Angle at Footstrike (°)	-68.73 ± 8.98	-70.03 ± 4.96	
Angle at Toe-off (°)	-50.14 ± 9.93	-54.04 ± 6.82	
Range of Motion (°)	18.32 ± 7.40	13.94 ± 7.24	
Excursion from footstrike to peak angle (°)	20.26 ± 10.84	15.99 ± 8.06	*
Peak Dorsi-Flexion (°)	-86.63 ± 3.17	-84.13 ± 4.80	
Y (+=inversion/-=eversion)			
Angle at Footstrike (°)	1.99±5.23	-4.50±10.11	
Angle at Toe-off (°)	5.15±7.41	-0.51±7.25	
Range of Motion (°)	3.16±4.97	3.99±8.03	
Excursion from footstrike to peak angle (°)	11.15±4.63	11.00±6.38	
Peak Eversion (°)	-9.16±7.83	-15.50±8.85	*
Z (-=internal/± =external)			
Angle at Footstrike (°)	-13.53±5.14	-9.62±4.93	*
Angle at Toe-off (°)	-10.39±5.96	-8.95±9.92	
Range of Motion (°)	3.15±3.54	0.67±11.23	
Excursion from footstrike to peak angle (°)	10.87±3.45	8.48±4.11	*
Minimum Internal Rotation (°)	-2.7 ± 4.3	-1.1 ± 3.0	

In the sagittal plane the results indicate that the excursion ROM was significantly greater $t_{(11)} = 2.45$, $p \leq 0.05$ in the overground condition in comparison to the treadmill. In the coronal plane the results indicate that the magnitude of peak eversion was significantly greater $t_{(11)} = 3.36$, $p \leq 0.01$ in the treadmill condition in comparison to overground. In the transverse plane it was observed that at footstrike external rotation magnitude was significantly $t_{(11)} = 3.30$, $p \leq 0.05$ greater in the overground condition. In addition the results indicate that this motion from footstrike to peak angle in terms of ROM was significantly greater $t_{(11)} = 2.57$, $p \leq 0.05$ in the overground condition in comparison to the treadmill.

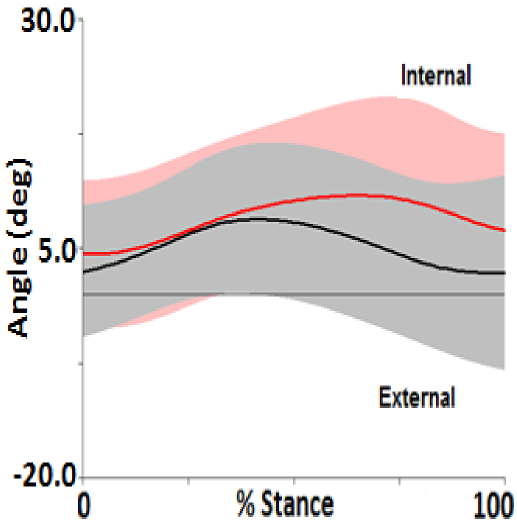

Figure 5.5 Mean and standard deviation tibial internal rotation kinematics for overground (black line) and treadmill (red line), running (shaded area is 1 ±SD, treadmill=grey shade and overground = red).

Table 5.9: Tibial internal rotation kinematics (mean and standard deviation) as a function of condition (* = Significant main effect).

	Overground	**Treadmill**	
Tibial Internal Rotation			
Z (+ =internal/ - =external)			
Angle at Footstrike (°)	3.19 ± 7.18	5.05 ± 8.05	
Angle at Toe-off (°)	2.69 ± 8.78	7.66 ± 8.80	
Range of Motion (°)	0.59 ± 4.03	2.48 ± 4.39	
Relative Range of Motion (°)	5.64 ± 3.81	10.01 ± 6.24	
Peak Tibial Internal Rotation (°)	8.90 ± 2.35	15.08 ± 8.92	*
EV/TIR ratio	1.91 ± 0.62	1.18 ± 2.24	

The results indicate that the treadmill condition was associated with significantly $t(11) = 2.36$, $p \leq 0.05$ greater peak tibial internal than the overground condition.

5.5.4 Discussion

The aim of this study was to provide a kinetic and 3-D kinematic comparison of treadmill and overground running. This study represents the first to comparatively examine the synchronous tibial acceleration and 3-D kinematic parameters when running in the two conditions. The results indicate that several kinetic and kinematic differences were observed between the two running modalities.

The results indicate that tibial accelerations were significantly lower during treadmill locomotion in comparison to overground. The transient shockwave in running is capable of generating significant forces in the joints and other structures of the lower limbs, and is linked to the aetiology of a variety of bony and soft tissue disorders (Whittle, 1999). It is important to acknowledge the link between these forces and pathological conditions, since the magnitude of these forces and by implication; the incidence of these conditions can be reduced by attenuating the impact magnitude (Whittle, 1999). The interaction between foot and surface has a significant effect in the development of lower overuse extremity injuries. Research investigating the cushioning properties of different surfaces suggests that surface stiffness may have a significant effect on the magnitude of the impact shock experienced during the landing phase of gait. It is hypothesized that the more compliant treadmill belt provided an additional deceleration mechanism in comparison to stiffer laboratory floor (Kim and Voloshin 1992).

Such impact shock patterns represent the capacity of the treadmill surface to attenuate the magnitude of the impact shock that is applied to the lower extremities (Logan, 2007). These findings lead to the conclusion that treadmill running may be associated with a lower incidence of impact related overuse injuries (Misevich and Cavanagh 1984). Furthermore, it appears based on these findings that older runners and heavy pounders who characteristically elect high impact forces may be less susceptible to overuse injuries if they choose to conduct their training on the treadmill rather than overground.

It has been proposed that the mechanics of treadmill locomotion are similar to overground provided that velocity remains constant (Van Ingen Schenau, 1980). However, in this study significant differences between overground and treadmill running were found for sagittal plane hip rotation. Overground running was associated with increased peak hip flexion and flexion angle at initial contact. This concurs with the findings of Schache et al. (2001) who observed similar increases in hip flexion during overground running.

Overground running in this experiment was also associated with an increased ROM in hip flexion-extension, which was a product of increased hip flexion at footstrike during overground running, as hip flexion at toe-off was found to be similar for the two conditions. This finding agrees with the findings of Frishberg (1983), Gamble et al. (1988) and Schache et al. (2001). These findings may be attributable to the reduced stride lengths that have been observed previously during treadmill running (Wank et al. 1998). Furthermore, it is hypothesized that the slatted treadmill belt may have acted as a visual cue which served to further accentuate this adaptation causing the large difference between the two conditions. Future, research may therefore wish to investigate the influence of both slatted and smooth treadmill belts of the 3-D kinematics of running.

Furthermore, Alton et al. (1998) hypothesized that participants utilized these mechanics as a means of avoiding falling off the back of the treadmill and/or keeping up with the belt speed. The results of the current investigation appear to oppose this notion in that participants did not exhibit similar patterns, despite moving at a greater velocity, as fear of falling and pressure to maintain a stipulated speed would theoretically be amplified by an increased belt velocity. It is also probable that the length of the treadmill utilized during this investigation (1.0m longer than that reported by Alton et al. 1998), decreased participants concern that they might fall off the treadmill. Future investigations may wish to assess subjective feedback from participants in order to determine the underlying mechanisms behind gait alterations.

The increase in peak knee flexion during overground running has not been reported previously. It is proposed that this finding is attributable to the difference in COM progression during overground running as the COM moves over the stance limb the proximal end of the tibia must move forwards, facilitating an increase in knee flexion. Similarly, the significant increase in the relative ROM from footstrike to peak dorsiflexion has not been reported previously within the literature. It is proposed that this is also attributable to the increase in COM progression in the overground condition. Given that the foot is fixed during the majority of the stance phase, forward motion of the COM forces the tibia to move over the ankle joint creating the dorsiflexion ROM. This finding may also relate to differences in surface hardness between the two conditions. The increase in dorsiflexion ROM in conjunction with peak knee flexion may act as a deceleration mechanism which serves to reduce loading of the lower extremity structures (Bobbert et al. 1992).

During treadmill running, the ankle was found to be slightly more dorsiflexed at footstrike. This finding contrasts the findings of Wank et al. (1998), Fellin et al. (2010a) and Nigg et al. (1995), who found decreased ankle dorsiflexion at footstrike. This change in sagittal plane ankle position at foot contact may relate to a change in strike pattern as plantar/dorsiflexion of the ankle is one of the mechanisms by which leg stiffness is regulated (Bishop et al. 2006). It is hypothesized that the reduced stiffness of the treadmill surface may have led to the increased dorsiflexion at footstrike as runners have been found to adjust their leg stiffness in response to differences in surface hardness (Bishop et al. 2006).

The significant increase in eversion and associated tibial internal rotation magnitudes are in contrast to the observations of Fellin et al. (2010a) who reported no differences in rearfoot eversion parameters between treadmill and overground running. This finding may relate to the deformation characteristics of the surface during the treadmill condition and has potential clinical significance. These findings have potential clinical significance and suggest that running on this type of treadmill may be associated with an increased risk from injury as rearfoot eversion and tibial internal rotation are implicated in the aetiology of a number of overuse injuries (Willems et al. 2004; Lee et al. 2010; Taunton et al. 2003; Duffey et al. 2000). Therefore treadmill runners may be at a greater risk from overuse syndromes such as tibial stress syndrome, Achilles tendinitis, patellar tendonitis, patellofemoral pain, illiotibial band syndrome and plantar fasciitis (Willems et al. 2004; Lee et al. 2010; Taunton et al. 2003; Duffey et al. 2000).

A number of previous investigations examining the mechanics of treadmill and overground locomotion attribute the differences between the two conditions to a lack of familiarization to the treadmill protocol (Wall and Charteris, 1981). Matsas et al. (2000) proposes studies reporting significant differences between the two conditions locomotion have generally put little emphasis on subject familiarisation to treadmill locomotion and concluded that differences may disappear following an appropriate accommodation period. The results of this study appear to oppose this claim as a number of significant differences were observed despite the utilization of a five minute accommodation period. Furthermore, the findings of the current investigation appear to be representative and as Matsas et al. (2000) found that reliable kinematic measurements could be obtained following 4 minutes of treadmill habituation.

There are a number of limitations to the current investigation that should be acknowledged. The means by which footstrike and toe-off were determined differed from conventional methods as the treadmill did not feature an integrated force platform. Given this limitation the stance and swing phases were separated using kinematic data using the Dingwell et al. (1998) method. A number of methods have been utilized for the determination of gait events using kinematic data (Alton et al. 1998; Hreljac and Stergiou.; 2000, Zeni et al. 2008; O'Connor et al. 2003; Schache et al. 2001). However, although these computational methods are repeatable they are known to be associated with error when contrasted to the gold-standard method using force platform data (Fellin et al. 2010b).

A possible limitation is that this study observed right foot contact only. Bilateral studies are considered to be more appropriate as symmetry between limbs is unlikely (Cavanagh and Lafortune, 1980). Another prospective restriction of the current investigation is that the results are specific exclusively to the treadmill and surface conditions as well as the velocity of motion and variations in these parameters would likely cause changes in the runners movement strategy, additional work should therefore be conducted examining the effect of different treadmills on gait mechanics.

Based on the results of this investigation it appears that new footwear models are not necessary for treadmill locomotion. Rather, it appears that existing footwear models are appropriate provided that footwear with suitable mechanical characteristics are utilized in the appropriate condition. The results of this study suggest that the treadmill surface may serve to attenuate impact shock but at the expense of greater rearfoot motion. Therefore, it appears that for overground locomotion a cushioned shoe with design features focussed towards shock attenuation would be more appropriate. A large variety of materials are utilized in the cushioning systems of modern running shoes (Shorten, 2000). These include materials such as foamed polymers, viscoelastic materials, air, gases, gels and moulded springs. Materials are selected based on their ability to attenuate shock. Although cushioning materials vary considerably, the principles of cushioning are common to all of them. In addition, given the significant increase in rearfoot motion during treadmill running it is recommended that runners utilize a shoe with design features aimed towards the reduction of calcaneal eversion. Since the very early 1980's most technical running shoes have incorporated anti-pronation or subtalar control devices. Such devices include harder cushioning, heel counters, insole boards, medially posted midsoles and vagus wedges (Shorten, 2000). Shorten, (2000) proposed that anti-pronation devices are designed to work in two ways. Some are designed to stiffen the shoe and thus physically restrain the movement of the subtalar joint. Others modify the shoe cushioning to reduce the lever arm of the GRF in an attempt to decrease the amount of torque that leads to pronation.

The results of this study also suggest that treadmill should be utilized with caution within clinical and research settings in terms of its ability to mimic the mechanics of overground running. Furthermore, given that injury patterns may to differ between the two conditions it is also recommended that runners consider their primary method of training when selecting the most appropriate footwear for their needs as treadmill runners are likely to require footwear with additional medial stability properties, aimed at reducing rearfoot eversion whilst overground runners should consider footwear with more advanced midsole cushioning properties designed to reduce the magnitude of impact transients.

6. <u>Gender differences in distance running: Implications for footwear design.</u>

Publications

1. Sinclair J, Greenhalgh, A, Edmundson CJ, and Hobbs SJ (2012). Gender differences in the kinetics and kinematics of distance running: implications for footwear design, International Journal of Sports Science and Engineering. Vol. 06, No. 02, pp. 118-128

2. Sinclair, J., Taylor, P.J., Edmundson, C.J., Brooks, D., and Hobbs, S.J. (2012). Gender differences in tibiocalcaneal kinematics. Journal of Biomechanics, 40, 628.

Conference Presentations

3. Sinclair J, Taylor PJ, Edmundson, CJ, Brooks and Hobbs SJ (2012). Gender differences in tibiocalcaneal kinematics European congress of Biomechanics, Lisbon, Portugal.

6.1.1 Introduction

Running is the sport of choice for millions of people, both males and females alike (Taunton et al. 2003). A rapid growth in distance running participation has been witnessed amongst the female population (Lilley et al. 2011). This increase in women's running activities has stimulated many sport scientists to investigate the various aspects of female running performance. Although there are numerous health benefits associated with running, the risk of injury is also well documented (Taunton et al. 2003). There are several notable anatomical and physiological differences between males and females that may influence running biomechanics. The average mature male is greater in both height and mass and has a lower body fat percentage (Atwater, 1990). In a study providing anatomical reference data Morris et al. (1982) found that males are on average 0.12m taller than females and 18kg heavier, whilst carrying on average 9% less body fat. Increased muscular mass in males is attributable to the higher levels of testosterone, whilst increases in oestrogen contribute to the higher body fat percentage found in females (Morris et al. 1982).

It has been postulated that differences in structure may predispose females to variations in running mechanics which, over many repetitions, may cause females to sustain different injury characteristics than age matched males. Evidence suggests that females are almost twice as likely to sustain a running related injury such as patellofemoral pain syndrome, stress fractures, iliotibial band syndrome or gluteus medius injury (Geraci and Brown, 2005; Taunton et al. 2003), yet the gender specific aetiology of these injuries are not fully understood (Taunton et al. 2003). Gender differences in kinetics and lower extremity kinematics during running have been suggested as a contributing factor (Ferber et al. 2003;

Schache et al. 2003) and whilst gender differences in lower extremity structure have been studied, little attention has been devoted to differences in running mechanics between genders.

Only a small number of investigations to date have investigated differences in lower extremity joint mechanics between genders during running. Malinzak et al. (2001) investigated gender differences in coronal and sagittal plane knee motion. It was demonstrated that whilst the coronal plane knee excursion was similar between genders, women were found to exhibit less peak knee flexion and a lower ROM in the knee compared to men. Ferber et al. (2003) examined the gender differences in 3-D kinematics of the hip and knee. Female runners exhibited greater peak hip adduction, hip internal rotation and knee abduction compared to men. Whilst informative, these studies did not investigate ankle kinematics or observe the kinetic loading parameters between genders. There has yet to be an investigation which has examined both the kinetics and 3-D kinematics of the lower extremities of male and female runners.

The running shoe acts as the primary interface between the runner and the road, and thus has an important role to play in the management of injuries. A key concern is the demands of specific running footwear for females when compared to men's shoes. Given the relative susceptibility of females to overuse running injuries, a key issue within the discipline of footwear biomechanics that has yet to be addressed is the specific demands of athletic footwear for females. Footwear manufacturers frequently produce footwear for females on the basis of data collected using male participants. This has led to women's running shoes being habitually designed using a scaled down version of a man's shoe with all dimensions reduced proportionally according to the length of the foot (Wunderlich and Cavanagh, 2002). Thus, it is possible that there is a paucity of footwear models that meet the specific needs of female runners both in terms of protection from injury and appropriate fit. As participation in distance running amongst females has increased, new information regarding the biomechanical aspects of female distance running mechanics would be of both theoretical and practical significance. A greater understanding of the differences in running mechanics between male and female runners may also provide an insight into the aetiology of different injury patterns and how these injuries may be attenuated using appropriate footwear designs.

The present study aimed to provide both a kinetic and 3-D kinematic comparison of male and female runners in order to determine 1) the relative susceptibility of females to the proposed mechanisms of overuse injuries and 2) whether females require more specific footwear designs to meet their needs. This examination presents information that may aid footwear manufacturers regarding the design of future shoe models for female runners.

6.1.2 Methods

6.1.2 (i) Participants

Twelve male participants and twelve female participants volunteered to take part in this investigation. All were injury free at the time of data collection and provided written informed consent. Participants were active recreational runners who completed 35km across a minimum of 3 training sessions per week. The mean characteristics of the participants were males; age 25.08 ± 5.30 years, height 1.78 ± 0.04 m and mass 71.33 ± 5.38 kg and females; age 25.04 ± 4.87 years, height 1.68 ± 0.04 m and mass 62.67 ± 3.75 kg. A statistical power analysis was conducted in order to reduce the likelihood of a type II error and determine the minimum number of participants needed for this investigation. It was found that the sample size was sufficient to provide more than 80% statistical power. The procedure was approved by the University of Central Lancashire, School of Psychology ethics committee.

6.1.2 (ii) Procedure

Participants completed overground running at 4.0m.s^{-1}±5% over a piezoelectric force plate (Kistler, Kistler Instruments Ltd., Alton, Hampshire) embedded in the floor (Altrosports 6mm, Altro Ltd,) of a 22 m biomechanics laboratory. Running velocity was quantified using infrared timing gates Newtest 300 (Newtest, Oy Koulukatu, Finland), a maximum deviation of ±5% from the set velocity was allowed. Participants completed a minimum of five successful trials. Stance time during contact with the force plate was determined as the time over which 20N or greater of vertical force was recorded. A successful trial was defined as one within the specified velocity range, where all tracking clusters were in view of the cameras, the foot made full contact with the force plate and with no evidence of gait modification due to the experimental conditions. Dynamic calibration with the same acceptance criteria as outlined in chapter 5 was conducted prior to data collection.

6.1.2 (iii) 3-D Kinematics

3-D kinematics from the lower extremities were obtained using the same protocol as the in chapter 5.

6.1.2 (iv) Tibial accelerations

Tibial accelerations were also obtained using the protocol outlined in chapter 5.

6.1.2 (v) Data Processing

Trials were processed in Qualisys Track Manager in order to identify anatomical and tracking markers then exported as C3D files. Kinematic parameters were quantified using Visual 3-D (C-Motion Inc., Germantown, USA) following the smoothing of marker data using a low-pass Butterworth 4th order zero-lag filter at a cut off frequency of 10Hz. This frequency was selected as being the frequency at which 95% of the signal power was below. 3-D kinematics of the hip knee and ankle joints were calculated using an XYZ cardan sequence of rotations (where X is flexion-extension; Y is ab-adduction and is Z is internal-external rotation). All data were normalized to 100% of the stance phase then processed gait trials were averaged. 3-D kinematic measures from the hip, knee and ankle which were extracted for statistical analysis were 1) angle at footstrike, 2) angle at toe-off, 3) ROM during stance, 4) peak angle during stance and 5) relative ROM from footstrike to peak angle. In addition to this, anatomical measures of pelvic width, hip width, femur length, hip width/femur length ratios and pelvis width/femur length ratios were obtained in accordance with the Horton and Hall (1989) recommendations. Finally the eversion/tibial internal tibial internal rotation (EV/TIR) ratio was quantified in accordance with De Leo et al. (2004) as the relative eversion ROM / the relative tibial internal rotation ROM.

A FFT analysis of the acceleration signal revealed that more than 95% of the signal power was below 60 Hz. Therefore, the acceleration signal was filtered using a 60Hz Butterworth 4^{th} order zero-lag filter in accordance with the Hennig and Lafortune, (1991) guidelines, to prevent any resonance effects on the acceleration signal. Peak positive axial tibial acceleration was defined as the highest positive acceleration peak measured during the stance phase. To analyze data in the frequency domain, a FFT transformation function was performed and median power frequency content of the acceleration signals were calculated. Forces were reported in bodyweights (BWs) to allow normalisation of the data between participants. From the force plate data, peak braking and propulsive forces, stance time, average loading rate, instantaneous loading rate, peak impact force and time to peak impact were calculated. Average loading rate was calculated by dividing the impact peak magnitude by the time to the impact peak. Instantaneous loading rate was quantified as the maximum increase in vertical force between frequency intervals.

6.1.2 (vi) Shoes

The shoes utilized during this study consisted of a Saucony Pro Grid Guide II; they differed in size only (sizes 6, 7 and 9 in men's shoe UK sizes).

6.1.2 (vii) Statistical Analysis

Descriptive statistics including means and standard deviations were calculated for both males and females. Differences in 3-D kinematic parameters, anatomical characteristics, impact shock and impact forces were examined using independent samples t-tests with significance accepted at the $p \leq 0.05$ level. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA). The Shapiro-Wilk statistic for each condition confirmed that the normal distribution assumption was met for the data set.

6.1.3 <u>Results</u>

Tables 6.1-6.5 and figures 6.1 and 6.2 present information regarding the kinetic and kinematic differences between male and female runners.

6.1.3 (i) Temporal, anatomical and kinetic parameters

Table 6.1: Kinetic and temporal variables (mean and standard deviation) as a function of gender (* = Significant main effect).

	Male	**Female**	
Vertical Impact Peak (BW)	1.81 ± 0.51	1.91 ± 0.30	
Instantaneous Loading Rate ($BW.s^{-1}$)	157.27 ± 59.61	155.27 ± 59.99	
Average Loading Rate ($BW.s^{-1}$)	68.43 ± 14.41	76.51 ± 29.21	
Time to Peak Impact (ms)	28.01 ± 6.37	27.92 ± 6.22	
Peak impact shock (g)	5.13 ± 2.67	6.51 ± 2.85	
Median Power Frequency (Hz)	14.03 ± 12.07	13.29 ± 8.33	
Stance time (ms)	210.60 ± 32.45	203.93 ± 29.33	
Peak Braking Force (BW)	0.51 ± 0.14	0.45 ± 0.08	
Peak Propulsive Force (BW)	0.38 ± 0.05	0.38 ± 0.05	
Peak Medial Force (BW)	0.13 ± 0.08	0.12 ± 0.08	
Peak Lateral Force (BW)	0.19 ± 0.03	0.19 ± 0.10	
Pelvis Width (m)	0.29 ± 0.02	0.28 ± 0.02	
Hip Width (m)	0.38 ± 0.05	0.40 ± 0.02	
Femur Length (m)	0.41 ± 0.03	0.39 ± 0.02	
Hip width-femur length ratio	0.92 ± 0.16	1.01 ± 0.10	
Pelvis width-femur length ratio	0.69 ± 0.08	0.71 ± 0.08	

The results indicate that no significant $p>0.05$ differences in kinetic, anatomical or temporal variables exist between male and female runners.

6.1.3 (ii) 3-D kinematic parameters

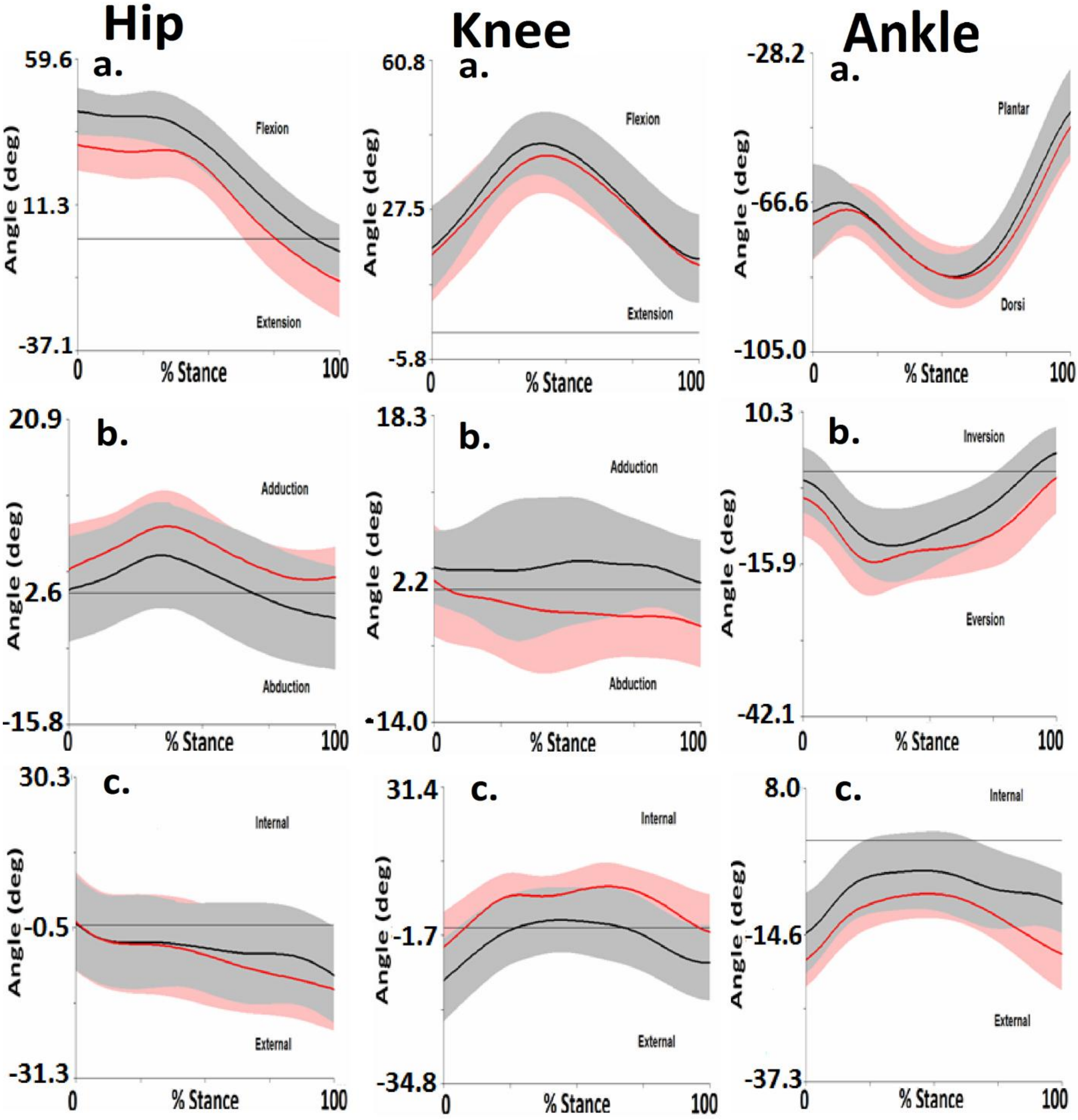

Figure 6.1: Mean and standard deviation hip, knee and ankle joint kinematics in the a. sagittal, b. coronal and c. transverse planes for males (black line) and females (red line), running (shaded area is 1 ±SD, males=grey shade and females = red).

Table 6.2: Hip kinematics (mean and standard deviation) as a function of gender (* = Significant main effect).

	Male	Female	
Hip			
X (+=flexion/-=extension)			
Angle at Footstrike (°)	43.23 ± 5.94	32.48 ± 9.93	*
Angle at Toe-off (°)	-3.63 ± 8.73	-13.26 ± 12.32	*
Range of Motion (°)	46.46 ± 7.52	45.61 ± 6.97	
Relative Range of Motion (°)	2.31 ± 2.86	1.26 ± 1.99	
Peak Flexion (°)	45.53 ± 6.21	33.61 ± 9.49	*
Y (+=adduction - =abduction)			
Angle at Footstrike (°)	1.28 ± 6.50	3.20 ± 4.63	
Angle at Toe-off (°)	-2.25 ± 6.37	2.20 ± 3.76	*
Range of Motion (°)	3.81 ± 2.32	2.15 ± 3.26	
Relative Range of Motion (°)	4.53 ± 3.15	6.72 ± 2.27	
Peak Adduction (°)	6.81 ± 6.41	10.93 ± 3.20	
Z (+=internal/- =external)			
Angle at Footstrike (°)	2.16 ± 9.33	2.00 ± 9.69	
Angle at Toe-off (°)	-12.40 ± 8.54	-8.98 ± 10.16	
Range of Motion (°)	13.33 ± 8.51	10.21 ± 9.42	
Relative Range of Motion (°)	10.41 ± 5.48	12.36 ± 5.77	
Peak external Rotation (°)	11.21 ± 6.27	14.81 ± 6.16	

In the sagittal plane the results indicate that the males exhibited significantly $t_{(22)} = 3.22$, $p \leq 0.01$ more hip flexion at initial contact than the female group. Furthermore, it was also found that peak hip flexion was significantly $t_{(22)} = 3.64$, $p \leq 0.01$ greater in the male group. Finally, the results indicate that the hip was significantly $t_{(22)} = 2.21$, $p \leq 0.05$ more flexed at toe-off in the male group. In the coronal plane a significant difference $t_{(22)} = 2.09$, $p \leq 0.05$ between genders was found at toe-off. The male group was found to exhibit abduction whilst the female group exhibited adduction.

Table 6.3: Knee kinematics (mean and standard deviation) as a function of gender (* = Significant main effect).

	Male	Female	
Knee			
X (+=flexion/-=extension)			
Angle at Footstrike (°)	19.84 ± 8.60	19.20 ± 10.50	
Angle at Toe-off (°)	17.43 ± 8.83	16.20 ± 8.34	
Range of Motion (°)	8.99 ± 4.58	7.95 ± 5.16	
Relative Range of Motion (°)	23.96 ± 9.77	21.95 ± 4.83	
Peak Flexion (°)	43.80 ± 7.18	41.15 ± 7.51	
Y (+=adduction - =abduction)			
Angle at Footstrike (°)	2.35 ± 3.60	1.64 ± 5.53	
Angle at Toe-off (°)	0.42 ± 3.87	-3.77 ± 4.63	*
Range of Motion (°)	3.19 ± 2.87	6.42 ± 5.27	
Relative Range of Motion (°)	3.73 ± 3.58	6.99 ± 5.18	
Peak Angle (°)	6.08 ± 5.91	-5.35 ± 4.68	*
Z (+=internal/- =external)			
Angle at Footstrike (°)	-13.53 ± 9.03	-4.89 ± 6.62	*
Angle at Toe-off (°)	-9.06 ± 8.73	-0.05 ± 7.34	*
Range of Motion (°)	6.39 ± 5.82	7.29 ± 4.88	
Relative Range of Motion (°)	15.70 ± 5.47	15.83 ± 4.43	
Peak Internal Rotation (°)	2.17 ± 7.59	10.94 ± 5.04	*

In the coronal plane a significant difference $t_{(22)} = 5.25$, $p \leq 0.01$ between genders was observed for the magnitude of peak coronal plane knee rotation. The male group exhibited adduction whilst the female group were found to exhibit abduction. Furthermore, a significant $t_{(22)} = 2.41$, $p \leq 0.05$ difference between males and females was observed at toe-off, once again females were found to exhibit abduction whilst males exhibited adduction. In the transverse plane male runners were found to be associated with significantly $t_{(22)} = 2.67$, $p \leq 0.05$ more external rotation at footstrike. Furthermore, females were found to be associated with significantly $t_{(22)} = 3.33$, $p \leq 0.01$ greater peak internal rotation magnitude whilst it was also observed that male runners exhibited significantly $t_{(22)} = 2.74$, $p \leq 0.05$ more external rotation at toe-off.

Table 6.4: Ankle kinematics (mean and standard deviation) as a function of gender (* = Significant main effect).

	Male	Female	
Ankle			
X (+ =plantar/- =dorsi)			
Angle at Footstrike (°)	-70. 35 ± 11.34	-71.77 ± 9.27	
Angle at Toe-off (°)	-43.12 ± 8.21	-48.07 ± 8.75	
Range of Motion (°)	28.51 ± 11.48	23.70 ± 10.80	
Relative Range of Motion (°)	15.97 ± 9.74	15.94 ± 7.24	
Peak Dorsi-Flexion (°)	-86.27 ± 5.75	-87.26 ± 7.18	
Y (+=inversion/ - =eversion)			
Angle at Footstrike (°)	-1.97 ± 5.25	-5.39 ± 7.34	
Angle at Toe-off (°)	3.15 ± 3.26	-1.94 ± 6.74	*
Range of Motion (°)	6.28 ± 3.17	5.17 ± 3.30	
Relative Range of Motion (°)	11.82 ± 3.15	12.12 ± 3.88	
Peak Eversion (°)	-11.97 ± 4.23	-17.51 ± 7.57	*
Z (- =internal/+ =external)			
Angle at Footstrike (°)	-14.27 ± 5.93	-18.18 ± 3.82	
Angle at Toe-off (°)	-10.00 ± 3.36	-17.68 ± 4.70	*
Range of Motion (°)	5.31 ± 2.84	4.55 ± 3.09	
Relative Range of Motion (°)	10.47 ± 3.10	10.66 ± 3.87	
Peak Angle (°)	-4.00 ± 4.52	-17.51 ± 7.66	

In the coronal plane female runners were found to be associated with a significantly $t_{(22)} = 2.21$, $p \leq 0.05$ greater magnitude of peak eversion. In addition a significant difference $t_{(22)} = 2.36$, $p \leq 0.05$ between genders was observed at toe-off, with male runners exhibiting inversion and female runners exhibiting eversion. Furthermore, in the transverse plane a significant difference t (22) =4.60, p≤0.01 between genders was observed at toe-off, with females exhibiting more external rotation.

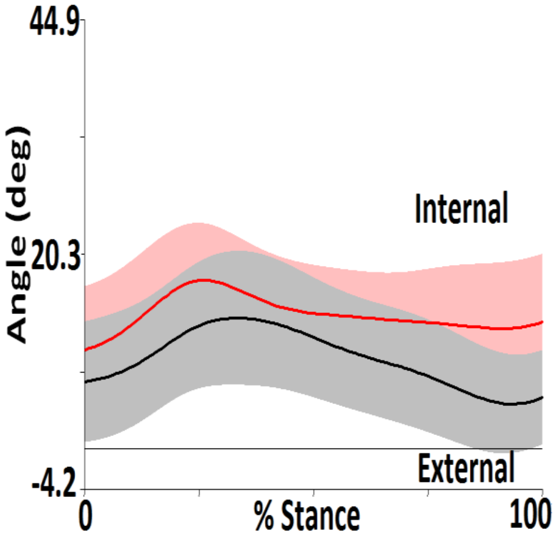

Figure 6.2: Mean and standard deviation tibial internal rotation kinematics for males (black line) and females (red line), running (shaded area is 1 ±SD, males=grey shade and females= red).

Table 6.5: Tibial internal rotation kinematics (mean and standard deviation) as a function of gender (* = Significant main effect).

	Male	**Female**	
Tibial Internal Rotation			
Z (+ =internal/ - =external)			
Angle at Footstrike (°)	7.09 ± 6.09	11.21 ± 6.70	
Angle at Toe-off (°)	5.55 ± 3.11	14.08 ± 6.85	*
Range of Motion (°)	1.65 ± 4.90	2.85 ± 2.48	
Relative Range of Motion (°)	7.93 ± 4.29	8.92 ± 3.00	
Peak Tibial Internal Rotation (°)	14.25 ± 7.36	20.17 ± 7.23	*
EV/TIR ratio	1.55 ± 1.76	1.32 ± 1.09	

The results indicate that female runners were associated with significantly greater tibial internal rotation at toe-off $t_{(22)} = 3.92$, $p \leq 0.01$. Furthermore, female were also found to be associated with significantly greater peak tibial internal rotation $t_{(22)} = 2.11$, $p \leq 0.05$.

6.1.4 Discussion

The purpose of this investigation was to determine whether female runners have different biomechanical characteristics than male runners and to use this information to provide recommendations for appropriate footwear design. This study represents the first to examine the potential necessity of different footwear designs for females.

Few investigations have been devoted to the differences in impact kinetics between male and females during running. The results of this study identified no significant kinetic differences in impact parameters between genders. The results of the current investigation appear to support the findings of both Decker et al. (2003) and Ryu, (2005) who reported no gender differences in either time or frequency domain impact parameters. However, they appear to oppose the findings of Hennig, (2005) and Stefanyshyn et al. (2003) who found that at matched velocities females were associated with significantly greater loading rates than males, although neither of these investigations examined gender differences in the frequency domain. Thus, it is concluded that gender differences in lower extremity running injuries do not appear to be related to variations in impact parameters. Therefore, with regards to the selection of appropriate footwear designs, it appears based on the findings of the current investigation with respect to shock attenuation; females do not require different footwear properties than males. This opposes the conclusions of Stefanyshyn et al. (2003) who suggested that females require footwear with additional shock attenuating properties. In addition Stefanyshyn et al. (2003) documented that in subjective ratings of heel cushioning females indicated that they would prefer more cushioning in the heel region. As such it may be that females perceive the cushioning properties of footwear differently which serves to influence their selection of running footwear. This is something that is difficult to quantify accurately due to its subjectivity, but nonetheless should be investigated further.

It is also emerging within biomechanical literature that females are at considerably greater risk of developing stress fractures, having up to four times the frequency when compared to age matched males (Pester and Smith, 1992; Queen et al. 2007). A relationship between increased vertical impact loading and the incidence of stress fractures (particularly at the tibia) has emerged within the epidemiological literature. In a number of retrospective studies, runners with a history of stress fractures have exhibited a higher tibial shock and vertical ground reaction force parameters than healthy controls (Grimston et al. 1991; Hreljac et al. 2000; Ferber et al. 2002). The results of the current investigation appear to provide only partial support for this conjecture; although a number of impact parameters were found to be higher in the female runners none were sufficiently greater to reach statistical significance. Bone exhibits both cellular and molecular remodelling responses to the mechanical stresses experienced during gait. This remodelling occurs throughout life and is affected by multiple factors. Therefore, it appears that the aetiology of stress fractures is complex and extends beyond increases in impact loading. Therefore, other factors such as bone structure, thigh and calf musculature, fitness level, body fat and hormonal variations, may also be significant (Hoch et al. 2005). Whilst these factors are beyond the scope of this investigation, future investigations examining how these factors influence the aetiology of stress fractures may assist in developing strategies to reduce the occurrence of such injuries.

Significant differences in 3-D kinematic parameters were observed between genders. With respect to sagittal plane motion of the hip, males were found to be associated with increased hip flexion throughout the stance phase. This evidence opposes the findings of Ferber et al. (2003), Schache et al. (2003) and Chumanov et al. (2008) who observed no gender differences in sagittal plane hip motion. It is difficult to elucidate to mechanisms behind this difference, however the experimental conditions in the aforementioned investigations differed from the current study. Schache et al. (2003) used a treadmill protocol in order to investigate gender differences in 3-D kinematics. Treadmill locomotion has been associated with different movement strategies in comparison to overground which may serve to diminish the differences between genders as it is not yet known to what extent male and female runners accommodate to treadmill running. Furthermore, none of the above investigations controlled for footwear amongst participants. This could potentially account for some of the differences between studies as footwear has been shown to have a significant influence of the kinematics of running (Hardin et al. 2004). It is further hypothesized that this finding relates to the greater absolute stride lengths commonly associated with male runners (Atwater, 1990). Hoffman, (1972) found moderate to strong correlations between absolute stride length and height in runners. Therefore, given that the male group were almost 10cm taller in the current investigation and as such would be expected to be associated with an increased stride length it is likely that increases in hip flexion associated with male runners are necessary to facilitate the increase in stride length.

With respect to anatomical variations between genders, this study refutes the notion that females have a wider pelvis and hips than males. The results indicate that pelvic width as measured from right ASIS to left ASIS was actually larger in the male group. This concurs with the observations of Horton and Hall, (1989) who found no significant differences between either hip or pelvic dimensions between genders. In addition no significant differences were observed between genders with respect to femoral length although femur length was greater in the male group. However, in support of Horton and Hall, (1989) larger (although non-significant) pelvic and hip to femoral length ratios were observed in the female group. Theoretically increases in pelvic/hip: femur length ratio contributes to a greater static femoral valgus in females and thus increases the Q angle. As such although no direct measures of Q-angle were made it appears that, the increases in femoral valgus indicate a trend toward increases in Q angle in the female group.

With respect to the knee joint complex, no significant differences were observed in the sagittal plane. Previous investigations have reported conflicting results with respect to sagittal plane knee kinematics; Maliznak et al. (2001) found that females exhibit less peak knee flexion and less knee flexion excursion in comparison to males, whilst Ferber et al. (2003) reported no gender differences in sagittal plane knee kinematics. Hewett et al. (2005) propose that females limit the amount of knee flexion during dynamic tasks, and instead, rely more on their passive restraints in the frontal plane to control these tasks. The results of the current investigation provide partial support for this notion in that females were associated with non-significant reductions in knee flexion and significant increases in frontal plane knee abduction. It has been hypothesized that females lack the strength and/or neuromuscular control of the sagittal plane musculature to effectively decelerate the body COM during landing and thus rely on frontal plane mechanics to a greater extent than males. Hewett et al.

(2005) found that both knee valgus motion and moments to be predictors of ACL injury. In general, the knee joint mechanics exhibited by females are thought to place them at a greater risk of ACL injury. Therefore, it appears that females are at greater risk from non-contact ACL injuries. The results of this study provide basis for future work examining the underlying mechanisms behind this movement strategy and geometric differences in the size and shape of the ACL and their influence on the aetiology of non-contact ACL injuries.

In the coronal plane females were found to be associated with significantly greater peak knee abduction and knee abduction at toe-off. This concurs with the findings of Cho et al. (2004), Ferber et al. (2003) and Hurd et al. (2004) who reported that females were associated with significant increases in knee abduction in comparison to male runners. The greater knee abduction in conjunction with increases (non-significant) in hip adduction associated with female runners may facilitate an increase in dynamic Q-angle. This supports the current conjecture with respect to gender differences in Q-angle (Aglietti et al. 1983; Horton and Hall, 1989; Hsu et al. 1990). Increases in dynamic Q-angle magnitude may enhance the lateral pull of the quadriceps on the patella (Horton and Hall, 1989), which serves to facilitate misalignment of the patellofemoral joint and produces compression of the lateral articular surface and is hypothesized to be associated with greater lateral patellar contact forces and may facilitate a greater incidence of patellofemoral disorders (Mizuno et al. 2001). As such, the results of the current investigation appear to at least partially explain the mechanisms behind the increases in susceptibility of female runners to patellofemoral disorders and pain (Almeida et al. 1999; DeHaven and Lintner, 1986).

With respect to the ankle joint complex, significant increases in ankle eversion and associated tibial internal rotation parameters were reported for the female group. These results concur with the findings of Hennig, (2001) and Kernozek et al. (2005) who also observed increases in ankle eversion in female runners. The significant increases in peak eversion and tibial internal rotation in female runners also has potential clinical significance. These findings suggest that female runners may be associated with an increased risk from stability related injury as excessive rearfoot eversion and associated tibial internal rotation are implicated in the aetiology of a number of overuse injuries such as Achilles tendinitis, patellar tendonitis, iliotibial band syndrome, patellofemoral pain and plantar fasciitis (Willems et al. 2004; Lee et al. 2010; Taunton et al. 2003; Duffey et al. 2000).

Significantly, iliotibial band pathology is considered to be the leading cause of lateral knee pain in runners (Taunton et al. 2003). Female runners are reported to be twice as likely to suffer from illiotibial band syndrome as males (Taunton et al. 2003). The increase in coronal plane eversion in the female condition serves to augment tension in the illiotibial band which is hypothesized by Noehren et al. (2006) as a being the mechanism by which illiotibial pathology occurs. As such this finding appears to explain the increased susceptibility of females to illiotibial band injury. Therefore, given the significant increase in rearfoot eversion observed in female runners it is recommended that females select running footwear with design characteristics aimed towards the reduction of calcaneal eversion. It is hypothesized based on the findings of the current investigations that this will serve to reduce the incidence of pathology in female runners.

With respect to the potential differences in coupling between ankle and tibia it was observed that females showed a trend towards having a lower ankle eversion to tibial internal rotation ratio in comparison to males. This suggests that differences between genders may exist in terms of the distal coupling mechanism between ankle and tibia. The rearfoot eversion/tibial internal rotation ratio (EV/TIR) is an important mechanism as it provides insight into where an injury is most likely to occur (Nigg et al. 1993). It is hypothesized that a greater EV/TIR ratio i.e. relatively greater rearfoot eversion in relation to tibial internal rotation may increase the stress placed on the foot and ankle (Nawoczenski et al. 1997; Nigg et al. 1993) and are thus at greater risk for foot injuries. Conversely, those with lower EV/TIR ratios (relatively more tibial motion in relation to rearfoot eversion) are at greater risk from knee related injuries (McClay and Manal, 1997; Nawoczenski et al. 1997; Williams et al. 2001). As such it appears that females are susceptible to knee injuries and males may be most susceptible to foot injuries.

In conclusion this study provides data not previously available comparing the impact kinetics and lower extremity 3-D kinematics of male and female runners. The current investigation provides insight into the aetiology of different injury patterns that may be observed between genders. Furthermore, this study supports the notion that females are more susceptible to overuse injuries than males, although further studies are required in order to determine whether gender differences in lower extremity kinematics are related to the incidence of injury. With regards to appropriate footwear, it appears based on the findings of the current investigation with respect to shock attenuation; females do not require different footwear properties than males. However, it is recommended that females select running footwear with design characteristics aimed towards the reduction of coronal plane ankle eversion in order to reduce the incidence of injury. Future research should focus on prospective studies whereby aetiological measures are determined before individuals obtain the injury and as such causative factors may be more accurately determined allowing footwear designs to be developed and prescribed more effectively.

7. Differences in the kinetics and kinematics of barefoot and barefoot inspired footwear in comparison to conventional footwear

Publications

1. Sinclair J, Greenhalgh, A, Edmundson CJ, and Hobbs SJ (2012). The influence of barefoot and barefoot inspired footwear on the kinetics and kinematics of running in comparison to conventional running shoes, Footwear Science, 5, 1-10, I First.
2. Sinclair J, Taylor PJ, Edmundson CJ, and Hobbs SJ (2012). Differences in tibiocalcaneal kinematics measured with skin and shoe mounted markers. Journal of Human Movement (In press).

Conference presentations

3. Sinclair, J (2011). Does footwear influence the way we run. Workshop, University of Essex June 2011.
4. Sinclair, J and Hobbs, S.J (2012). Is barefoot running beneficial. Science and Technology conference, University of Central Lancashire, July 2012.
5. Sinclair, J (2012). Barefoot and Barefoot inspired footwear in running biomechanics. Conferencia be biomechanica do Porto, University of Porto, Portugal, (Keynote address).

7.1 Pilot study 1 – Kinematic differences between shoe an skin mounted foot markers

7.1.1 Introduction

During running excessive motions of the ankle and tibia, have been implicated in the aetiology of a number of overuse injuries (Viitasalo and Kvist, 1983; van Mechelen, 1992; Taunton et al. 2003). Therefore, numerous investigations have been undertaken examining the 3-D kinematics of the foot with respect to the tibia (Clement et al. 1981). These studies are conducted to determine how different running shoe properties influence these parameters (Stacoff et al. 1991; Stacoff et al. 2000), quantify the coupling mechanism between eversion and tibial internal rotation and to investigate the potential relationship between kinematic parameters and running injuries (Nigg and Morlock, 1987; Hamill et al. 1992).

To quantify these movements, retro-reflective markers are typically attached through external palpation to the shoe. However, during dynamic movements such as running, the foot may move inside the shoe and thus these external markers may not accurately represent the movement of the foot itself (Stacoff et al. 1992). Therefore measurement errors, typically referred to as movement artefact, may be introduced as a function of this relative movement. Several techniques have been developed in order to overcome issues regarding the placement of markers on the shoe. Pins attached directly to bone via intercortical screws can be used to accurately quantify skeletal motion (Reinschmidt et al. 1997). However, the application of this technique is limited due to its invasiveness. Therefore, the currently accepted gold standard technique that does not require invasive techniques is to place markers onto the foot itself through windows in the shoe (Richards et al. 2008).

Previous investigations have been conducted which have examined the kinematic differences between externally mounted markers and those placed inside windows cut into the shoe. However these studies have examined limited discrete 3-D kinematic parameters and have not taken into account how the different techniqes influence the kinematic waveforms.

Given that this chapter focuses on barefoot (whereby markers are placed on the skin) and shod running (whereby markers are placed on the shoe) it is neccesary to determine the extent to which the two differ: in order to examine the efficacy of findings obtained from the main study. Therefore the aim of this pilot investigation was to compare the 3-D tibiocalcaneal kinematics between skin and shoe mounted markers using both kinematic waveform (intra-class correlations) and discrete variable (paired t-tests) analyses.

7.1.2 Methods

7.1.2 (i) Participants

Ten male participants (age = 23.4 ± 4.30 years; height = 1.79 ± 0.08 m; body mass = 71.7 ± 9.26 kg) were recruited for this investigation. All were injury free and provided written informed consent. Ethical approval for this study was granted from a University School of Psychology ethical panel.

7.1.2 (ii) Procedure

Kinematic parameters were obtained at 250 Hz via an eight camera motion analysis system (Qualisys™ Medical AB, Goteburg, Sweden) whilst participants ran at $4.0 m.s^{-1} \pm 5\%$.

Running velocity was monitored using infrared light-cells Newtest 300 (Newtest, Oulu Finland). Participants struck a force platform (Kistler, Kistler Instruments Ltd., Alton, Hampshire, UK; Model 9281CA) sampling at 1000 Hz, with their dominant limb in order to define the stance phase of running. Stance time was determined as the time over which 20 N or greater of vertical force was applied to the force platform.

In order to define the anatomical and technical reference frames of the foot and shank a static trial was captured allowing the anatomical frame to be referenced in relation to the technical frame. Following this, markers that were not used for tracking the segments during motion, were removed. Windows with length: width dimensions in accordance with the Schultz and Jenkyn (2012) guidelines were cut in the laboratory footwear (Saucony pro grid guide II) at the approximate positions of the 1st metatarsal, 5th metatarsal and calcaneus. To define the foot and tibial segment anatomical frame axes retro-reflective markers were attached to the right foot and shank in the following locations, medial and lateral malleoli, medial and lateral epicondyles of the femur (Figure 7.1). The foot segment was tracked 1. using markers positioned close to the 1st and 5th metatarsal heads and the calcaneus (Shoe) and 2. using markers positioned onto the skin within the shoe windows (Skin). The tibia was tracked via a cluster comprised of four 19mm spherical reflective markers mounted to a thin sheath of lightweight carbon fibre with a length to width ratio of 1.5-1, in accordance with the Cappozzo et al. (1997) recommendations.

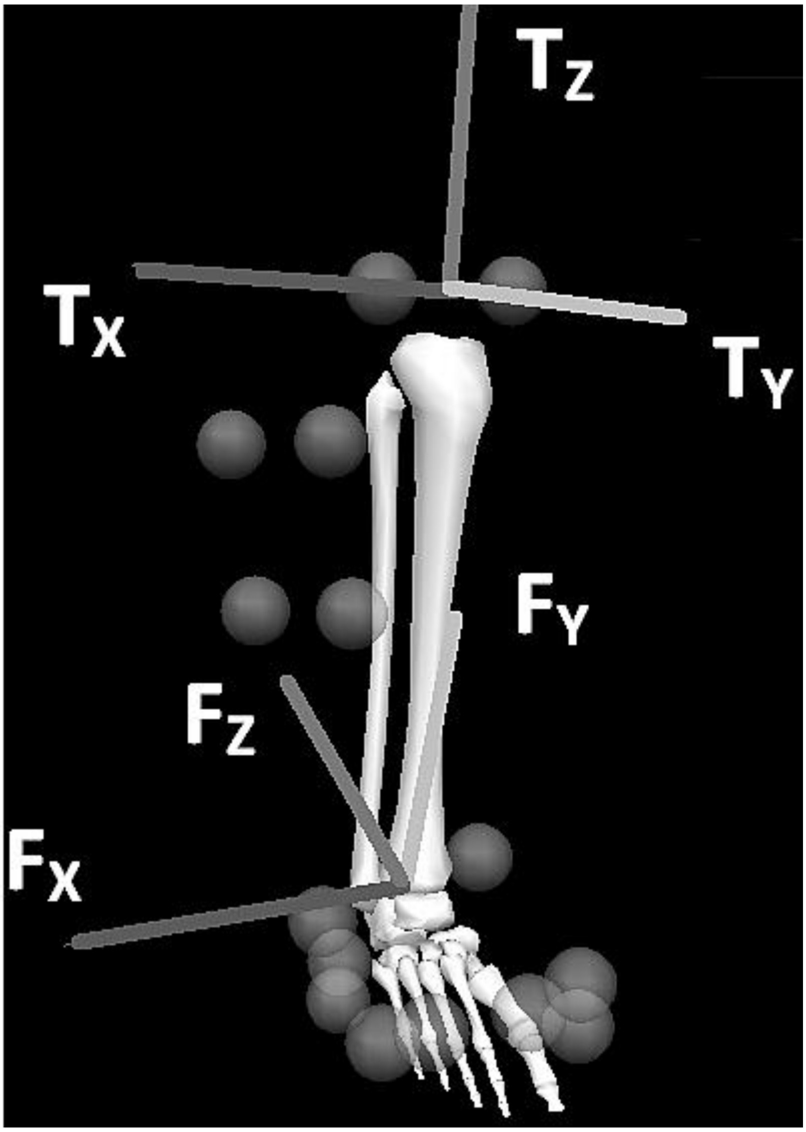

Figure 7.1: Tibial and foot segments, with reference segment co-ordinate system axes (T = tibia and F = foot).

7.1.2 (iii) Data Processing

The running trials were digitized using Qualisys Track Manager and then exported as C3-D files. Kinematic parameters were quantified using Visual 3-D (C-Motion Inc, Germantown, USA) after marker data was smoothed using a low-pass Butterworth 4th order zero-lag filter at a cut off frequency of 10 Hz. This was quantified as being the frequency at which 95% of the signal power was maintained following a FFT. 3-D kinematic parameters were calculated using an XYZ cardan sequence of rotations. Trails were normalized to 100% of the stance phase then processed gait trials were averaged. 3-D kinematic measures from the hip, knee and ankle which were extracted for statistical analysis were 1) angle at footstrike, 2) angle at toe-off, 3) ROM during stance, 4) peak angle during stance and 5) relative ROM from footstrike to peak angle, 6) velocity at footstrike, 7) velocity at toe-off, 8) peak velocity and 9) eversion/tibial internal rotation (EV/TIR) ratio. In addition, to assess the proprioceptive differences imposed by the modified footwear, participants were asked to subjectively rate (in relation to the left shoe which remained unmodified) on a scale of 1-10 with 1 being totally uncomfortable and 10 being totally comfortable.

7.1.2 (iv) Statistical Analyses

Descriptive statistics (mean ± standard deviation) were calculated for the outcome measures. To compare differences in stance phase 3-D tibiocalcaneal kinematic parameters between skin and shoe mounted markers paired t-tests were utilized with statistical significance accepted at the $p \leq 0.05$ level. Intra-class correlations were also utilized to compare skin and shoe sagittal, coronal and transverse plane waveforms. The Shapiro-Wilk statistic for each condition confirmed that the data were normally distributed. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA).

<u>7.1.3 Results</u>

Tables 7.1-7.3 and figures 7.2 and 7.3 present kinematic differences observed using skin and shoe mounted markers.

7.1.3 (i) Joint Angles

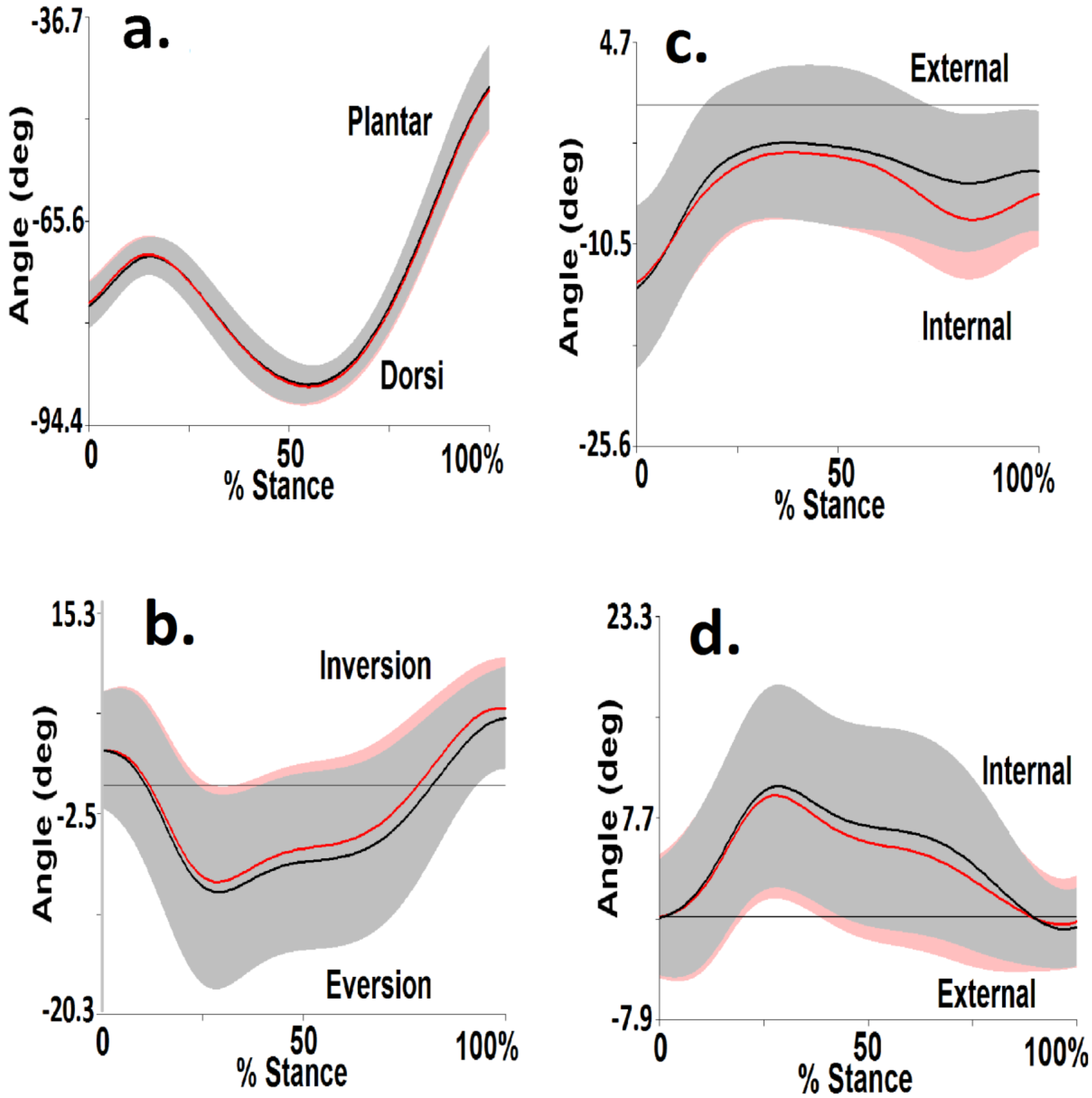

Figure 7.2: Mean and standard deviation kinematic parameters representing a. sagittal, b. coronal, c. transverse and d. tibial internal rotation movements for shoe (black line) and skin (red line), mounted markers (shaded area is 1 ± SD, shoe =grey shade and skin = pink).

Table 7.1: Ankle joint kinematics (mean ± standard deviation) in the sagittal, coronal and transverse planes as a function of the different foot tracking techniques (* = Significant main effect).

	Shoe	**Skin**	
Ankle			
X (+ =plantar/ - =dorsi)			
Angle at Footstrike (°)	-77.01 ± 2.73	-77.51 ± 2.92	
Angle at Toe-off (°)	-47.07 ± 5.48	-46.67 ± 5.38	
Range of Motion (°)	29.94 ± 3.88	30.84 ± 3.68	
Relative Range of Motion (°)	11.99 ± 2.46	11.24 ± 3.04	
Peak Dorsi-Flexion (°)	-89.01 ± 2.25	-88.75 ± 2.36	
Y (+ =inversion/ =eversion)			
Angle at Footstrike (°)	2.33 ± 5.01	2.31 ± 4.92	
Angle at Toe-off (°)	6.15 ± 4.15	5.30 ± 4.15	
Range of Motion (°)	4.19 ± 2.33	3.36 ± 2.13	
Relative Range of Motion (°)	12.61 ± 3.70	13.46 ± 4.14	*
Peak Eversion (°)	-10.28 ± 8.18	-11.15 ± 8.39	*
Z (- =internal/ + =external)			
Angle at Footstrike (°)	-13.58 ± 5.55	-13.88 ± 6.56	
Angle at Toe-off (°)	-6.82 ± 4.08	-4.99 ± 4.65	
Range of Motion (°)	6.76 ± 3.10	8.89 ± 3.39	*
Relative Range of Motion (°)	10.45 ± 1.19	12.12 ± 2.27	*
Peak Angle (°)	-3.13 ± 4.74	-1.76 ± 5.22	*

Table 7.2: Tibial internal rotation parameters (mean ± standard deviation) in the sagittal, coronal and transverse planes as a function of the different foot tracking techniques (* = Significant main effect).

	Shoe	**Skin**	
Tibial Internal Rotation			
Z (+ =internal/ - =external)			
Angle at Footstrike (°)	0.84 ± 4.96	0.72 ± 4.55	
Angle at Toe-off (°)	0.13 ± 3.68	-0.40 ± 3.24	
Range of Motion (°)	0.67 ± 1.39	1.22 ± 2.09	
Relative Range of Motion (°)	10.34 ± 3.74	11.13 ± 4.09	*
Peak Tibial Internal Rotation (°)	11.17 ± 7.91	11.85 ± 7.89	
EV/TIR ratio	1.27 ± 0.16	1.26 ± 0.20	

In the coronal plane the skin mounted markers produced a significantly greater relative ROM $t_{(9)}$ =3.16, p≤0.01 and peak eversion magnitude $t_{(9)}$ = 2.30, p≤0.05. In the transverse plane the skin mounted markers once again produced a significantly greater ROM t (9) = 7.06, p≤0.05, relative ROM $t_{(9)}$ = 3.27, p≤0.01 and peak angle $t_{(9)}$ = 2.46, p≤0.05. It was further observed that the skin mounted markers produced a significantly greater relative ROM for tibial internal rotation $t_{(9)}$ =3.32, p≤0.01. Comparisons between shoe and skin mounted angular kinematic waveforms for the ankle joint revealed strong correlations for the sagittal (R^2= 0.99), coronal (R^2=0.92) and transverse (R^2= 0.97) planes. Comparisons between tibial internal rotation waveforms also revealed strong correlations (R^2= 0.85).

7.1.3 (ii) Joint Velocities

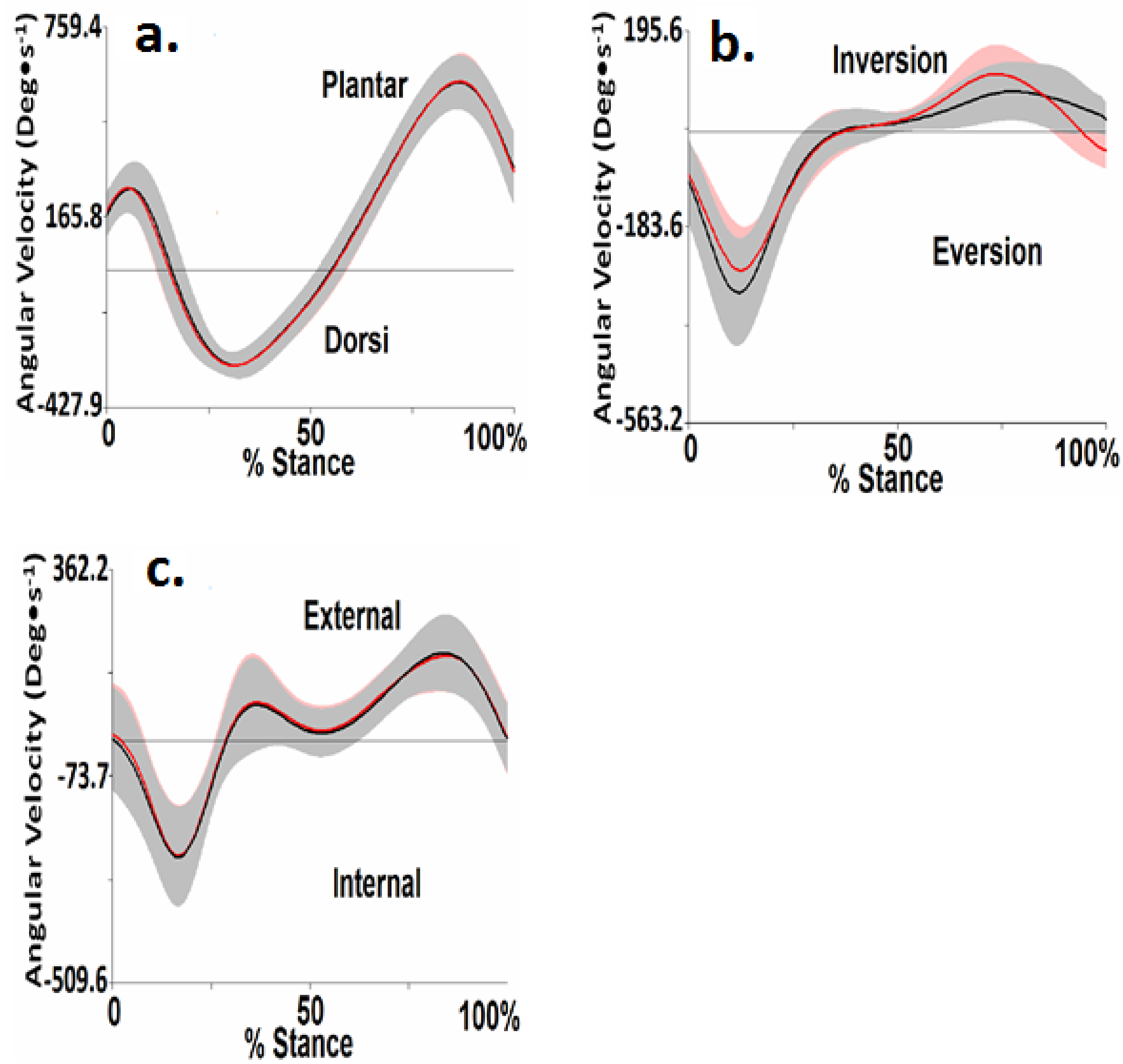

Figure 7.3: Mean and standard deviation velocities representing a. sagittal, b. coronal, and c. transverse of shoe (red line) and skin (black line), mounted markers (shaded area is 1 ± SD, shoe =red shade and skin = black).

Table 7.3: Ankle joint velocities (mean ± standard deviation) in the sagittal, coronal and transverse planes as a function of the different foot tracking techniques (* = Significant main effect).

	Shoe	Skin	
Ankle			
X (+ =plantar/ - =dorsi)			
Velocity at FootStrike (°.s^{-1})	189.55 ± 57.38	181.50 ± 76.27	
Velocity at Toe-Off (°.s^{-1})	313.12 ± 76.62	322.03 ± 85.37	
Peak Plantarflexion Velocity (°.s^{-1})	603.52 ± 79.74	604.03 ± 89.51	
Peak Dorsiflexion Velocity (°.s^{-1})	-314.22 ± 28.08	-318.47 ± 44.22	
Y (+ =inversion/ - =eversion)			
Velocity at FootStrike (°.s^{-1})	-90.35 ± 63.05	-107.11 ± 83.10	
Velocity at Toe-Off (°.s^{-1})	-36.34 ± 12.70	-26.04 ± 25.50	
Peak Inversion Velocity (°.s^{-1})	130.61 ± 51.54	107.90 ± 45.72	
Peak Eversion Velocity (°.s^{-1})	-294.56 ± 65.96	-338.37 ± 85.50	*
Z (- =internal/ + =external)			
Velocity at FootStrike (°.s^{-1})	-5.12 ± 114.74	-15.41 ± 117.05	*
Velocity at Toe-Off (°.s^{-1})	6.89 ± 83.15	-0.78 ± 55.39	
Peak Internal Rotation Velocity (°.s^{-1})	219.32 ± 28.08	219.48 ± 83.58	
Peak External Rotation Velocity (°.s^{-1})	-247.78 ± 99.13	-255.01 ± 99.52	

In the coronal plane peak eversion velocity was found to significantly $t_{(9)} = 5.11$, $p \leq 0.01$greater using shoe mounted markers. Furthermore in the transverse plane skin mounted markers were associated with a significantly $t_{(9)} = 2.56$, $p \leq 0.05$ greater internal rotation velocity at footstrike. Comparisons between shoe and skin mounted angular kinematic waveforms for the ankle joint revealed strong correlations for the sagittal (R^2= 0.99), coronal (R^2=0.96) and transverse (R^2= 0.96) planes.

7.1.3 (iii) Ratings of shoe comfort

Participants rated the modified footwear at a comfort level of 6.5 ± 1.18.

7.1.4 Discussion

The aim of this pilot investigation was to determine the kinematic differences between skin and shoe mounted markers. This pilot study represents the first to statistically examine the differences in stance phase waveforms and discrete kinematic parameters using skin and shoe mounted markers.

The results indicate that the different foot tracking mechanisms have no significant influence on sagittal plane kinematic parameters. This is further substantiated by the intra-class correlation analyses which show very high agreement $R^2 \geq 0.99$ between shoe and skin mounted waveforms. This concurs with the findings of Reinschmidt et al. (1997) who also found that sagittal plane kinematics were minimally affected by different methods of tracking the foot segment, although they did observe an increase in peak dorsi-flexion when quantifying kinematics using shoe mounted markers.

However, when quantifying tibiocalcaneal motions in the coronal and transverse planes significant differences between the discrete kinematic parameters were observed. It was observed that placing markers on the running shoe lead to a significant underestimation of coronal and transverse plane rotations. This opposes the findings of Reinschmidt et al. (1997)

who found the shoe mounted foot tracking techniques served to overestimate the motions of the ankle and tibia in the coronal and transverse planes. This may potentially be attributable to the fact that Reinschmidt et al. (1997) removed the heel cap from their experimental footwear thus greatly increasing the potential for relative shoe to foot movement.

Whilst the findings of the current investigation disagree with the observations of Reinschmidt et al. (1997), the implications of the current study are similar in that shoe mounted markers are not representative of true foot movement measured with markers placed directly onto the skin. The findings of the current investigation have potential clinical significance as lower extremity movements of excessive ankle eversion and tibial internal rotation are implicated in the aetiology of a number of lower extremity pathologies. Therefore, any mis-representation of these parameters may serve to confound the efficacy of epidemiological analyses.

Clinical gait analyses such as the current investigation have typically considered the foot as a single rigid segment (Richards et al. 2008). However this technique may not allow 3-D kinematics to be collected for the joints within the foot which are also susceptible to injury and dysfunction (Hunter and Prentice, 2001). Therefore, whilst this study provides important information regarding the differences between skin and shoe mounted markers for a single segment foot model, future work should be conducted examining the differences between the two tracking mechanisms when using a multiple segment foot model.

In conclusion although previous studies have compared shoe to skin mounted markers, the current knowledge is limited in terms of the parameters that have been observed. Given that significant differences were observed between skin and shoe mounted markers in key coronal and transverse plane parameters it is concluded that the results of studies using shoe mounted markers should be interpreted with caution, particularly when performing clinical analyses. However, given that cutting holes in the experimental footwear has been proposed to reduce the structural integrity of the upper, and that participants indicated that the modifications affected their subjective feel of the shoe it was determined that the most appropriate technique for the current investigation was to place markers onto the shoe surface. To modify the experimental footwear in order to place markers onto the skin would have more of a negative impact on ecological validity than would using shoe mounted markers as proprioception can have a large influence on the kinetics and kinematics of running.

7.2 Main study

7.2.1 Introduction

In recent years the concept of barefoot running has been the subject of much attention in footwear biomechanics literature. Furthermore, a number of well-known athletes have competed barefoot, most notably Zola Budd-Pieterse and the Abebe Bikila who both held world records for the 5000m and marathon events respectively. This demonstrates that barefoot running does not appear to prevent athletes from competing at the highest levels (Warburton, 2000). Barefoot locomotion presents a paradox in footwear literature (Robbins and Hanna, 1987); and has been used for many years both by coaches and athletes (Nigg 2009) based around the supposition that running shoes are associated with an increased incidence of running injuries (Lieberman et al. 2010; Robbins and Hanna, 1987; Warburton, 2000).

Based on such research and taking into account the barefoot movement's recent rise in popularity, shoes have been designed in an attempt to transfer the perceived advantages of barefoot movement into a shod condition (Nigg, 2009). Yet, given the popularity of barefoot running, surprisingly few investigations have specifically examined both the impact kinetics and 3-D kinematics of the lower extremities of running barefoot and in barefoot inspired footwear in comparison to shod. Furthermore, there is a paucity of research reporting the prospective epidemiological investigations into the aetiology of injury in runners and how footwear may affect the frequency of injury. This study provides a comparison of the kinetics and 3-D kinematics of running: barefoot, in conventional running shoes and in barefoot inspired footwear, in order to highlight the differences among conditions.

The aim of the current investigation was therefore to determine 1: whether differences in impact kinetics during running exist between the footwear conditions and 2: whether shoes which aim to simulate barefoot movement patterns can closely mimic the 3-D kinematics of barefoot running.

<u>7.2.2 Methods</u>

7.2.2 (i) Participants

Twelve male runners, volunteered to take part in this study. Participants were active runners engaging in training at least three times per week; completing a minimum of 35 km. All were injury free at the time of data collection and provided written informed consent. The mean characteristics of the participants were; age 24.34 ± 1.10 years, height 1.78 ± 0.05 m and body mass 76.79 ± 8.96 kg. A statistical power analysis was conducted using the Hopkins method in order to reduce the likelihood of a type II error and determine the minimum number participants needed for this investigation. It was found that the sample size was sufficient to provide more than 80% statistical power. The procedure utilized for this investigation was approved by the University of Central Lancashire, School of Psychology, ethical committee.

7.2.2 (ii) Procedure

Participants ran at 4.0 $m.s^{-1}$ over a force plate (Kistler, Kistler Instruments Ltd., Alton, Hampshire) embedded in the floor (Altrosports 6mm, Altro Ltd,) of a 22 m biomechanics laboratory. Running velocity was quantified using Newtest 300 infrared timing gates (Newtest, Oy Koulukatu, Finland); a maximum deviation of ±5% from the set velocity was allowed. Stance time was defined as the time over which 20 N or greater of vertical force was applied to the force platform. A successful trial was defined as one within the specified velocity range, where all tracking clusters were in view of the cameras, the foot made full contact with the force plate and no evidence of gait modifications due to the experimental conditions. Runners completed a minimum of five successful trials in each footwear condition. Participants were non-habitual barefoot runners and were thus given time to accommodate to the barefoot and barefoot inspired footwear prior to the commencement of data collection. This involved 5 minutes of running through the testing area without concern for striking the force platform. Dynamic calibration with the same acceptance criteria as outlined in chapter 5 was conducted prior to data collection.

7.2.2 (iii) 3-D Kinematics

3-D kinematics from the lower extremities were obtained using the same protocol as the in chapter 5. Static trials were conducted for each footwear condition.

7.2.2 (iv) Tibial accelerations

Tibial accelerations were also obtained using the protocol outlined in chapter 5.

7.2.2 (v) Data Processing

Trials were processed in Qualisys Track Manager in order to identify anatomical and tracking markers then exported as C3D files. Kinematic parameters were quantified using Visual 3-D (C-Motion Inc, Germantown, USA) after marker data were smoothed using a low-pass

Butterworth 4th order zero-lag filter at a cut off frequency of 10Hz. This frequency was selected as being the frequency at which 95% of the signal power was below. 3-D kinematics of the hip knee and ankle joints were calculated using an XYZ cardan sequence of rotations. All data were normalized to 100% of the stance phase then processed gait trials were averaged. 3-D kinematic measures from the hip, knee and ankle which were extracted for statistical analysis were 1) angle at footstrike, 2) angle at toe-off, 3) ROM during stance, 4) peak angle during stance and 5) relative ROM from footstrike to peak angle. In addition to this the eversion/tibial internal tibial internal rotation (EV/TIR) ratio was quantified in accordance with De Leo et al. (2004) as the relative eversion ROM / the relative tibial internal rotation ROM.

The acceleration signal was filtered using a 60 Hz Butterworth zero-lag 4th order low pass filter in accordance with the Lafortune and Hennig, (1992) recommendations to prevent any resonance effects on the acceleration signal. Peak positive axial tibial acceleration was defined as the highest positive acceleration peak measured during the stance phase. To analyze data in the frequency domain, a FFT function was performed and median power frequency content of the acceleration signals were calculated in accordance with (Lafortune and Hennig, 1995).

Forces were reported in bodyweights (BWs) to allow normalisation of the data among participants. From the force plate data, stance time, average loading rate, instantaneous loading rate, peak impact force and time to peak impact were calculated. Average loading rate was calculated by dividing the impact peak magnitude by the time to the impact peak. Instantaneous loading rate was quantified as the maximum increase in vertical force between frequency intervals.

7.2.2 (vi) Shoes

The shoes utilized during this study consisted of a Saucony Pro Grid Guide II and a Nike Free 3.0. The shoes were the same for all runners; they differed in size only (sizes 6, 7 and 9 in men's shoe UK sizes).

7.2.2 (vii) Statistical Analysis

Descriptive statistics including means and standard deviations of 3-D kinematic, impact shock and impact force parameters were calculated for each footwear condition. Differences between the parameters were examined using repeated measures ANOVA's with significance accepted at the $p \leq 0.05$ level. Appropriate post-hoc analyses were conducted using a Bonferroni correction to control for type I error. Effect sizes were calculated using a η^2. If the sphericity assumption was violated then the degrees of freedom were adjusted using the Greenhouse Geisser correction. The Shapiro-Wilk statistic for each footwear condition confirmed that all data were normally distributed. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA).

7.2.3 Results

Tables 7.4-7-8 and figures 7-4-7.5 present the kinetic and 3-D kinematic information obtained as a function of footwear.

7.2.3 (i) Kinetic and temporal parameters

Table 7.4: Kinetic and temporal variables (mean and standard deviation) as a function of footwear (* = Significant main effect).

	Barefoot	**Barefoot Inspired Footwear**	**Conventional**	
Vertical Impact Peak (BW)	1.94 ± 0.92	1.95 ± 0.37	1.76 ± 0.48	
Instantaneous Loading Rate ($BW.s^{-1}$)	422.38 ± 226.36	207.97 ± 51.17 ¥	139.50 ± 62.02 ¥†	*
Average Loading Rate ($BW.s^{-1}$)	182.08 ± 128.61	100.61 ± 33.89	67.43 ± 17.93 ¥†	*
Time to Peak Impact (ms)	17.07 ± 4.02	22.32 ± 5.12	30.44 ± 4.09 ¥†	*
Peak impact shock (g)	9.17 ± 2.96	10.2 ± 3.48	6.60 ± 3.65 ¥†	*
Median Power Frequency (Hz)	15.62 ± 3.61	14.38 ± 4.72	12.01 ± 3.53 ¥†	*
Stance time (ms)	193.33 ± 26.01	196.78± 31.93	207.98 ± 30.10 ¥†	*
Peak Braking Force (BW)	0.42 ± 0.16	0.45 ± 0.09	0.49 ± 0.10	
Peak Propulsive Force (BW)	0.42 ± 0.09	0.38 ± 0.07	0.39 ± 0.07	
Peak Medial Force (BW)	0.19 ± 0.03	0.21 ± 0.08	0.20 ± 0.07	
Peak Lateral Force (BW)	0.11 ± 0.06	0.06 ± 0.10	0.10 ± 0.07	

¥ Significantly different from the barefoot condition

† Significantly different from the barefoot inspired condition

The results indicate that a significant main effect was observed for the instantaneous loading rate $F_{(1.08, 11.88)}$, = 20.05, $p \leq 0.01$, η^2=0.65. Post-hoc analyses revealed that the instantaneous loading rate was significantly higher in the barefoot condition in comparison to the barefoot inspired footwear (p=0.011) and conventional shoe (p=0.001) conditions). Furthermore the post-hoc analysis also showed that barefoot inspired footwear were associated with a significantly (p=0.001) higher instantaneous loading rate than the conventional shoe condition. In addition a significant main effect was also observed for the average loading rate $F_{(1.08, 11.84)} = 9.19$, $p \leq 0.01$, $\eta^2 = 0.46$. Post-hoc analyses revealed that the average loading rate was significantly lower in the conventional shoe condition in comparison to the barefoot inspired footwear (p=0.004) and barefoot conditions (p=0.02) which did not differ significantly (p=0.084) from one another. A significant main effect was observed for the time to impact peak $F_{(1.23, 13.58)} = 7.94$, $p \leq 0.01$, $\eta^2 = 0.41$. Post-hoc analyses revealed that the time to impact peak was significantly greater in the conventional shoe condition in comparison to the barefoot inspired footwear (p=0.006) and barefoot (p=0.042) conditions which did not differ significantly (p=0.504) from one another. Finally, a significant main effect $F_{(1.21, 13.35)} = 15.81$, $p \leq 0.01$, $\eta^2=0.59$ was found for the magnitude of peak axial impact shock. Post-hoc analysis revealed that peak impact shock was significantly greater in the barefoot (p=0.021) and barefoot inspired footwear (p=0.01) conditions in comparison to the conventional shoe condition. The spectral analysis of the acceleration signal revealed that a significant main effect $F_{(1.29, 14.14)}$ 14.09, $p \leq 0.01$, $\eta^2=0.56$ existed for the median frequency content. Post-hoc analysis revealed that the conventional shoe condition was associated with a significantly lower frequency content than the barefoot (p=0.001) and barefoot inspired footwear (p=0.0001). No significant differences were observed between the barefoot and barefoot inspired footwear (p=0.35). Finally, a significant main effect $F_{(2, 22)} = 8.10$, $p \leq 0.01$, $\eta^2=0.42$ was found for the stance time duration. Post-hoc analysis revealed that stance times were significantly shorter in the barefoot (p=0.003) and the barefoot inspired footwear (p=0.008) conditions in comparison to the conventional shoe condition. No significant differences (p=0.512) were found between the barefoot and barefoot inspired footwear.

7.2.3 (ii) 3-D Kinematic parameters

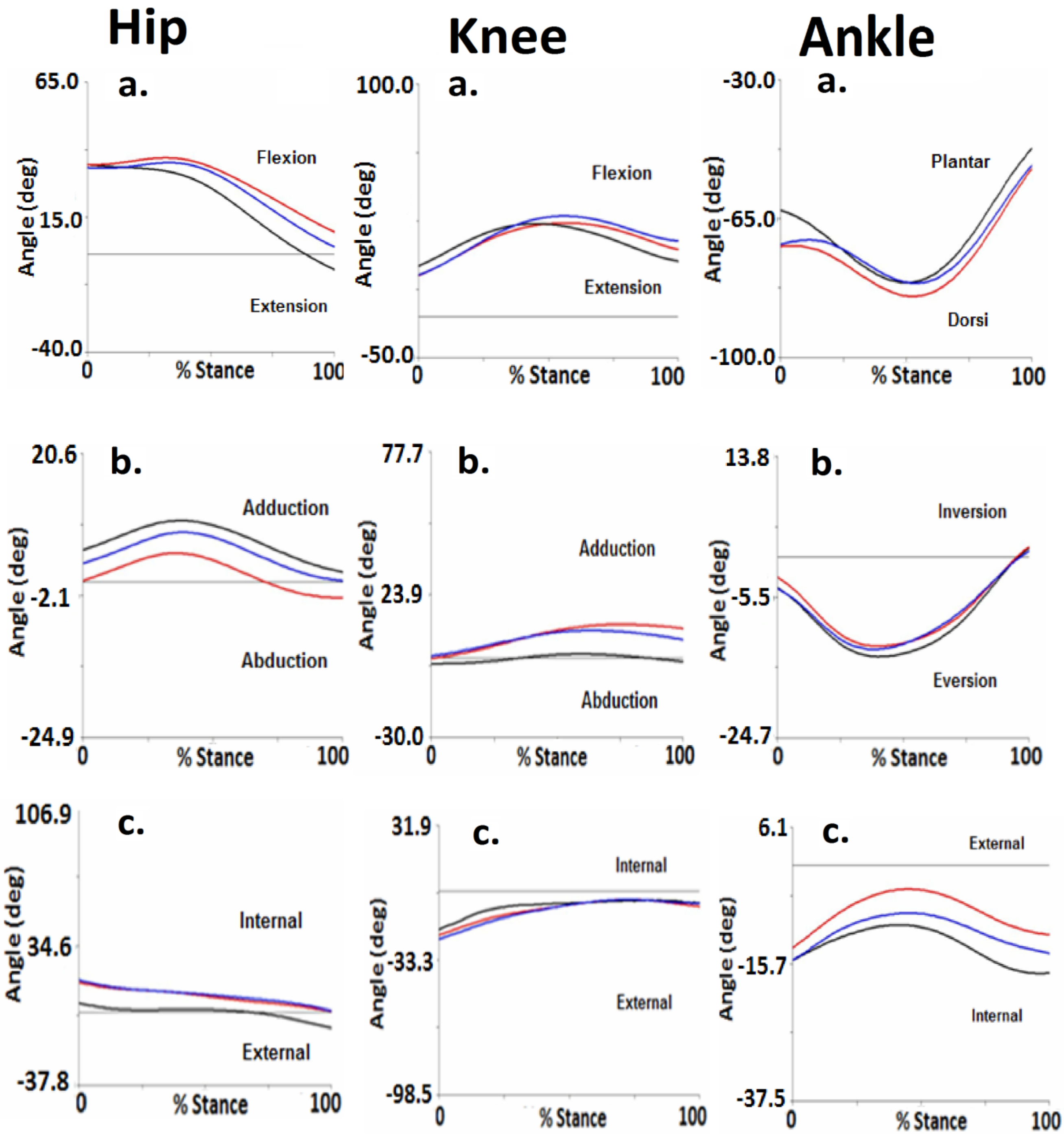

Figure 7.4: Mean hip knee and ankle kinematics as a function of footwear in the a. sagittal, b. coronal and c. transverse planes (black=barefoot, red=barefoot inspired footwear and blue =Saucony.)

Table 7.5: Hip joint kinematic (mean and standard deviation) as a function of footwear (* = Significant main effect).

	Barefoot	Barefoot Inspired Footwear	Conventional	
Hip				
X (+ = flexion/ - = extension)				
Angle at Footstrike (°)	44.39 ± 17.06	43.75 ± 18.44	45.02 ± 19.23	
Angle at Toe-off (°)	-5.44 ± 12.16	3.78 ± 22.08	1.49 ± 19.24	
Range of Motion (°)	50.14 ± 15.27	40.27 ± 8.29	44.02 ± 6.57	
Relative Range of Motion (°)	2.31 ± 6.88	5.63 ± 20.86	5.83 ± 17.88	
Peak Flexion (°)	46.00 ± 12.44	48.39 ± 9.14	49.85 ± 8.58 ¥	*
Y (+=adduction/-=abduction)				
Angle at Footstrike (°)	4.94 ± 3.96	0.62 ± 5.40	3.19 ± 5.57	
Angle at Toe-off (°)	1.05 ± 5.50	-1.34 ± 6.80	-0.04 ± 5.60	
Range of Motion (°)	4.74 ± 4.51	4.46 ± 2.88	3.86 ± 3.09	
Relative Range of Motion (°)	5.11 ± 2.20	5.14 ± 2.36	5.03 ± 3.04	
Peak Adduction (°)	10.04 ± 4.45	5.76 ± 7.05	8.22 ± 5.32	
Z (+=internal /-=external)				
Angle at Footstrike (°)	5.11 ± 11.99	15.36 ± 31.56	13.28 ± 22.74	
Angle at Toe-off (°)	-7.60 ± 12.61	-0.39 ± 29.86	-1.24 ± 21.17	
Range of Motion (°)	12.72 ± 5.62	15.75 ± 6.30	14.52 ± 6.16	
Relative Range of Motion (°)	1.09 ± 3.33	2.40 ± 5.90	1.79 ± 7.02	
Peak Internal rotation (°)	6.12 ± 4.99	17.59 ± 13.54	15.09 ± 14.21	

¥ Significantly different from the barefoot condition

† Significantly different from the barefoot inspired condition

A significant main effect $F_{(1.25,\ 13.73)} = 5.24$, $p \leq 0.05$, $\eta^2 = 0.32$ was found for peak flexion. Post-hoc analysis revealed that peak flexion was significantly p=0.039 greater in the conventional shoe condition, in comparison to the barefoot condition.

Table 7.6: Knee joint kinematic (mean and standard deviation) as a function of footwear (* = Significant main effect)

	Barefoot	Barefoot Inspired Footwear	Conventional	
Knee				
X (+ = flexion/ - = extension)				
Angle at Footstrike (°)	22.07 ± 4.73	17.74 ± 9.43	18.47 ± 9.43	
Angle at Toe-off (°)	24.89 ± 10.42	28.91 ± 25.92	29.54 ± 26.65	
Range of Motion (°)	8.86 ± 8.49	19.55 ± 26.57	20.17 ± 27.57	
Relative Range of Motion (°)	22.59 ± 9.31	31.82 ± 24.18	32.39 ± 26.06	
Peak Flexion (°)	44.66 ± 7.84	49.56 ± 20.17	50.86 ± 19.77	
Y (+=adduction/- =abduction)				
Angle at Footstrike (°)	-1.91 ± 5.38	-1.33 ± 7.84	0.76 ± 3.96	
Angle at Toe-off (°)	-1.12 ± 9.67	9.93 ± 22.93	4.93 ± 15.47	
Range of Motion (°)	1.82 ± 7.92	15.45 ± 24.17	9.11 ± 14.39	
Relative Range of Motion (°)	6.88 ± 8.50	16.84 ± 26.63	10.61 ± 17.07	
Peak Adduction (°)	4.98 ± 10.50	15.51 ± 22.93	11.37 ± 16.57	
Z (+=internal/- =external)				
Angle at Footstrike (°)	-16.66 ± 9.14	-21.98 ± 26.71	-20.51 ± 19.19	
Angle at Toe-off (°)	-4.23 ± 8.70	-6.54 ± 19.46	-6.49 ± 10.61	
Range of Motion (°)	12.43 ± 11.59	15.71 ± 21.49	14.74 ± 19.74	
Relative Range of Motion (°)	16.94 ± 10.25	22.02 ± 18.48	20.93 ± 17.54	
Peak Internal Rotation (°)	0.27 ± 7.98	0.05 ± 17.47	0.43 ± 7.96	

¥ Significantly different from the barefoot condition

† Significantly different from the barefoot inspired condition

No significant (p≤0.05) differences were in knee joint kinematics were found among footwear conditions.

Table 7.7: Ankle joint kinematics (mean and standard deviation) as a function of footwear (* = Significant main effect).

	Barefoot	Barefoot Inspired Footwear	Conventional	
Ankle				
X (+ =plantar/- =dorsi)				
Angle at Footstrike (°)	-64.08 ± 8.66	-72.12 ± 9.18 ¥	-73.38 ± 7.52 ¥	*
Angle at Toe-off (°)	-45.49 ± 7.47	49.12 ± 8.75	-48.97 ± 9.12	
Range of Motion (°)	18.58 ± 7.95	23.00 ± 10.19	24.40 ± 9.33	
Relative Range of Motion (°)	20.82 ± 7.67	12.89 ± 4.86 ¥	14.68 ± 4.28 ¥	*
Peak Dorsiflexion (°)	-84.89 ± 5.34	-85.01 ± 8.17	-88.06 ± 5.97	
Y (+=inversion/ -=eversion)				
Angle at Footstrike (°)	-3.79 ± 3.80	-2.55 ± 4.21	-3.99 ± 3.30	
Angle at Toe-off (°)	1.39 ± 4.34	2.39 ± 5.83	0.83 ± 2.94	
Range of Motion (°)	5.21 ± 4.74	4.94 ± 2.98	4.83 ± 3.58	
Relative Range of Motion (°)	10.55 ± 3.94	10.37 ± 3.60	9.45 ± 3.17	
Peak Eversion (°)	-14.34 ± 3.12	-12.92 ± 5.44	-13.44 ± 4.57	
Z (+=internal/- =external)				
Angle at Footstrike (°)	-14.84 ± 4.92	-12.66 ± 3.65	-14..64 ± 4.20	
Angle at Toe-off (°)	-16.71 ± 4.57	-11.23 ± 3.97 ¥	-13.41 ± 4.68	*
Range of Motion (°)	3.07 ± 2.65	1.86 ± 2.38	2.86 ± 2.22	
Relative Range of Motion (°)	6.16 ± 2.44	9.55 ± 2.20	5.00 ± 7.23	
Minimum External Rotation (°)	-8.67 ± 4.81	-3.11 ± 4.25 ¥	-6.60 ± 4.26	*

¥ Significantly different from the barefoot condition

† Significantly different from the barefoot inspired condition

A significant main effect $F_{(2, 22)} = 7.91$, $p \leq 0.01$, $\eta^2=0.42$ was observed for the magnitude of plantarflexion at foot strike. Post-hoc analysis revealed that in the barefoot condition the ankle was significantly more plantarflexed than in both the conventional (p=0.01) and the shoes designed to simulate barefoot running (p=0.015). A significant main effect $F_{(1.06, 11.66)} =8.23$, $p \leq 0.01$, $\eta^2=0.43$ existed for the movement from footstrike to peak dorsiflexion in terms of ROM. Post-hoc analyses revealed that the motion was significantly greater in the barefoot condition in comparison to the footwear designed to simulate barefoot running (p=0.011) and conventional shoe (p=0.013) conditions.

The results indicate that a significant main effect $F_{(2, 22)} = 7.23$, $p \leq 0.01$, $\eta^2 = 0.40$ exists for the magnitude of peak axial rotation. Post-hoc analysis revealed that the barefoot condition was significantly p=0.001 more externally rotated in comparison to the shoes designed to simulate barefoot running. The results indicate that a significant main effect $F_{(2, 22)} = 6.09$, $p \leq 0.01$, $\eta^2=0.36$ exists for the magnitude of axial rotation at toe-off. Post-hoc analysis revealed that external rotation was significantly (p=0.001) greater in the barefoot condition in comparison to the barefoot inspired footwear.

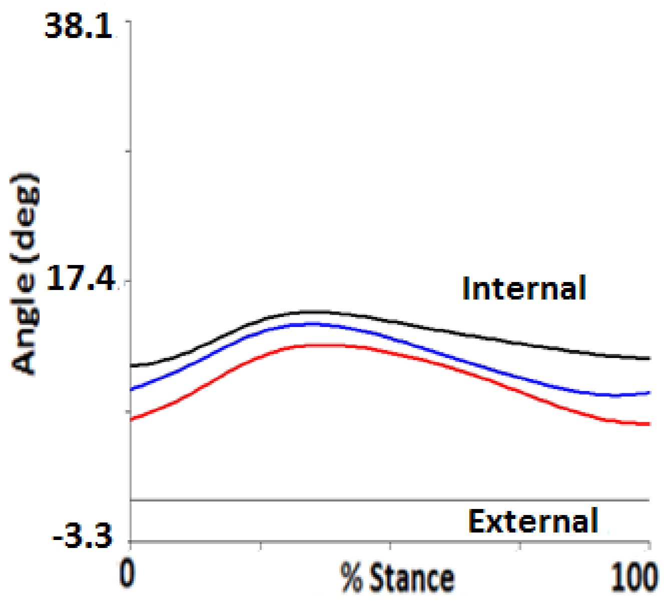

Figure 7.5: Mean tibial internal rotation kinematics of the stance limb as a function of footwear (black=barefoot, red=barefoot inspired footwear and blue =Saucony.)

Table 7.8: Tibial internal rotation kinematics (mean and standard deviation) as a function of footwear (* = Significant main effect).

	Barefoot	**Barefoot Inspired Footwear**	**Conventional**	
Tibial Internal Rotation				
Z (+ =internal/ - =external)				
Angle at Footstrike (°)	10.17 ± 3.92	6.41 ± 3.92	8.57 ± 4.35	
Angle at Toe-off (°)	10.90 ± 5.63	6.32 ± 4.05	8.27 ± 4.61	
Range of Motion (°)	0.68 ± 3.40	0.19 ± 0.93	0.32 ± 1.99	
Relative Range of Motion (°)	6.69 ± 4.04	6.93 ± 3.20	6.59 ± 3.12	
Peak Tibial Internal Rotation (°)	16.31 ± 4.07	13.05 ± 5.76	15.15 ± 5.61	
EV/TIR ratio	1.62 ± 1.06	1.49 ± 0.63	1.40 ± 0.98	

7.2.4 Discussion

This study represents is the first to synchronously examine the alterations in 3-D kinematics, force and axial impact shock associated with running barefoot, in conventional footwear and in footwear designed to simulate barefoot running.

The results from the kinetic analysis indicate that the conventional shoes were associated with lower impact parameters than running barefoot. This finding corresponds with the results of previous investigations (Dickinson et al. 1985; De Koning and Nigg 1993; De Clercq et al. 1994; De Wit et al. 2000) who reported significantly greater impact parameters when running barefoot. This however opposes the findings of Squadrone and Gallozzi (2009) and Lieberman et al. (2010) who observed that those running barefoot were associated with smaller collision forces than shod. Moreover, that instantaneous loading rate was found to be significantly greater in the barefoot condition in comparison to the barefoot inspired shoes opposes the findings of Squadrone and Gallozzi (2009) who reported that impact forces did not differ significantly between barefoot and barefoot inspired footwear.

These observations may relate to the differences in barefoot running experience between studies. Squadrone and Gallozzi (2009) and Lieberman et al. (2010) utilized habitual barefoot runners which is in contrast to the non-habitual barefoot runners examined in the current investigation. Therefore the kinetic observations in barefoot analyses may relate to the experience of the participants in barefoot locomotion, this is an interesting notion and future research may wish to replicate the current investigation using habitually barefoot runners. Furthermore, both Squadrone and Gallozzi, (2009) and Lieberman et al. (2010) utilized a treadmill protocol which is in contrast to the current investigation whereby participants conducted overground trials. This may also have influenced the differences between studies as the treadmill may allow a larger number footfalls to be captured and ensures that continuous movement kinematics are obtained (Fellin et al. 2010a).

The results also indicate that stance times were significantly shorter whilst running barefoot and in barefoot inspired footwear in comparison to the conventional running shoe condition. This also corresponds with previous investigations with respect to shorter stance times being associated with barefoot running (De Wit et al. 2000; Warburton, 2000). Furthermore it would also appear to confirm that the barefoot condition was associated with a greater step frequency/reduced step lengths, as De Wit et al. (2000) found stance times to be strongly correlated with step length. With respect to the hip joint complex, in the sagittal plane a significant increase in peak flexion during the early stance phase was found in the conventional shoe condition in comparison to the barefoot condition. It is surmised that this finding is attributable to the mechanical alterations that runners make when running barefoot. Runners traditionally take longer steps when running in traditional footwear, so their COM moves through a greater horizontal displacement during each step. As such, during early stance the hip must flex to a greater extent in order to reduce the horizontal distance from the stance leg to the COM to maintain balance during the early stance phase.

The results indicate that the ankle was significantly more plantarflexed at initial contact in the barefoot condition in comparison to the conventional shoe and barefoot inspired footwear, suggesting a mid or forefoot strike pattern. This concurs with the findings of (De Wit et al. 2000; Hartveld and Chockalingam 2001; Griffin et al. 2007) findings. Barefoot running or running in shoes with less midsole cushioning is proposed to facilitate increases in plantar discomfort which are sensed and moderated (Robbins and Gouw, 1991). Footwear with greater cushioning i.e. the conventional and barefoot inspired footwear conditions provoke a reduction in shock-moderating behaviour as evidenced by the increased dorsiflexion angle at footstrike (Robbins and Hanna, 1987; Robbins et al. 1989; Robbins and Gouw, 1991). This may lend support to the supposition that the body adapts to a lack of cushioning via kinematic measures. However, it appears that these measures do not offer the same shock attenuating properties as do cushioned midsoles found in conventional footwear.

The increase in plantarflexion at footstrike associated with barefoot running is considered to be the primary mechanism by which runners adjust to this condition (De Wit et al. 2000; Warburton, 2000; Griffin et al. 2007). Thus, it appears that the barefoot inspired footwear do not closely mimic the kinematics of barefoot running with respect to the ankle joint complex. It is proposed that this finding is attributable to the perceptual effects of increased cushioning in the barefoot inspired footwear which were found to have increased shock attenuating properties. This finding opposes the observations of Squadrone and Gallozzi, (2009) who found that barefoot inspired footwear where effective in imitating barefoot conditions. However, Squadrone and Gallozzi, (2009) utilized the Vibram five-fingers design as their barefoot inspired footwear condition which are characterized by their minimalist features in contrast to the Nike Free footwear utilized in the current investigation which aims to simulate barefoot locomotion through a flexible outsole construction. It appears that barefoot inspired footwear between different designs and manufacturers cannot be considered analogous. Future research is therefore necessary to examine the efficacy of the various conceptual shoe models which aim to replicate barefoot locomotion.

Interestingly, no significant differences were found between the three footwear conditions, in terms of the peak eversion magnitude during stance. This is appears to oppose the findings of Warburton, (2001), Shorten (2000), Edington et al. (1990), Stacoff et al. (1991) and Smith et al. (1986) who reported that ankle eversion is greater during shod running. Greater ankle eversion is reputed to be due to a reduction in stability caused by the cushioned midsole (Shorten, 2000). However like most modern footwear, both the conventional and barefoot inspired footwear encompass features such as stiffer cushioning, stiff heel counters, insole boards, medially posted midsoles, varus wedges designed to control excessive ankle eversion (Shorten, 2000). Therefore, whilst it appears logical that cushioning will lead to increased ankle eversion the results of this investigation suggest that a combination of cushioning and features designed to control pronation can be effective.

There is a paucity of research directly comparing injury rates in shod and barefoot running. However, the findings of this study in conjunction with epidemiological analyses suggest that running in conventional footwear may lower the incidence of impact related overuse injuries as increases in impact parameters have been linked to the aetiology of a number overuse pathologies (Hardin et al. 2004; Misevich and Cavanagh 1984). Furthermore, the results of the kinetic analysis suggest that the barefoot inspired footwear offer shock attenuating properties that are superior to barefoot conditions, but inferior to the conventional running shoe. It appears based on these findings that the barefoot inspired footwear places runners at greater risk of musculoskeletal injury compared to the conventional footwear yet at a lesser risk in comparison to barefoot running at comparable velocities. However, given that previous investigations have shown habitually barefoot runners to be less susceptible to impact related overuse injuries (Warburton, 2001); it is proposed that the experience of the runners may influence their susceptibility of injury. Therefore, future work should be carried out examining participants before and after their habituation to barefoot conditions to determine whether this affects their impact kinetics.

That this investigation quantified barefoot locomotion with skin mounted markers and shod motion using shoe mounted markers may serve as a limitation of the current investigation. Pilot study 7.1 found that differences exist between shoe and skin mounted markers, thus it is questionable as to whether anatomical markers located on the shoe provide comparable results to those placed on the foot itself Stacoff et al. (1992). Thus potentially reducing the efficacy of the comparison between the shoe conditions where markers were placed onto the shoe and barefoot condition where makers were placed directly onto the foot. However, given that cutting holes in the shoes in order to attach markers to skin would likely cause further problems by compromising the structural integrity of the upper, and that participants indicated that the holes affected the feel of the shoe it was determined that the current technique was the most appropriate.

In conclusion although previous studies have compared barefoot and shod running, the current knowledge with respect to the degree in which these modalities differ is limited. The present study adds to the current knowledge of barefoot running by providing a comprehensive kinetic and 3-D kinematic evaluation. Given that significant differences were observed between running barefoot and in barefoot inspired footwear, it was determined that they do not closely mimic the mechanics of barefoot running. Future research will serve to determine the efficacy of footwear designed to mimic barefoot running. Finally, although further investigation is necessary concerning additional barefoot inspired shoe models it appears in this case that conventional shod running is superior to both barefoot running and shoes designed to mimic barefoot running, in terms of protection from running injuries. Future research should focus on providing prospective epidemiological analyses of barefoot and shod runners and the influence of different footwear conditions on the aetiology of running injuries.

8. The influence of footwear cushioning properties on steady state energy expenditure.

Publications

1. Sinclair J, Brooks, D, A, Edmundson CJ, and Hobbs SJ (2012). The efficacy of EMG MVC normalization techniques for running analyses. Journal of Biomechanics, 41, 621-623.
2. Sinclair, J., Taylor, P.J., Edmundson, C.J. Brooks, D., and Hobbs, S.J. (2012). The influence of footwear kinetic, kinematic and electromyographical parameters on the energy requirements of steady state running. Movement and Sports Science, 79, 1-14, I First.

Conference Presentations

3. Sinclair J, Edmundson, CJ, Brooks and Hobbs SJ (2012). The efficacy of EMG MVC normalization techniques for running analyses European congress of Biomechanics, Lisbon, Portugal.
4. Sinclair J, Taylor PJ & Hobbs SJ (2013*). Development of a novel technique to assess shoe centre of mass. Accepted for presentation at 11TH Footwear Biomechanics Symposium, July 31st – August 2nd 2013 in Natal, Brazil.

8.1 Pilot study 1 Optimal filtering of EMG data

8.1.1 Introduction

Although there are a number of mechanisms by which the elecctromyographic signal can be processed, the most appropriate technique has yet to be fully established (Hug and Dorel, 2008). However, it is recommended that the EMG signal be full wave rectified and then filtered in order to remove background noise and tissue artefact, creating a linear envelope (Bartlett 1997). One method of smoothing the EMG signal is to pass the data through both low and high pass digital filters (Bartlett, 1997), a second technique which is also frequently utilized within the literature involved smoothing the signal using only a digital low-pass 4th order filter at a cut-off frequency of 3-20Hz (Shiavi et al. 1998; Winter 1990; Merletti and Parker, 2004). Alternative methods for data analysis which are used less frequently include using a high pass filter with a relatively high cut-off of frequency of 25-30Hz, which may remove some, but not all, of the noise due to skin movement artefacts. A moving average (Acierno et al. 1995; Latash, 1998) or root mean square of the signal (De Luca and Knaflitz, 1990) may also be used. Integration filtering allows the signal strength to be assessed in terms of the total voltage through the muscle and can be related to muscle force (Enoka, 2002; Nigg and Herzog, 2005). Based on the recommendations of Richards et al. (2008) and Nigg and Herzog, (2005) a low pass filter will be used throughout this thesis, yet the most appropriate cut-off frequency for running analyses is not yet known. The aim of the current pilot investigation was to determine the most appropriate cut-off frequency in order to filter the EMG signals.

8.1.2 Methods

8.1.2 (i) Participants

Seven male participants (Age: 21.32 ± 2.51years, mass: 72.57 ± 8.56kg, Height 1.75 ± 0.21) took part in this investigation. All were injury free and provided written consent. Ethical approval for this project was obtained from the University of Central Lancashire School of Psychology ethics committee.

8.1.2 (ii) Procedure

Participants completed five trials at $4.0 m.s^{-1} \pm 5\%$ striking a force platform with their right foot. Muscle activity from the Medial Gastrocnemius, Vastus Lateralis, Vastus Medialis and Tibialis Anterior from the right leg was obtained at 1000Hz using the protocol outlined in detail in section 8.5. In accordance with the SENIAM guidelines the electrodes were placed on the bellies of the appropriate muscles. The skin was prepared by abrading the skin with a paper towel to remove dead skin and cleaning with an isopropyl alcohol wipe.

8.1.2 (iii) Data Processing

Trials were processed in Qualisys Track Manager then exported as C3D files. The EMG signal parameters were quantified using Visual 3-D (C-motion Inc, Germantown, USA). Following full wave rectification, the EMG signal was filtered at 4, 6, 8, 10, 15, 20, 40, 60, 80, 100, 120, 140 and 160Hz using a Butterworth low pass 4^{th} order zero-lag filter to create a linear envelope.

8.1.2 (iv) Statistical Analysis

Descriptive statistics (mean ± SD) were calculated for the outcome measures. Differences in mean and peak EMG amplitude from the stance phase were analysed using repeated measures ANOVA's with significance set at $p \leq 0.05$.

8.1.3 Results

Tables 8.1-8.4 and figures 8.1-8.4 present the mean and peak stance phase activation as a function of frequency cut-off

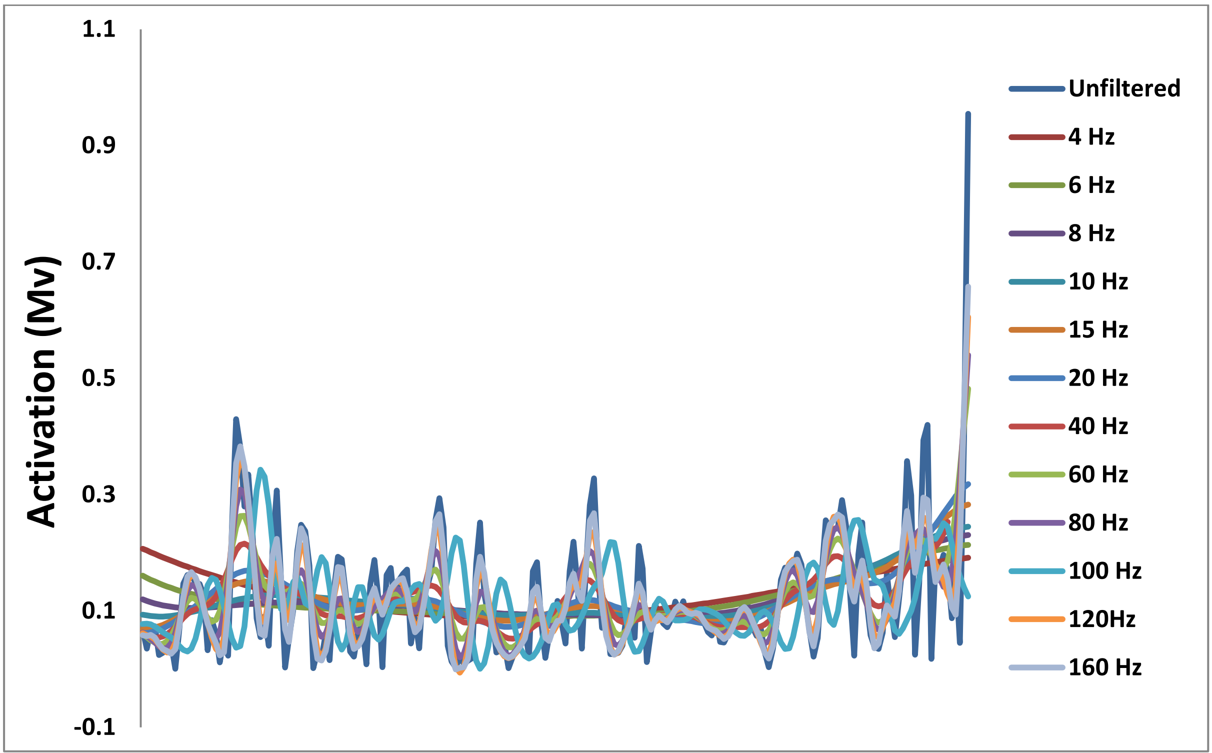

Figure 8.1: Representative stance phase EMG signals obtained from the Tibialis Anterior as a function of cut-off frequency.

Table 8.1: Muscle activation (mean and standard deviation) from the Tibialis anterior as a function of cut-off frequency (* = significant main effect).

	Tibialis Anterior (mV)	
	Peak	**Mean**
Unfiltered	0.87 ± 0.57	0.20 ± 0.21
4Hz	0.42 ± 0.40	0.22 ± 0.22
6Hz	0.42 ± 0.40	0.21 ± 0.22
8Hz	0.43 ± 0.41	0.21 ± 0.21
10Hz	0.45 ± 0.43	0.21 ± 0.21
15Hz	0.51 ± 0.48	0.20 ± 0.21
20Hz	0.56 ± 0.52	0.20 ± 0.21
40Hz	0.67 ± 0.60	0.20 ± 0.21
60Hz	0.74 ± 0.65	0.20 ± 0.21
80Hz	0.77 ± 0.66	0.20 ± 0.21
100Hz	0.81 ± 0.67	0.20 ± 0.21
120Hz	0.83 ± 0.67	0.20 ± 0.21
140Hz	0.85 ± 0.67	0.20 ± 0.21
160Hz	0.86 ± 0.66	0.20 ± 0.21
	*	

The results indicate that for mean stance phase activation that no significant $F_{(1.10,\ 6.63)} = 4.25$, $p > 0.05$, $\eta^2=0.41$ differences exist between the different cut-off frequencies. However a significant main effect $F_{(13,\ 78)} = 5.52$, $p \leq 0.05$, $\eta^2=0.48$ was observed for the magnitude of peak stance phase activation.

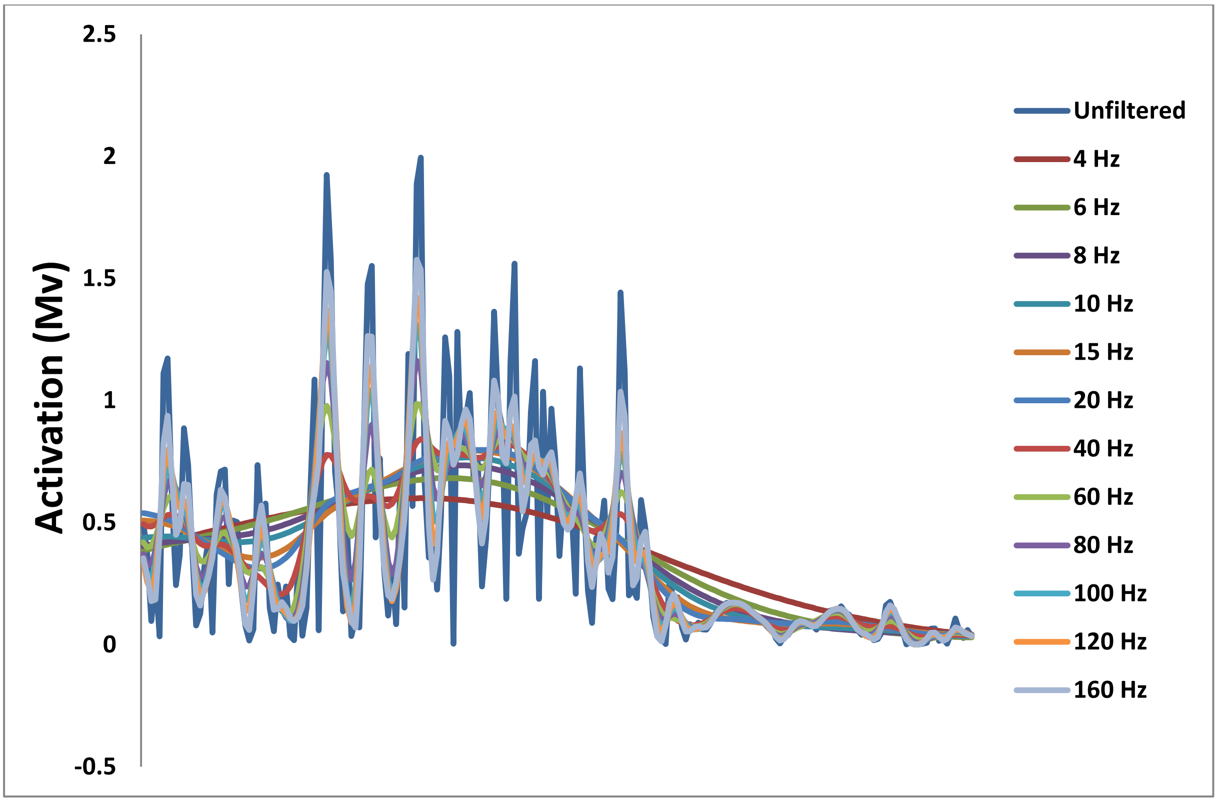

Figure 8.2: Representative stance phase EMG signals obtained from the Medial Gastrocnemius as a function of cut-off frequency.

Table 8.2: Muscle activation (mean and standard deviation) from the Medial Gastrocnemius as a function of cut-off frequency (* = significant main effect).

	Medial Gastrocnemius (mV)	
	Peak	**Mean**
Unfiltered	1.14 ± 0.42	0.28 ± 0.11
4Hz	0.42 ± 0.18	0.29 ± 0.11
6Hz	0.47 ± 0.20	0.29 ± 0.11
8Hz	0.50 ± 0.21	0.29 ± 0.11
10Hz	0.52 ± 0.22	0.29 ± 0.11
15Hz	0.57 ± 0.23	0.29 ± 0.11
20Hz	0.61 ± 0.24	0.29 ± 0.11
40Hz	0.73 ± 0.27	0.29 ± 0.11
60Hz	0.83 ± 0.30	0.29 ± 0.11
80Hz	0.91 ± 0.33	0.29 ± 0.11
100Hz	0.98 ± 0.35	0.29 ± 0.11
120Hz	1.04 ± 0.38	0.29 ± 0.11
140Hz	1.07 ± 0.39	0.29 ± 0.11
160Hz	1.09 ± 0.41	0.29 ± 0.11
	*	

The results indicate that for mean stance phase activation that no significant $F_{(13, 78)} = 1.46$, $p > 0.05$, $\eta^2=0.20$ differences exist between the different cut-off frequencies. However a significant main effect $F_{(13, 78)} = 64.29$, $p \leq 0.01$, $\eta^2=0.92$ was observed for the magnitude of peak stance phase activation.

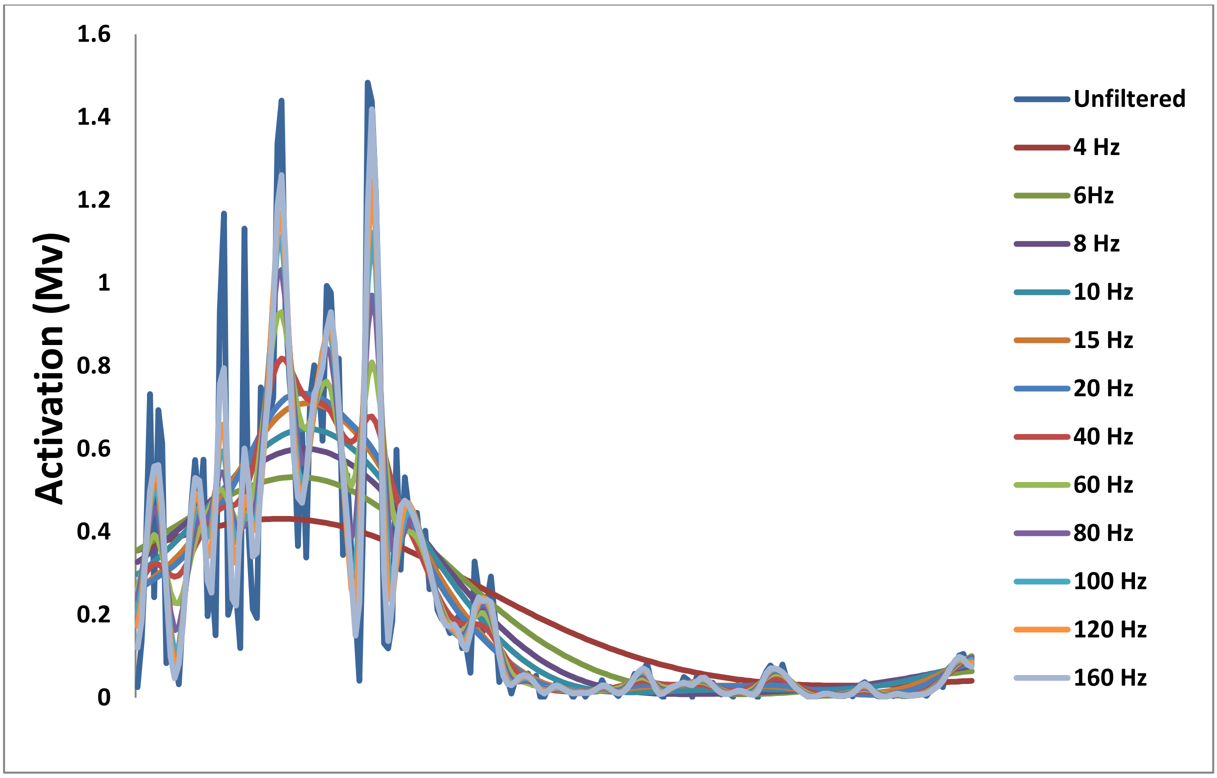

Figure 8.3: Representative stance phase EMG signals obtained from the Vastus Lateralis as a function of cut-off frequency.

Table 8.3: Muscle activation (mean and standard deviation) from the Vastus Lateralis as a function of cut-off frequency (* = significant main effect).

	Vastus Lateralis (mV)	
	Peak	**Mean**
Unfiltered	1.29 ± 0.46	0.26 ± 0.18
4Hz	0.43 ± 0.24	0.21 ± 0.10
6Hz	0.44 ± 0.20	0.21 ± 0.10
8Hz	0.48 ± 0.22	0.21 ± 0.10
10Hz	0.51 ± 0.24	0.21 ± 0.10
15Hz	0.56 ± 0.26	0.21 ± 0.10
20Hz	0.58 ± 0.26	0.21 ± 0.10
40Hz	0.72 ± 0.31	0.21 ± 0.10
60Hz	0.82 ± 0.35	0.21 ± 0.10
80Hz	0.91 ± 0.39	0.21 ± 0.10
100Hz	0.99 ± 0.42	0.21 ± 0.10
120Hz	1.05 ± 0.43	0.21 ± 0.10
140Hz	1.09 ± 0.46	0.21 ± 0.099
160Hz	1.14 ± 0.47	0.21 ± 0.099
	*	

The results indicate that for mean stance phase activation that no significant $F_{(13, 78)} = 0.75$, $p>0.05$, $\eta^2=0.11$ differences exist between the different cut-off frequencies. However a significant main effect $F_{(13, 78)} = 38.86$, $p\leq0.01$, $\eta^2=0.87$ was observed for the magnitude of peak stance phase activation.

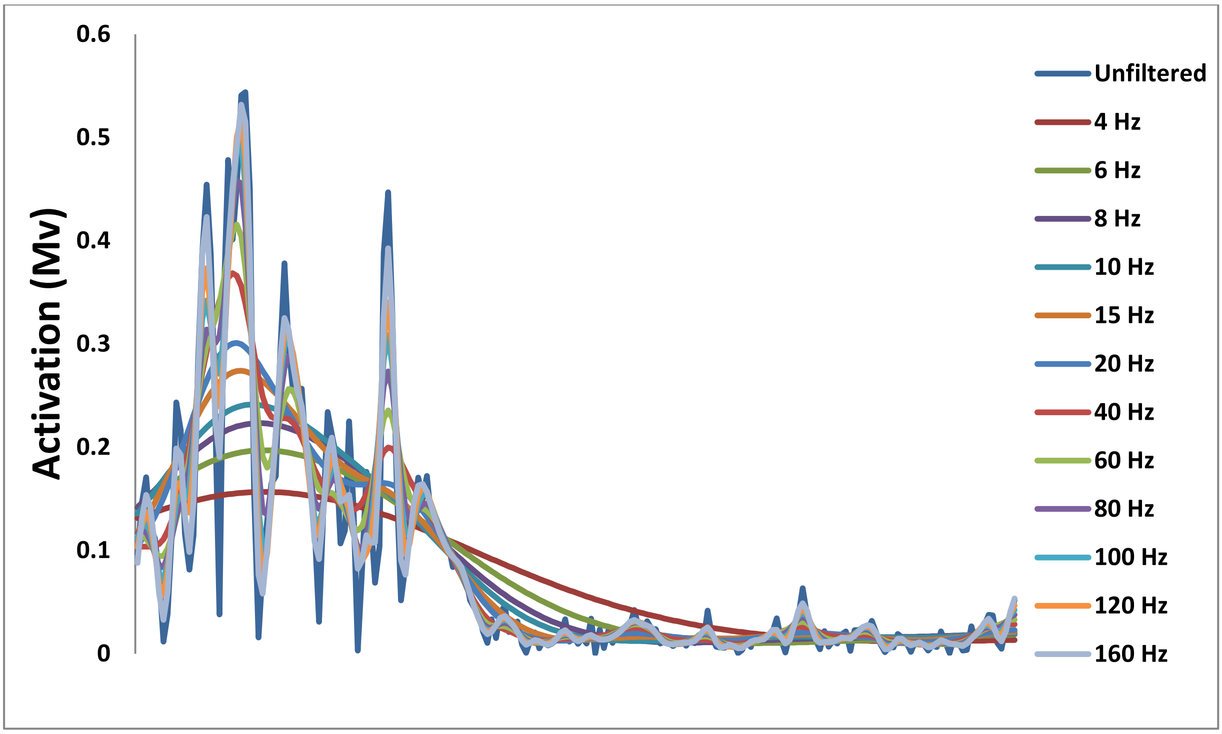

Figure 8.4: Representative stance phase EMG signals obtained from the Vastus Medialis as a function of cut-off frequency.

Table 8.4: Muscle activation (mean and standard deviation) from the Vastus Medialis as a function of cut-off frequency (* = significant main effect).

	Vastus Medialis (mV)	
	Peak	**Mean**
Unfiltered	0.91 ± 0.58	0.21 ± 0.21
4Hz	0.30 ± 0.25	0.28 ± 0.28
6Hz	0.33 ± 0.25	0.28 ± 0.28
8Hz	0.36 ± 0.29	0.28 ± 0.28
10Hz	0.39 ± 0.31	0.28 ± 0.28
15Hz	0.44 ± 0.35	0.28 ± 0.28
20Hz	0.48 ± 0.38	0.28 ±0.27
40Hz	0.59 ± 0.46	0.28 ± 0.27
60Hz	0.69 ± 0.52	0.28 ± 0.27
80Hz	0.75 ± 0.55	0.28 ± 0.27
100Hz	0.80 ± 0.57	0.28 ± 0.27
120Hz	0.83 ± 0.58	0.28 ± 0.27
140Hz	0.86 ± 0.58	0.28 ± 0.27
160Hz	0.87 ± 0.59	0.28 ± 0.27
	*	

The results indicate that for mean stance phase activation that no significant $F_{(1.01, 6.07)} = 2.37$, $p > 0.05$, $\eta^2=0.28$ differences exist between the different cut-off frequencies. However a significant main effect $F_{(13, 78)} = 19.48$, $p \leq 0.01$, $\eta^2=0.77$ was observed for the magnitude of peak stance phase activation.

8.1.4 Discussion

It is clear from this pilot investigation that different cut-off frequencies can significantly affect the interpretation of the resultant EMG amplitudes. Thus, selecting the correct cut-off frequency is essential in order to be able to draw meaningful conclusions from the data. Similar to previous pilot investigations examining the kinematic and tibial acceleration signals the linear EMG envelopes evidences both over and under smoothing of the data.

The results of this pilot investigation appear at least partially oppose the general cut-off frequency guideline of 3-6 Hz recommended by Winter (1990) as the lower cut-off frequencies i.e. 4 and 6 Hz indicate that the EMG curve has been over smoothed which suggests that this general recommendation with regards to the optimal cut-off frequency cannot be applied to running. It appears based on both the Merletti and Parker (2004) and Shiavi, et al, (1998) recommendations and observation of the EMG profiles that a cut off frequency of 20Hz appears to be the most appropriate for the filtering of the EMG signals.

8.2 Pilot study 2 Determining the most appropriate technique for MVC normalization of EMG signals

8.2.1 Introduction

Normalization of EMG is employed to reduce inter-subject variability and provide an empirically meaningful representation of the signal amplitude. EMG is normalized in relation to a maximum voluntary contraction (MVC) (Burden et al. 2003). MVC normalization is traditionally accomplished using isometric contractions but dynamic actions are also used. The utilization of different normalization techniques may alter the magnitude of the resultant EMG (Burden et al. 2003), yet the most effective method for normalization is unknown and many different techniques are used. The aim of this investigation was to determine the most appropriate technique from the relevant methods available within the literature.

8.2.2 Methods

8.2.2 (i) Participants

Eight male participants (Age: 23.67 $\pm$ 5.71years, mass: 70.76 $\pm$ 8.21kg, Height 1.72 $\pm$ 0.31m) volunteered to take part in this investigation. Ethical approval was obtained from the University of Central Lancashire School of Psychology ethics committee.

8.2.2 (ii) Procedure

Surface EMG activity was obtained during the stance phase at 1000Hz from the Vastus Lateralis (VL), Vastus Medialis (VM), Tibialis Anterior (TA), Gastrocnemius (GM) and Biceps Femoris (BF) using the same protocol outlined in 8.5 as participants ran at 4.0m.s^{-}

[1]±5% over a force platform. 5 running and isometric MVC trials were collected twice (A and B) without removing the electrodes. The following normalization techniques were examined:

Dynamic Mean Task (DMT): Mean stance phase EMG obtained from ensemble average

Dynamic Peak Task (DPTa): Peak stance phase EMG obtained from ensemble average.

Dynamic Peak Task (single) (DPTb): Peak stance phase EMG obtained a single trial which encompasses the peak value from all five trials.

Arbitrary isometric MVC (ArbISO): Maximum EMG amplitude from a maximal isometric MVC, from a self-selected joint angle.

Angle specific isometric MVC (AngISO): Maximum EMG amplitude from a maximal isometric MVC, from a pre-determined joint angle.

Isometric MVC's were obtained using an isokinetic dynamometer (Isocom, Eurokinetics) in accordance with Norcross et al. (2009).

8.2.2 (iii) Data processing

Following full wave rectification EMG signals were filtered using a 20Hz zero lag 4th order low-pass filter to create a linear envelope.

8.2.2 (iv) Statistical analysis

Pearson's correlations were employed in order to determine the strength of the relationship (i.e. reliability) between A and B EMG amplitudes from each normalization method. In addition, repeated measures ANOVA's were utilized for each muscle to examine the differences in MVC reference amplitude (mV) from the 5 different MVC techniques Comparative analyses were conducted on the mean values from both the A and B trials in accordance with Sinclair et al. (2012).

8.2.3 Results

Table 8.5: Correlations (R) for each muscle from each MVC normalization procedure (* = Significant correlation)

	TA	**GM**	**VL**	**VM**	**BF**
DMT	0.83 *	0.77 *	0.35*	0.77 *	0.1
DPTa	0.98 *	0.95 *	0.93*	0.92 *	0.97 *
DPTb	0.76 *	0.73 *	0.48	0.86 *	0.88 *
ArbISO	0.83 *	0.46	0.73 *	0.61	0.24
AngISO	0.31	0.25	0.74 *	0.76 *	0.55

Table 8.6: EMG MVC amplitudes (mV) (mean and standard deviation) of each muscle from each MVC normalization procedure (* = Significant main effect).

	TA	**GM**	**VL**	**VM**	**BF**
DMT (mV)	0.08 ± 0.04	0.05 ± 0.01	0.09 ± 0.02	0.09 ± 0.02	0.07 ± 0.02
DPTa (mV)	0.17 ± 0.04	0.30 ± 0.07	0.26 ± 0.04	0.25 ± 0.06	0.18 ± 0.08
DPTb (mV)	0.19 ± 0.06	0.40 ± 0.19	0.28 ± 0.08	0.32 ± 0.13	0.21 ± 0.09
ArbISO (mV)	0.36 ± 0.06	0.14 ± 0.09	0.17 ± 0.07	0.27 ± 0.12	0.19 ± 0.06
AngISO (mV)	0.48 ± 0.17	0.15 ± 0.03	0.19 ± 0.07	0.22 ± 0.50	0.19 ± 0.06
	*	*	*	*	*

The results indicate that significant MVC amplitude main effects exist for the TA $F_{(1.55, 10.85)} = 36.78$, $p \leq 0.01$, $\eta^2=0.82$, GM $F_{(1.38, 9.68)} = 18.35$, $p \leq 0.01$, $\eta^2=0.72$, VL $F_{(1.65, 11.58)} = 12.97$, $p \leq 0.01$, $\eta^2=0.56$, VM $F_{(2.01, 14.29)} = 18.21$, $p \leq 0.01$, $\eta^2=0.72$ and BF $F_{(4, 28)} = 13.89$, $p \leq 0.01$, $\eta^2=0.67$.

8.2.4 Discussion

The aim of this study was to determine the most effective EMG MVC normalization technique for running analyses. The results of this investigation suggest that different normalization techniques can significantly influence the interpretation of the normalized EMG magnitude, thus selection of the most appropriate technique is critical to achieve empirically meaningful findings. Importantly, this investigation observed that isometric MVC normalization techniques exhibited low reliability which is concerning given their frequent application. This concurs with the observations of Yang and Winter (1983). The results of this pilot study indicate that the most reliable normalization technique was the DPTa method and thus appears to be the most dependable in order to provide meaningful EMG data. Whilst this technique does mean that peak stance phase amplitude cannot be reported as it will always result in a normalized amplitude of one i.e. 100% MVC, its utilization is still justified given the lack of reliability of the remaining techniques and thus it will be utilized throughout this chapter.

8.3 Pilot study 3 Reliability of metalyzer gas analysis system

8.3.1 Introduction

In order for the findings/conclusions from the gas analysis aspect of the project to be reliable, it must be evidenced that the protocol for the acquisition of VO_2 information is repeatable. The aim of this pilot investigation was to determine the within and inter-session reliability of the VO_2 acquisition protocol.

8.3.2 Methods

8.3.2 (i) Participants

Five male participants (Age: 23.2 ± 0.84years, height: 1.73 ± 0.08m, body mass: 66.8 ± 3.63kg) took part in this investigation. All provided written informed consent. Ethical approval for this project was obtained from the University of Central Lancashire School of Psychology ethics committee.

8.3.2 (ii) Procedure

Participants were required to attend on two separate occasions. In the first session participants performed two running trials which were contrasted allowing intra reliability to be examined. On the second session participants performed a further trial which was contrasted against the first trial obtained in the first session allowing inter reliability to be examined. Gas analysis was obtained using a Metalyzer 3B (Cortex INC Germany) system using the same protocol outlined later in this chapter. Six minutes of steady state VO_2 was obtained whilst participants completed treadmill running at 4.0m.s^{-1}.

For within session reliability the same protocol that was utilized in the main study was observed. Inter-session analyses were conducted over two sessions, with at a minimum of 24 hours between sessions. Testing sessions were conducted at the same time of day to reduce the potential variation in VO_2 due to circadian rhythms.

8.3.2 (iii) Statistical Analysis

Differences in mean steady state from both analyses (within and inter-session) VO_2 were compared using paired samples t-test's with significance set at $p \leq 0.05$. Finally, to further assess reliability Pearson's correlations were conducted between the test and retest data.

8.3.3 Results

The results indicate that no significant differences (p=0.85) were found for within session test (VO_2= 42.24 ± 3.30ml.min.kg^{-1}) and retest (VO_2=42.66 ± 3.20ml.kg.min^{-1}) sessions indicating a difference of 1.01%. Furthermore, no significant differences (p=0.56) were found for inter-session test (VO_2= 42.09± 3.40ml.min.kg^{-1}) and retest (42.59 ± 3.49) testing sessions with a difference of 1.19%.

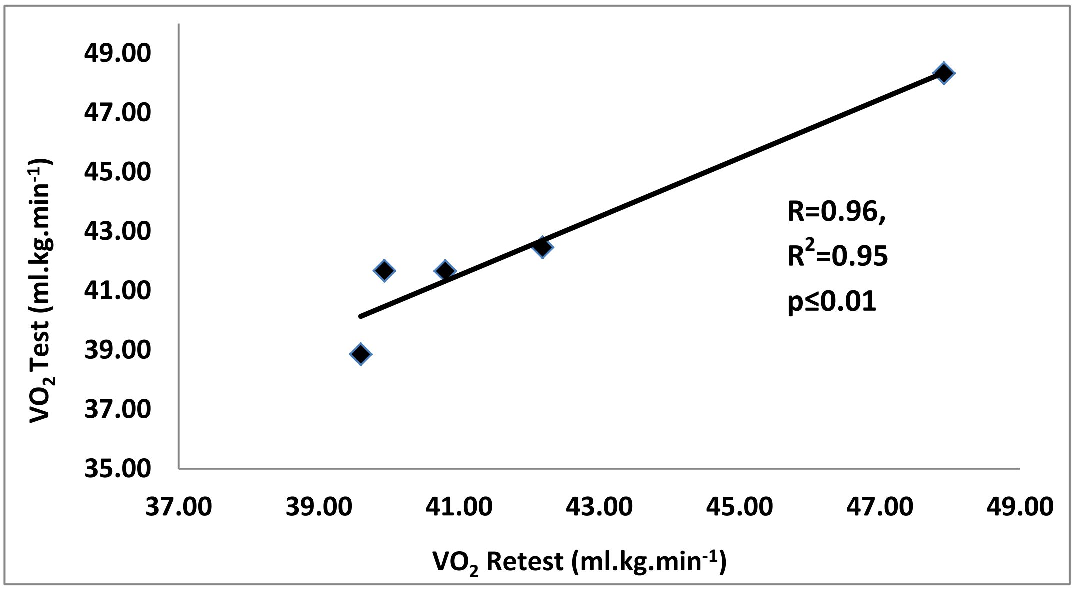

Figure 8.5: Relationship between within session test and retest VO_2 values.

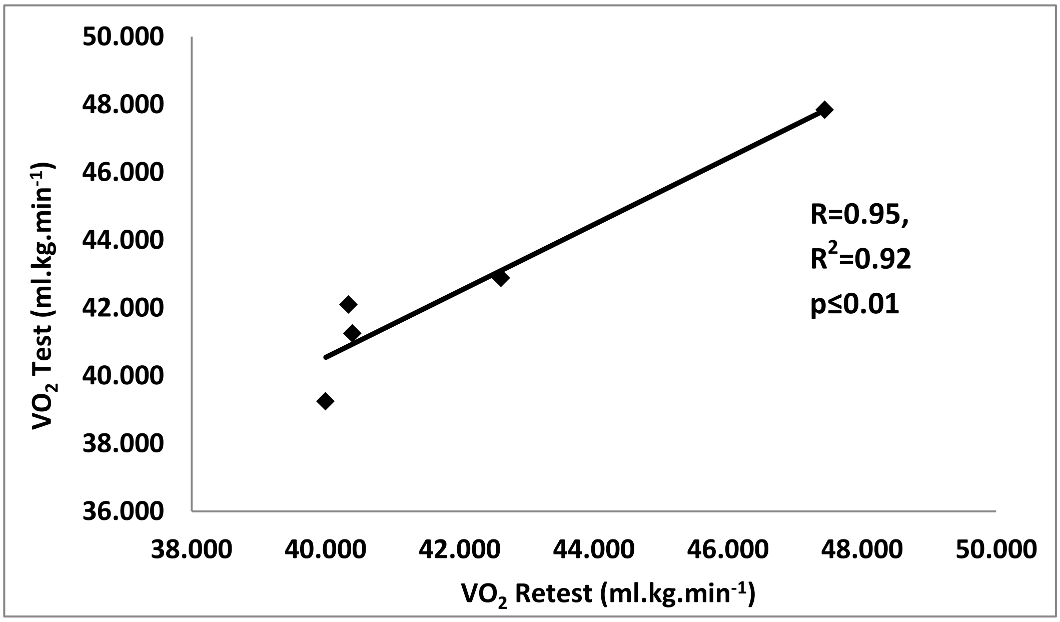

Figure 8.6: Relationship between inter-session test and retest VO_2 values.

8.3.4 Discussion

The results of this pilot investigation suggest that the observed VO_2 parameters can be reliably obtained. The ≤1.19% difference between test and retest sessions, suggests that any differences in the main study outside this deviation can be considered to be a true effect. This deviation concurs with the findings of Frederick et al. (1986), but is smaller than the 2.48% reported by Armstrong and Costill, (1985). There are several possible explanations for this finding i.e. why the data from this investigation exhibited lower variability between sessions, 1. participants were familiarized with the treadmill and metalyzer protocol prior to data collection and as such may have been better acquainted with the protocol and 2. participant's weights were adjusted day-day in order to correct for any variations in mass that may have influenced the results. Furthermore, very strong correlations were observed between the test and retest VO_2 data which suggests that the measures obtained from the metalyzer gas analysis system can be obtained consistently between testing sessions, thus confirming the efficacy of the findings from the main study.

8.4 Pilot study 4– Shoe COM deviation following addition of mass

8.4.1 Introduction

One of the aims of chapter 8 is to determine how footwear with different shock attenuating properties (i.e. Saucony and Nike Free) influence the energy cost of steady state running. However, the tested shoes weighed different amounts and as such the metabolic VO_2 analysis would be confounded by this deviation in mass (Frederick, 1986). As such it was determined to increase the validity of the investigation that the mass of the lighter Nike Free shoes would be increased, to match the Saucony footwear. However, it must be shown that in placing this additional mass on the shoes that it does not affect the balance of the shoe. As such the aim of this pilot investigation was to determine the effect that the addition of this mass has on the balance of the shoes.

8.4.2 Methods

8.4.2 (i) Shoes

In order to add mass to the Nike Free footwear lead tape was used (Figure 8.7). In the first instance the tape was placed strategically around the shoe so that the balance of the shoe would be minimally affected.

Figure 8.7: Nike Free shoes fitted with lead tape

8.4.2 (ii) Assessing shoe balance

As the size of the measurement volume was small a miniature calibration L-frame and wand were used to calibrate the required area (Figure 8.8). The reduced calibration volume allied with the smaller calibration apparatus meant that that a very accurate calibration could be obtained. In order to determine the shoes balance prior to and subsequent to the addition of weight to the footwear, each shoe was placed inside a very lightweight plastic box and stuck down with rigid tape and didn’t move. The box was suspended from the ceiling.

Figure 8.8: Miniature calibration wand and L-frame.

The box was also fitted with smaller 12mm retro-reflective markers to each corner attached allowing it to be tracked very accurately using the 3-D camera system (Figure 8.9) and defined as a rigid segment within visual 3-D (C-Motion Inc, Germantown, USA) in order to determine the position of its COM in all three axes (X, Y, Z) with respect to the lab co-ordinate system origin (Figure 8.10). The box was hung from all three of its axes and the COM in the vertical Z axis was obtained before and after the addition of the lead tape. Numerous data captures were required as the position of the lead tape was finely adjusted following acquisition of the shoe COM until an acceptable level of accuracy was obtained.

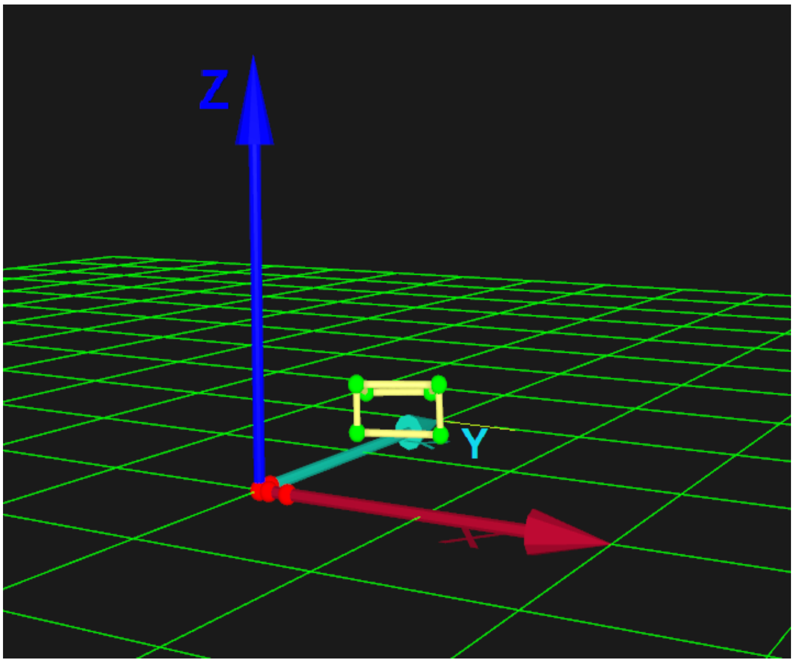

Figure 8.9: Box tracked using Qualisys track manager with respect to the lab co-ordinate axes.

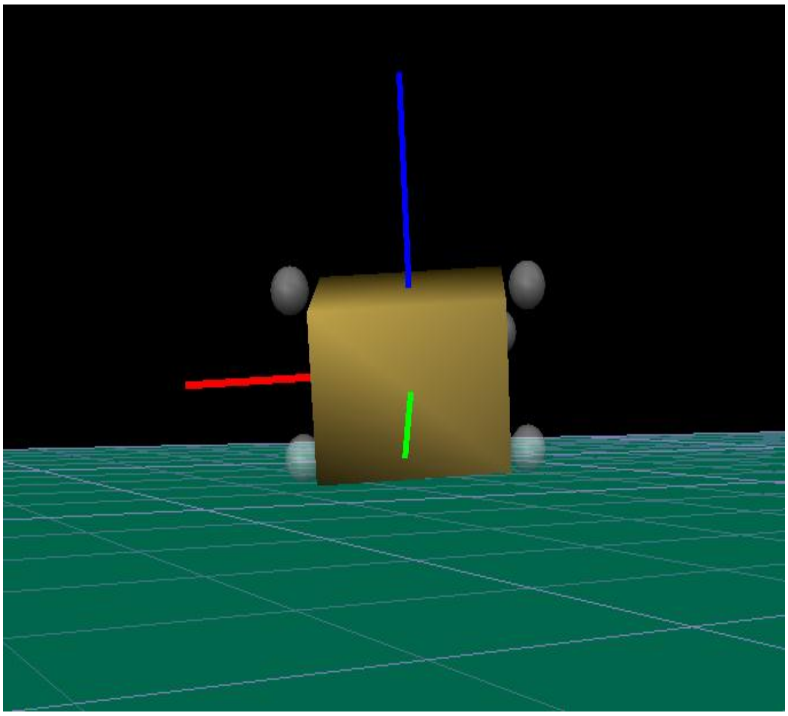

Figure 8.10: Box containing the weighted/un-weighted footwear defined as a segment within visual 3-D.

8.4.3 Results

Table 8.7: Shoe COM deviation in all three axes from the centre of the lab co-ordinate system

		X	Y	Z
Size 9 - Right Shoe	No Weight (m)	0.06848	0.12373	0.02139
	Weighted (m)	0.06853	0.12379	0.02137
Size 9 - Left Shoe	No Weight (m)	0.06964	0.12367	0.02188
	Weighted (m)	0.06978	0.1237	0.02188
Size 7 - Right Shoe	No Weight (m)	0.08661	0.12428	0.02124
	Weighted (m)	0.08651	0.12438	0.02123
Size 7 - Left Shoe	No Weight (m)	0.08675	0.12507	0.02105
	Weighted (m)	0.08675	0.12511	0.02100

8.4.4 Discussion

The results of this pilot investigation suggest that following the placement of additional mass to the shoe, that the COM of the shoes have been minimally affected. This suggests that excluding the influence that the additional mass may have on the inertial properties of the shoes that the mechanics of the shoes are unaltered.

8.5 Main study

8.5.1 Introduction

The economy of running which is a reflection of the amount of oxygen required to maintain a given velocity, is considered a very important factor for the determination of distance running performance (Bransford, 1977; Morgan et al. 1989; Weston 2000). As such it is hypothesized that improvements in running economy would be of significant value to those wishing to improve running performance.

Given the influence of running economy on running performance, a number of investigations have aimed to determine the biomechanical parameters that may affect running economy (Cavanagh and Williams 1982; Williams 1986; Williams 1987; Anderson 1994; Kyrolainen, 2001; Heise, 2001). Several discrete kinematic parameters have been linked to the economy of running. Cavanagh, (1977) observed that economic runners were associated with reduced vertical oscillation during the gait cycle and were more symmetrical compared to less economic athletes. Williams, (1986) found runners associated with high running economy exhibited reduced knee flexion at heelstrike and an increased maximal velocity of plantarflexion. Williams, (1987) contrasted low, medium and high VO_2 runners, it was observed that low VO_2 runners were associated with a higher tibial angle at heelstrike, reduced plantarflexion angle at toe-off and increased peak knee flexion. Kyrolainen, (2001) attempted to relate several kinetic, kinematic and electromyographical parameters to running economy. None of the observed parameters were found to be related to running economy.

However, although significant differences and trends have been observed between running economy and some biomechanical parameters, these connections are often weak and lack consistency between investigations (Anderson, 1996; Saunders, 2004). This is due to the complex interrelationships amongst the multitude of discrete mechanical descriptors of the running technique that globally influence running economy. The mechanical factors related to running economy do not act independently and parameters that contribute to increased VO_2 may be counterbalanced by another element of overall running mechanics. Therefore, definitive conclusions regarding the influence of biomechanical parameters on running economy have yet to be drawn, thus further investigation using 3-D kinematics, and muscle electromyographic parameters and more appropriate statistical techniques to facilitate multiple variables are required.

The possibility that athletic footwear can influence the economy of distance running has also been examined previously (Frederick et al. 1986). A number of studies allude to the assumption that it is more energetically economical to run in footwear with appropriate mechanical characteristics (Frederick et al. 1986). Footstrike induced transient impact forces produce muscle vibrations, that are attenuated by muscle tuned pre-activation (Nigg, 1997; Wakeling and Nigg, 2001). Therefore, the amount of work done during running may be influenced by the foot ground interface, (Nigg and Anton, 1995). Different footwear shock attenuating properties have the potential to reduce impact forces, alter running kinematics, change muscle activity, and thus influence running economy. Several investigations suggest the midsole characteristics of the running shoe can influence energy expenditure. Bosco and Rusco, (1983) found that shoes incorporating a shock absorbing viscoelastic material increased running economy during treadmill running. Frederick et al. (1986) reported that running in a shoe with a gas-inflated cushioning system reduced the economy of treadmill running by 2.4% when compared with a conventional shoe incorporating foam cushioning.

Nigg et al. (2003) examined the influence of footwear with different midsole material characteristics on muscle activation and running economy, it was observed that wearing viscoelastic midsoles as opposed to hard midsoles reduced VO_2 by 2% but the result did not reach statistical significance.

These findings suggest therefore that footwear with different midsole cushioning properties have the potential to produce alterations in running economy through changes in kinematics and muscle activity. However, to date there have been no investigations which have related running economy to simultaneous measurements of 3-D kinematics, impact kinetics and muscular activation parameters. Therefore the aim of the current investigation was to determine the influence of footwear with different shock attenuating properties on the energy requirements of distance running and to examine the biomechanical parameters which have the strongest association with running economy using regression analyses.

8.5.2 Methods

8.5.2 (i) Participants

Twelve experienced male runners volunteered to take part in this study. Participants were active runners engaging in training at least three times per week; completing a minimum of 35 km and had previous experience of treadmill running. All were free from pathology at the time of data collection and provided written informed consent. The mean characteristics of the participants were; age 23.74 ± 2.34 years, height 1.77 ± 0.06 m and body mass 75.57 ± 7.56 kg. An a priori statistical power analysis was conducted using the Hopkins method based on a moderate effect size and a power measure of 80%, which suggested that 12 subjects were adequate for the design. The procedure was approved by the University of Central Lancashire, School of Psychology ethics committee.

8.5.2 (ii) Procedure

Participants completed treadmill running at $4.0 m.s^{-1}$ on a Woodway™ (ELG, Steinackerstrasse D-79576 Weil Rhein-Germany) high powered treadmill maintained at a gradient of 0%. Participants were given a 5 minute period of habitation, in which they ran at the required velocity prior to the commencement of data collection in accordance with (Hanson et al. 2010). In accordance with the protocol described in chapter 5 footstrike and toe-off events were determined using the Dingwell et al. (2001) method.

Kinematics, EMG and tibial acceleration data were synchronously collected. Kinematic data was captured at 250 Hz via a ten camera motion analysis system (Qualisys Medical AB, Goteburg, Sweden). Dynamic calibration with the same acceptance criteria as outlined in chapter 5 was conducted prior to data collection.

8.5.2 (iii) 3-D Kinematics

3-D kinematics from the lower extremities were obtained using the same protocol as in chapter 5. Static trials were conducted for each footwear condition.

8.5.2 (iv) Tibial Accelerations

Tibial accelerations were also obtained using the protocol outlined in chapter 5.

8.5.2 (v) Surface EMG

Surface EMG activity was obtained at 1000Hz from the Vastus Lateralis (VL), Vastus Medialis (VM), Tibialis Anterior (TA), Gastrocnemius (GM) and Biceps Femoris (BF) muscles using bipolar electrodes with an inter-electrode distance of 20 mm. connected to an interface unit. To minimize myoelectric contributions from close synergistic muscles, all electrodes were placed on the bellies on the appropriate muscles in alignment with the muscle pennation according to the SENIAM recommendations (Freriks et al. 1999). The skin was shaved and prepared with abrasive paper and ethanol to lower the skin impedance and favour proper recordings of the muscle potentials. To minimize artefact, the electrodes and the cables from electrodes to the interface unit were fixed to the body surface using surgical tape. The electrodes and electrode wires were wrapped on thigh and shank with an elastic bandage do prevent dislocation during running.

8.5.2 (vi) VO_2 Gas Analysis

Breath-by-breath measurements of respiratory gases were made using a Metalyser 3B® (Cortex Biophysic, Leipzig, Germany). Heart rate (HR) was monitored during rest between footwear conditions with a Polar watch (Polar Electro Oy, Kempele, Finland). HR was not examined during dynamic trials as it was influenced by the camera system. Before each testing session the system was calibrated by inputting the atmospheric pressure, following which the pneumotach volume sensor was also calibrated using a 3 litre syringe (Hans Rudolph Inc, Kansas city USA). Lastly, the gas sensors were calibrated using ambient air and known gas concentrations of 5.09% O_2 and 14.46% CO_2. All testing sessions were conducted at a similar time of day to eliminate the potential variation in VO_2 due to circadian rhythm. The testing protocol itself consisted of steady-state runs of 6 min duration in accordance with the Nigg et al. (2003) procedure. In agreement with Frederick et al. (1986) participants reported to the laboratory a minimum of 4 hours postprandial and the order of wearing each footwear condition was counterbalanced. Participants completed their runs in both footwear conditions within the same testing session with rest in between each. In accordance with the Hanson et al. (2010) protocol the subsequent testing condition was not started until the participant's HR was less than 110 beats per minute and they felt ready to begin the next testing condition.

8.5.2 (vii) Data Processing

Trials were digitized in Qualisys Track Manager to identify anatomical and tracking markers then exported as C3-D files. Kinematic parameters were quantified using Visual 3-D (C-Motion Inc, Germantown, USA) after marker data was smoothed using a low-pass Butterworth 4th order zero-lag filter at a cut off frequency of 10 Hz. 3-D kinematics of the hip knee and ankle joints were calculated using an XYZ cardan sequence of rotations (where X is sagittal; Y is coronal and is Z is transverse plane rotation). Trails were normalized to 100% of the stance phase then processed gait trials were averaged. 3-D kinematic measures from the hip, knee and ankle which were extracted for statistical analysis were 1) angle at footstrike, 2) angle at toe-off, 3) ROM during stance, 4) peak angle during stance and 5) relative ROM from footstrike to peak angle, 6) velocity at footstrike, 7) velocity at toe-off and 8) peak velocity.

Peak positive axial tibial acceleration was defined as the highest positive acceleration peak measured during the stance phase. To analyze data in the frequency domain, a FFT function was performed and median power frequency content of the acceleration signals were calculated in accordance with (Lafortune and Hennig, 1995).

The EMG signals from each muscle were full wave rectified and filtered using a 20 Hz Butterworth zero lag low-pass 4th filter to create a linear envelope. EMG data from each muscle were normalised to an MVC which was obtained from for each participant as the peak stance phase EMG amplitude obtained from an ensemble average of participants completed trials. Electromyographic measures extracted from each muscle for statistical analysis were; mean normalized amplitude during the stance phase and for the 50 ms prior to footstrike in accordance with Nigg et al. (2004) and Arndt et al. (2005).

8.5.2 (viii) Shoes

The shoes utilized during this study consisted of a Saucony Pro Grid Guide II and a Nike Free 3.0 as they were known to be associated with different shock attenuating properties in both mechanical and human testing as evidenced by chapters 4.10 and 7.2. The shoes were the same for all runners; they differed in size only (sizes 7 and 9 in men's shoe UK sizes).

8.5.2 (ix) Statistical analysis

Descriptive statistics including means and standard deviations were calculated for each footwear condition. Differences in 3-D kinematics, impact shock and EMG parameters were examined using paired t-tests with significance accepted at the $p \leq 0.05$ level. The Shapiro-Wilk statistic for each footwear condition confirmed that the normal distribution assumption was met in all cases.

Correlation coefficients to peak tibial acceleration were also quantified for the normalized EMG amplitude in the 50 time window prior to footstrike (pre-activation) in accordance with Nigg et al. (2003) and Arndt et al. (2005). Finally, multiple regression analyses with VO_2 as criterion and the biomechanical parameters as independent variables were carried out for each footwear using a forward stepwise procedure with significance accepted at the $p \leq 0.05$ level. The independent variables were examined for co-linearity prior to entry into the regression model and those exhibiting high co-linearity $R \geq 0.7$ were removed. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA).

8.5.3 Results

8.5.3 (i) Running economy

For VO_2, no significant ($p=0.87$) differences were observed between footwear conditions; Saucony= 42.72 ± 2.17 and Nike Free = 42.75 ± 1.95 $ml.kg.min^{-1}$. A small mean difference of 0.09% was thus observed between footwear, although individual participants did exhibit % differences as high as 2.50% (See Appendix E).

8.5.3 (ii) Regression analyses

Two biomechanical parameters were obtained for each footwear condition as significant predictors of VO_2. For the Saucony shoes the regression model yielded and Adj R^2 of 0.81, $p \leq 0.01$. Peak stance phase hip flexion (B=0.62, t=5.21) Adj R^2=0.62, $p \leq 0.01$ and peak sagittal plane ankle velocity (B=0.47, t=3.60) Adj R^2=0.19, $p \leq 0.01$ were found to be significant predictors of VO_2. For the Nike Free footwear the stepwise regression model yielded and Adj R^2 of 0.82, p≤0.01. Peak sagittal plane knee excursion (*B*=0.88, t=6.91) Adj R^2=0.49, p≤0.01 and stance phase activation of the VM (*B*=0.59, t=4.62) Adj R^2=0.33, p≤0.01 were found to be significant predictors of VO_2.

8.5.3 (iii) Kinetic, temporal and EMG parameters

Table 8.8: Kinetic and electromyographic parameters (means, standard deviations) and a function of footwear (* = Significant main effect).

	Saucony	**Nike Free**	
Peak Axial impact shock (g)	4.28 ± 4.28	5.47 ± 1.83	*
Median Power Frequency (Hz)	12.63 ± 0.66	13.42 ± 1.02	*
Stance time (ms)	211.25 ± 14.43	210.33 ± 13.76	
Tibialis Anterior 50ms (MVC)	0.51 ± 0.15	0.55 ± 0.21	
Gastrocnemius 50ms (MVC)	0.14 ± 0.09	0.11 ± 0.06	
Biceps Femoris 50ms (MVC)	0.26 ± 0.13	0.27 ± 0.12	
Vastus Lateralis 50ms (MVC)	0.16 ± 0.10	0.17 ± 0.08	
Vastus Medialis 50ms (MVC)	0.23 ± 0.13	0.21 ± 0.16	
Tibialis Anterior stance (MVC)	0.15 ± 0.07	0.17 ± 0.07	
Gastrocnemius stance (MVC)	0.34 ± 0.07	0.34 ± 0.08	
Biceps Femoris stance (MVC)	0.24 ± 0.06	0.27 ± 0.09	
Vastus Lateralis stance (MVC)	0.27 ± 0.08	0.27 ± 0.07	
Vastus Medialis stance (MVC)	0.26 ± 0.08	0.31 ± 0.09	

Table 8.9: Correlation coefficients between peak tibial acceleration and EMG prior to footstrike.

	Saucony	**Nike Free**
TA (EMG 50)	R=0.77, R^2=0.59, p≤0.01	R=0.70, R^2=0.49, p≤0.01
GM (EMG 50)	R=0.09, R^2=0.15, p>0.05	R=0.39, R^2=0.15, p>0.05
VL (EMG 50)	R=0.31, R^2=0.10, p>0.05	R=0.4, R^2=0.16, p>0.05
VM (EMG 50)	R=0.02, R^2=0.004, p>0.05	R=0.08, R^2=0.006, p>0.05
BF (EMG 50)	R=0.53, R^2=0.28, p>0.05	R=0.58, R^2=0.34, p>0.05

8.5.3 (iv) 3-D kinematic parameters

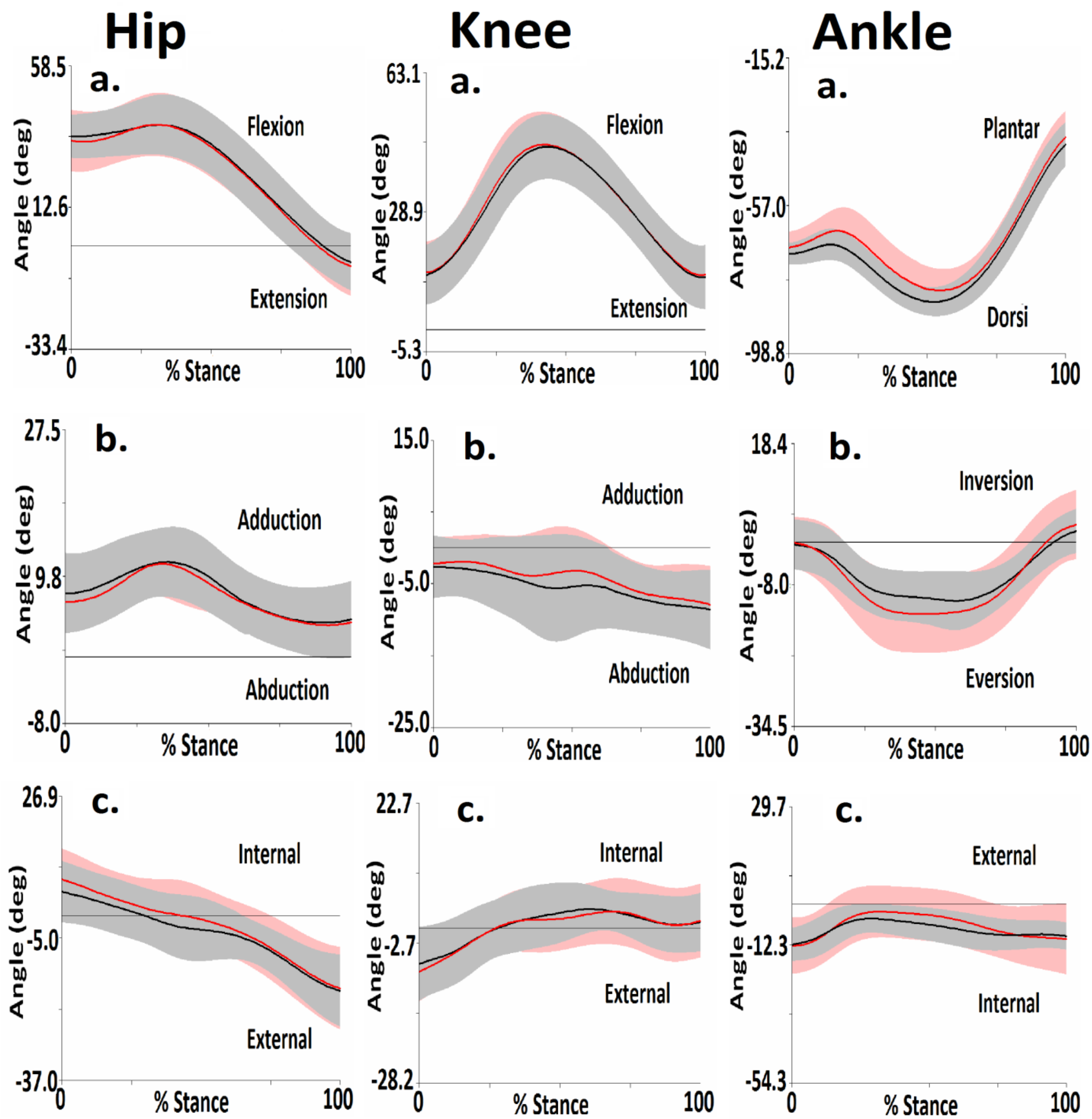

Figure 8.11: Mean and standard deviation hip, knee and ankle joint kinematics in the a. sagittal, b. coronal and c. transverse planes the Saucony (black line) and Nike free (red line), footwear conditions (shaded area is 1 ± SD, black= Saucony and red = Nike free).

Table 8.10: Hip joint kinematics (mean and standard deviation) from the stance limb as a function of footwear (* = Significant main effect).

	Saucony	Nike Free	
Hip			
X (+=flexion/-=extension)			
Angle at Footstrike (°)	33.59 ± 8.53	31.87 ± 10.19	
Angle at Toe-off (°)	-7.95 ± 10.07	-88.34 ± 10.20	
Range of Motion (°)	41. 52 ± 4.69	40.21 ± 5.49	
Relative Range of Motion (°)	5.53 ± 4.38	6.83 ± 5.60	
Peak Flexion (°)	39.17 ± 11.05	38.69 ± 9.46	
Y (+=adduction - =abduction)			
Angle at Footstrike (°)	6.57 ± 4.57	6.20 ± 3.07	
Angle at Toe-off (°)	4.26 ± 5.02	4.23 ± 2.49	
Range of Motion (°)	3.96 ± 2.15	3.43 ± 2.15	
Relative Range of Motion (°)	5.14 ± 2.43	5.84 ± 2.11	
Peak Adduction (°)	11.23 ± 3.26	12.04 ± 2.87	
Z (+=internal/- =external)			
Angle at Footstrike (°)	7.19 ± 6.09	9.69 ± 6.85	
Angle at Toe-off (°)	-16.12 ± 6.73	-15.29 ± 7.72	
Peak external Rotation (°)	8.06 ± 5.20	10.70 ± 5.69	
Relative Range of Motion (°)	1.04 ± 1.43	1.09 ± 1.99	
Range of Motion (°)	23.58 ± 4.66	24.98 ± 4.36	

Table 8.11: Knee joint kinematics (mean and standard deviation) from the stance limb as a function of footwear (* = Significant main effect).

	Saucony	**Nike Free**	
Knee			
X (+=flexion/-=extension)			
Angle at Footstrike (°)	13.95 ± 7.08	13.27 ± 7.01	
Angle at Toe-off (°)	13.04 ± 6.47	13.48 ± 4.25	
Range of Motion (°)	5.91 ± 2.99	6.68 ± 3.43	
Relative Range of Motion (°)	32.03 ± 5.63	32.14 ± 6.49	
Peak Flexion (°)	46.17 ± 8.55	45.42 ± 8.25	
Y (+=adduction - =abduction)			
Angle at Footstrike (°)	-1.34 ± 4.74	-1.09 ± 4.11	
Angle at Toe-off (°)	-7.29 ± 5.38	-6.91 ± 5.59	
Range of Motion (°)	5.99 ± 2.27	5.84 ± 2.19	
Relative Range of Motion (°)	7.10 ± 2.58	6.54 ± 1.90	
Peak Angle (°)	-8.45 ± 5.55	-7.62 ± 5.21	
Z (+=internal/- =external)			
Angle at Footstrike (°)	-7.63 ± 6.05	-8.90 ± 5.34	
Angle at Toe-off (°)	0.56 ± 4.74	1.01 ± 5.90	
Range of Motion (°)	5.91 ± 2.99	9.97 ± 4.16	
Relative Range of Motion (°)	13.60 ± 4.92	14.09 ± 3.54	
Peak Internal Rotation (°)	6.06 ± 1.97	5.19 ± 5.12	

Table 8.12: Ankle joint kinematics (mean and standard deviation) from the stance limb as a function of footwear (* = Significant main effect).

	Saucony	**Nike Free**	
Ankle			
X (+ =plantar/- =dorsi)			
Angle at Footstrike (°)	-70.85 ± 2.41	-69.28 ± 3.06	
Angle at Toe-off (°)	-39.78 ± 4.25	40.25 ± 4.77	
Range of Motion (°)	29.64 ± 4.85	30.60 ± 3.71	
Relative Range of Motion (°)	14.05 ± 3.83	14.76 ± 3.19	
Peak Dorsi-Flexion (°)	-83.62 ± 4.21	-85.62 ± 5.22	
Y (+=inversion/ -=eversion)			
Angle at Footstrike (°)	-0.94 ± 4.27	-1.20 ± 3.82	
Angle at Toe-off (°)	0.75 ± 3.48	1.35 ± 6.67	
Range of Motion (°)	5.20 ± 2.37	4.35 ± 2.22	
Relative Range of Motion (°)	13.12 ± 3.19	15.40 ± 4.49	
Peak Eversion (°)	-13.73 ± 4.26	-17.11 ± 6.96	
Z (-=internal/ +=external)			
Angle at Footstrike (°)	-11.77 ± 5.38	-14.01 ± 5.99	
Angle at Toe-off (°)	-9.87 ± 3.68	-12.59 ± 9.84	
Range of Motion (°)	5.03 ± 3.32	5.70 ± 4.06	
Relative Range of Motion (°)	9.35 ± 3.06	12.26 ± 5.43	
Peak Angle (°)	-2.50 ± 4.60	-1.75 ± 6.88	

8.5.3 (v) Joint velocities

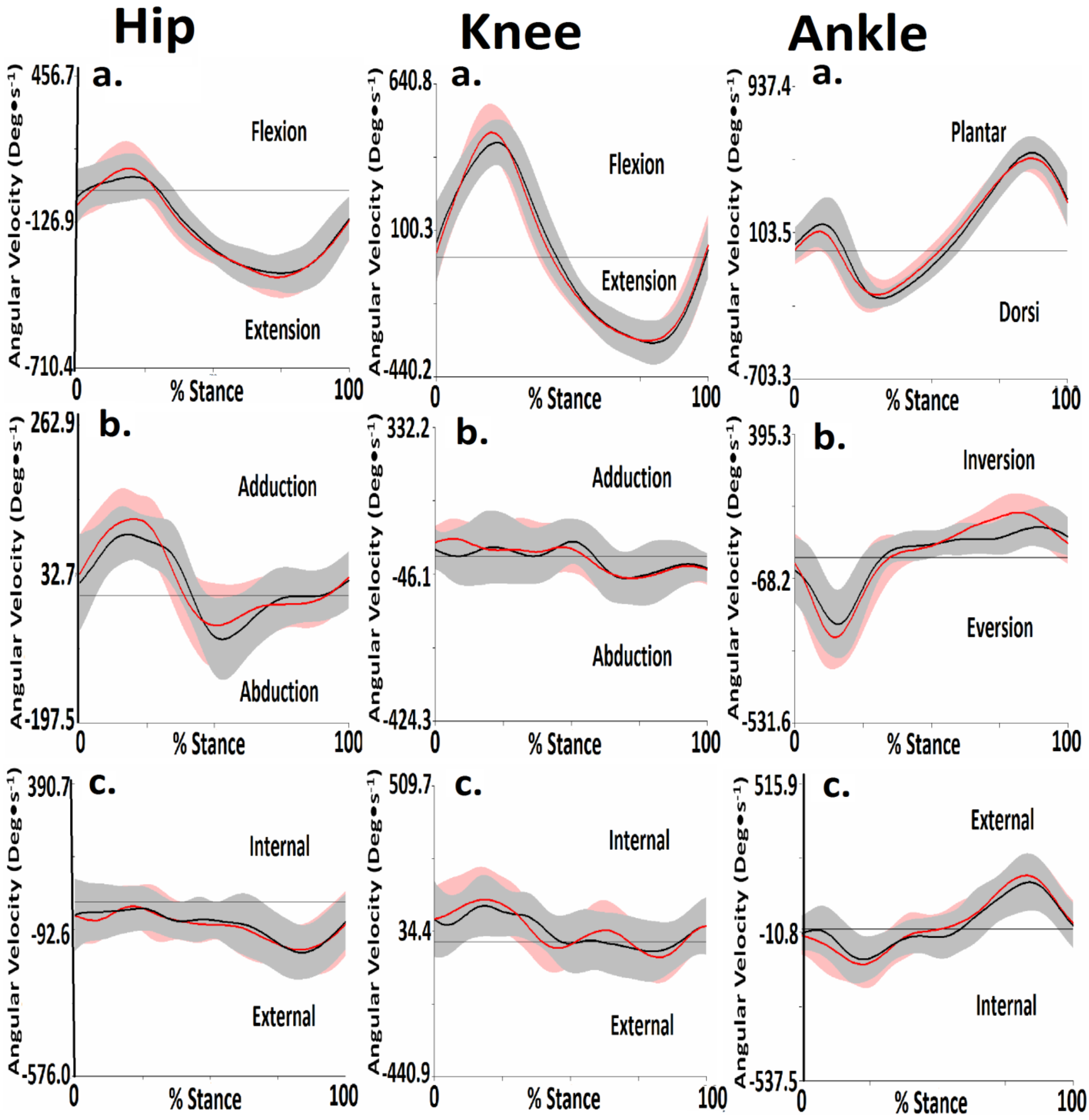

Figure 8.12: Mean and standard deviation hip, knee and ankle joint velocities in the a. sagittal, b. coronal and c. transverse planes the Saucony (black line) and Nike free (red line), footwear conditions (shaded area is 1 ±SD, black= Saucony and red = Nike free).

Table 8.13: Hip velocities (mean and standard deviation) from the stance limb as a function of footwear (* = Significant main effect).

	Saucony	**Nike Free**	
Hip			
X (+=flexion/-=extension)			
Velocity at FootStrike (°.s^{-1})	-22.63 ± 81.30	-77.74 ± 51.91	
Velocity at Toe-Off (°.s^{-1})	-116.42 ± 41.25	-125.49 ± 38.59	
Peak Flexion Velocity (°.s^{-1})	126.95 ± 80.96	148.08 ± 83.44	
Peak Extension Velocity (°.s^{-1})	-407.76 ± 80.16	-402.35 ± 67.98	
Y (+=adduction - =abduction)			
Velocity at FootStrike (°.s^{-1})	29.27 ± 54.35	36.19 ± 34.35	
Velocity at Toe-Off (°.s^{-1})	21.88 ± 27.21	18.32 ± 34.97	
Peak Adduction Velocity (°.s^{-1})	139.76 ± 31.02	153.92 ± 28.76	
Peak Abduction Velocity (°.s^{-1})	-81.76 ± 43.36	-94.03 ± 44.42	
Z (+=internal/- =external)			
Velocity at FootStrike (°.s^{-1})	-55.33 ± 86.25	-70.74 ± 70.01	
Velocity at Toe-Off (°.s^{-1})	-56.96 ± 75.12	-74.88 ± 83.46	
Peak Internal Rotation Velocity (°.s^{-1})	82.30 ± 43.94	76.91 ± 32.92	
Peak External Rotation Velocity (°.s^{-1})	-243.81 ± 69.67	-254.38 ± 58.22	

Table 8.14: Knee velocities (mean and standard deviation) from the stance limb as a function of footwear (* = Significant main effect).

	Saucony	**Nike Free**	
Knee			
X (+=flexion/-=extension)			
Velocity at FootStrike (°.s^{-1})	73.83 ± 85.44	23.01 ± 44.59	
Velocity at Toe-Off (°.s^{-1})	23.33 ± 32.14	48.21 ± 33.22	
Peak Flexion Velocity (°.s^{-1})	473.25 ± 53.32	491.27 ± 65.89	
Peak Extension Velocity (°.s^{-1})	-346.93 ± 57.27	-332.61 ± 35.22	
Y (+=adduction - =abduction)			
Velocity at FootStrike (°.s^{-1})	16.39 ± 38.96	36.18 ± 30.13	
Velocity at Toe-Off (°.s^{-1})	-21.73 ± 26.26	-24.15 ± 22.17	
Peak Adduction Velocity (°.s^{-1})	120.01 ± 42.36	128.80 ± 23.81	
Peak Abduction Velocity (°.s^{-1})	-138.79 ± 53.97	-119.04 ± 31.13	
Z (+=internal/- =external)			
Velocity at FootStrike (°.s^{-1})	90.57 ± 96.20	104.19 ± 103.11	
Velocity at Toe-Off (°.s^{-1})	33.18 ± 81.13	63.02 ± 59.68	
Peak Internal Rotation Velocity (°.s^{-1})	224.42 ± 41.81	257.85 ± 59.50	
Peak External Rotation Velocity (°.s^{-1})	-135.99 ± 37.26	-131.46 ± 35.07	

Table 8.15: Ankle velocities (mean and standard deviation) from the stance limb as a function of footwear (* = Significant main effect).

	Saucony	**Nike**	
Ankle			
X (+ =plantar/- =dorsi)			
Velocity at FootStrike (°.s^{-1})	35.67 ± 74.68	-5.19 ± 66.22	
Velocity at Toe-Off (°.s^{-1})	293.66 ± 70.46	274.72 ± 83.68	
Peak Plantarflexion Velocity (°.s^{-1})	568.35 ± 63.75	556.40 ± 37.59	
Peak Dorsiflexion Velocity (°.s^{-1})	-305.83 ± 34.88	-278.46 ± 31.77	
Y (+=inversion/ - =eversion)			
Velocity at FootStrike (°.s^{-1})	-58.28 ± 78.84	-13.67 ± 54.52	
Velocity at Toe-Off (°.s^{-1})	89.11 ± 71.46	58.11 ± 54.95	
Peak Inversion Velocity (°.s^{-1})	154.99 ± 65.86	207.48 ± 67.58	
Peak Eversion Velocity (°.s^{-1})	-303.85 ± 80.66	-348.20 ± 59.97	
Z (-=internal/ +=external)			
Velocity at FootStrike (°.s^{-1})	-21.37 ± 63.67	-27.80 ± 71.28	
Velocity at Toe-Off (°.s^{-1})	-18.67 ± 60.57	-6.18 ± 68.26	
Peak Internal Rotation Velocity (°.s^{-1})	232.96 ± 54.2	225.33 ± 44.55	
Peak External Rotation Velocity (°.s^{-1})	-196.91 ± 76.51	-219.66 ± 59.95	

8.5.4 Discussion

The aim of the current investigation was to examine the influence that footwear with different shock attenuating properties have on the energy requirements of distance running. This study also aimed to identify the kinetic, 3-D kinematic and electromyographical parameters that have strongest influence on running economy for each footwear condition.

The results of this study confirmed that steady state VO_2 did, not differ significantly (Saucony= 42.72 ± 2.17 $ml.kg.min^{-1}$ and Nike Free = 42.75 ± 1.95$ml.kg.min^{-1}$) between the two shoe conditions. Although, individual participants did exhibit differences of up to 2.75 % in oxygen consumption for the two footwear conditions. This concurs specifically with the observations of Nigg et al. (2003) who also found that VO_2 did not differ significantly between footwear conditions yet subject specific differences did occur. This observation however opposes the results of Frederick et al. (1986), Burkett et al. 1985 Hanson et al. (2010) and Squaradone et al. (2009) who found that different footwear cushioning systems can significantly reduce steady state VO_2.

However, in these investigations no correction was made for the variations in shoe mass that may have biased the results, which may explain the differences between studies as shoe mass has been shown to influence steady state energy expenditure (Frederick et al. 1986). It is further hypothesized that investigations that have observed differences between footwear may have compared footwear with extremes of midsole cushioning and that footwear within the normal commercially available confines of cushioning properties does not significantly influence steady stage energy expenditure.

That some participants were associated with lower VO_2 when wearing Saucony and others when wearing Nike Free is in line with theoretical predictions made earlier that footwear midsole material may, under certain conditions, influence the energy requirements for a given running velocity (Nigg and Anton, 1995). Taking into account the fact that the shoes utilized in the current investigation were selected arbitrarily in that the goal was to have one shoe with greater shock attenuating properties than the other, it could by hypothesized that differences in VO_2 may relate to the sole material being optimally tuned to each runner. The criteria for optimal participant/midsole tuning remains un-resolved at the current time.

The influence of the footwear conditions on VO_2 can be contextualized by taking the observed differences and examining their influence on performance as Hanson et al. (2010) did. Burkett et al. (1985) observed that every 1.0% increase in steady state energy requirements produced a subsequent 0.17 $km.h-^{1}$ reduction in running velocity. Therefore, the application of finding to the 0.09 % variation in VO_2 observed in the current study indicates that running in Saucony would allow athletes to run $0.015km.h^{-1}$ faster than in Nike Free. This translates to a 12 second reduction in marathon time and 2.85 second reduction in 10 km time, which is far less meaningful than Hanson et al. (2010) who observed potential marathon time reductions of 18 minutes.

The regression analysis revealed that peak hip flexion, knee relative ROM, activation of the vastus lateralis muscle and peak plantarflexion velocity were the best predictors of $VO_{2^{\cdot}}$ The fit of the multiple regression analysis for both footwear conditions ($R^2 \geq 0.81$) suggest that variance in running economy may be significantly influenced by running technique. This concurs with the findings of Cavanagh and Williams (1982), Williams (1986) and Williams

(1987) but opposes the observations of Kyrolainen (2001) who found that biomechanical parameters could not be related to running economy.

That peak hip flexion and knee relative ROM were found to influence VO_2 is to be expected as both of these parameters are likely to be associated with vertical oscillation of the body (Williams and Cavanagh, 1987). This concurs with the findings of Cavanagh (1977), Gregor and Kirkendall, (1978) and Saunders, (2004) who noted that increases vertical oscillation of the COM were associated with reduced running economy. Furthermore, that VM was also found to be a strong predictor of VO_2 is also predictable and in line with the previously described predictors of VO_2 given its role as concentric flexor of the hip and eccentric flexor and stabilizer of the knee joint (Kisner and Colby, 2002). This finding is in line with the observations of Martin and Morgan (1992) who noted that variables describing muscular effort have the greatest potential in determining metabolic energy demands.

That peak plantarflexion velocity also strongly influences running economy opposes the findings of Williams and Cavanagh (1987) who found that increases in plantarflexion were associated with reduced VO_2. This may relate to the differences in methods used as Williams and Cavanagh (1987) used a simple 2-D method as opposed to the 3-D technique in the current investigation. However, increases in plantarflexion velocity are likely to relate to greater mechanical work by the ankle plantarflexors, thus facilitating the increases in VO_2.

It is interesting to note that neither peak tibial acceleration or the median power frequency content of the tibial acceleration signal were found to significantly influence VO_2. This is interesting given that it is supposed that soft soled midsoles serve to improve running economy (Frederick et al. 1986). Thus, it appears based on the findings of the current investigation that the long held assumption that shock attenuating footwear are associated with reduced steady state VO_2 is incorrect.

The results of this study do however provide support for the concept of muscle tuning during dynamic activities as significant correlations were observed between muscle pre-activation and peak tibial accelerations. The muscle activation in the 50-100ms prior to impact is a preparatory mechanism for the expected impact transient generates joint stiffness and controls soft tissue vibrations (Nigg, 1997. The extent of the relationships between impact accelerations and EMG pre-activation are not as strong as those observed previously by Nigg et al. (2004) and Arndt et al. (2005). However, this may relate to the differences in methodological approach utilized in these studies in comparison to the current investigation as both studies used different running velocities and surfaces. Arndt et al. (2005) quantified tibial accelerations invasively using Hoffman bone pins mounted directly to the tibia. A mentioned previously this is considered to be a more direct technique of measuring tibial shock and thus may further explain the weaker relationships observed in the current study. Furthermore, Nigg et al. (2004) used wavelet transformation for their EMG data to examine the intensities of the signals prior to footstrike which is again in contrast to the current study.

The findings of this study show that changes in the heel material characteristics of running shoes are associated with subject specific changes in oxygen consumption however, that a significant proportion of the variance was explained by a small number of parameters; suggest that these parameters are clearly pertinent to running economy. Therefore, it is conceivable that runners may benefit from exposure to training techniques geared towards the modification of running mechanics specific to this study. Whilst the outcomes of previous analyses featuring biomechanical feedback to reduce energy expenditure have generally not been positive; future work should focus on implementing interventions to improve running economy.

9. Summary and recommendations for future research

9.1 Summary

This thesis was designed to meet a series of aims and objectives defined in the introductory chapter.

9.1.1 Summary of the contribution of this work to the literature

The research conducted during this thesis identified that:

- The ISB recommended XYZ cardan sequence of rotations is associated with the lowest amount of planar cross-talk from the sagittal plane when quantifying the 3-D kinematics of the lower extremities during running.

- The anatomical co-ordinate axes of the lower extremities can be defined consistently, ensuring that reliable and anatomically relevant kinematic data was obtained.

- The Bell et al. (1989) technique for the determination of hip joint centre was the most reliable technique of locating the hip joint centre.

- Gait events during running were accurately and consistently obtained in the absence of force platform information. Footstrike and toe-off events were identified with acceptable limits using kinematic data. Furthermore, this thesis also presents a new method of accurately defining gait events using a shank mounted accelerometer.

- Peak EMG obtained from ensemble average represents the most reliable technique for the normalization of EMG signals from the lower extremity muscles during running.

- Treadmill locomotion did not closely mimic overground locomotion. Thus, it is recommended that the treadmill is used cautiously in both clinical and research settings when it is being used as a substitute for overground analyses. Furthermore, on the basis that treadmill running was associated with increased rearfoot eversion/tibial internal rotation and overground runners exhibited greater peak tibial accelerations; it is recommended that runners consider their primary method of training when selecting the most appropriate footwear for their needs. Treadmill runners are likely to require footwear with additional medial stability properties, aimed at reducing rearfoot eversion whilst overground runners should consider footwear with more advanced midsole cushioning properties designed to reduce the magnitude of impact transients.

- From current knowledge females appear to be more susceptible to overuse injuries than males. Based on the findings of this investigation, with respect to shock attenuation; females do not require different footwear properties than males. However, on the basis that female runners were associated with increased rearfoot eversion/tibial internal rotation, it is recommended that females select running footwear with design characteristics aimed towards the reduction of coronal plane ankle eversion in order to reduce their incidence of injury.

- Barefoot running did not reduce the impact parameters associated with overuse injuries in comparison to conventional shod running. Furthermore, the barefoot inspired footwear tested in this study did not closely mimic the kinematics of barefoot running. Future work is still necessary to determine whether habitual barefoot running can reduce the incidence of chronic injuries.

- Footwear with different shock attenuating properties did not have a notable influence on steady state energy requirements of distance running. However, it was observed that kinetic, kinematic and electromyographic parameters of can be used to predict steady stage energy requirements. Finally, it was also shown that muscle activation in the period prior to footstrike was related to the magnitude of the impact transient experienced by the body.

9.1.2 Summary of footwear biomechanics research

In the late 1970's when footwear biomechanics emerged as a legitimate academic discipline in its own right, many fundamental questions were formulated regarding the relationship between footwear properties, performance and injury. Whilst significant progress has been made from addressing these questions, these early studies provide only the foundation of footwear biomechanics research.

Researchers commonly presume that additional knowledge from further studies will lead to a decreased frequency of injury (Novacheck, 1998). These promises are routinely un-quantified and incompletely substantiated. Nigg and Bobbert (1990) alluded to the fact that there is, as yet, no evidence that the vast amount of footwear biomechanics research has contributed to a decreased frequency of running injuries. The sport and clinical footwear fields of footwear biomechanics have until relatively recently been distinct bodies of knowledge with little interaction between disciplines. Similarly relationships between academia and industry have at times been strained, limiting communication and productivity. These divisions have restrained the pace of progress and thus it is recommended that additional work should be conducted aimed at reducing these divisions in order to simultaneously and uniformly advance the injury prevention and performance enhancement properties of running shoes.

Furthermore, whilst it appears logical that the running shoe has a role in the incidence and prevention of overuse injuries, the aetiology of injury is commonly unknown. Despite advances in footwear technology over the past forty years research is yet to confirm the extent to which footwear can reduce the risk of running injury. This is primarily because definitive studies would unethically increase the runner's risk of injury. Whilst many investigations have been conducted examining the aetiological factors that contribute to running injuries many of these have been retrospective in nature. Therefore more longitudinal prospective investigations are necessary whereby cause and effect relationships between kinetic and kinematics parameters and the incidence of specific injuries can be established. In direct evidence has been used effectively whereby running shoes have been shown to influence the hypothesized causes of injury (Shorten, 2000). Such evidence is only as good as the purported injury aetiology however, and more research is necessary in this area. In addition whilst different shoe constructions have been shown to influence running performance and economy, it remains uncertain as to the exact footwear properties that may affect performance.

9.1.3 Recommendations for additional research

Despite the popularity of the treadmill as an exercise and training modality, no epidemiological information for treadmill runners currently exists. Whilst this thesis provides information regarding the potential susceptibility of treadmill runners to overuse injuries, it is recommended that prospective epidemiological analyses be conducted for treadmill runners in order to establish the kinetic and kinematic mechanisms associated with the development of injury. Whilst prospective analyses are expensive they would be of substantial clinical significance given the number of runners who utilize the treadmill. Comparative analyses of treadmill and overground conditions are specific exclusively to the treadmill and surface conditions, and variations in these parameters would likely cause changes in runners movement strategy. Nonetheless, additional work should be conducted examining the effect of different treadmills on running gait mechanics in order to better understand the differences between the two conditions and make more appropriate footwear for the numerous commercially available treadmills.

Chapter 6 observed that females exhibit differences in 3-D kinematics that may predispose them to different overuse injury patterns to male runners. A number of structural and anatomical differences exist between males and females yet little is currently known regarding the underlying mechanisms behind the differences in running mechanics between male and female runners. Therefore it is recommended that future work be carried out investigating the mechanisms behind these variations in movement patterns between genders using correlational and co-variate analyses. This will provide insight into the distinct aetiology of the different injury patterns that may be observed between genders.

The findings of chapter 7 whereby non-habitual barefoot runners were found to be associated with increased impact parameters in comparison to shod differ from those who have examined habitual barefoot runners. It is therefore recommended that future investigations examine the differences between habitual and non-habitual barefoot runners and the duration over which the habituation period occurs. It is also recommended that future research examine the 3-D kinematics of habitual barefoot runners as the majority of work has examined only their impact kinetics. This may provide more insight into the mechanisms behind habitual barefoot runner's proposed reduction in susceptibility to injury. Finally whilst comparative analyses have been conducted which have examined barefoot inspired footwear and barefoot running itself, no investigations have contrasted the kinetics and 3-D kinematics of the numerous barefoot inspired shoe models against one another. Given the current popularity of these contemporary footwear, finding the most effective brand/model could have both practical and clinical benefits.

Whilst it is clear that certain shoe constructions can affect performance positively or negatively; this thesis did not find that cushioning properties had a noticeable influence on energy expenditure. Further research should therefore aim identify the mechanisms behind these effects in order to improve performance through footwear interventions. Furthermore, the concept of biological adaptation is a relatively new one in running literature. Whilst chapter 8 found that kinetic, kinematic and electromyographic parameters can have a large influence steady state energy expenditure. Whilst the regression analyses in chapter 8 indicates that a sizeable proportion of the variance may be explained by discrete parameters of peak hip flexion, vastus medialis activation, peak plantarflexion velocity and sagittal plane knee excursion; additional work should be conducted in order to determine additional factors that may contribute to steady state energy expenditure during running.

It is widely agreed that excessive impact loading can lead to musculoskeletal pathology, and that a loading window exists whereby the musculoskeletal system acts positively to these applied forces. However, it is not yet established the extent to which forces acting upon the lower extremity structures during running are within this window. The confines for an acceptable level of impact loading which is beneficial to the musculoskeletal system and avoids the overuse injuries associated with running represent a complex interaction between the magnitude and rate of the impact loading, duration of the loading cycles, and the amount of rest allowed between loading cycles. At the current time these complex interactions are not well researched or understood. Examining where typical impact loads during running fall in relation to the optimal window of loading is difficult to quantify due to the associated difficulties in measuring the stress/strain parameters of the musculoskeletal system using external measurements of vertical force loading, tibial acceleration parameters, or musculoskeletal modelling. Therefore, future research should focus more extensively on the magnitude and frequency components of the impact phase of running gait in vivo to determine how the impact phase specifically influences the musculoskeletal structures.

The application of forces to the musculoskeletal structures of the human body are clearly implicated in the aetiology of injury. Robbins and Gouw, (1990) determined that impulsive loading produces not only mechanical but also biochemical changes in the composition of body structures such as cartilage. Therefore, future work may wish to examine these changes using functional biological/biochemical markers to measure the reaction and adaptation of body structures to repeated impacts in sports such as running. This will improve the current knowledge regarding the aetiology of injury development.

It is also recommended that future research should pursue the concept of intelligent footwear. Footwear that adapts and understands the needs of the individual has been proposed a number of times in the literature but has yet to be fully investigated biomechanically. Footwear of this nature has been developed previously by Adidas in the form of their intelligence range. This footwear possesses a built-in 8- bit microcontroller sensor which measures heel compressions during running gait, and relays information to a motor which moderates a cushioning adaptation. To facilitate this cushioning adaptation, the shoe features a cushioning element, that is able to deform vertically via regulation by a motor-driven cable system which is can adjust the attenuation setting by turning a screw which lengthens or shortens the cable. These shoes were not successful commercially however as they were confounded by weight and cost issues and thus were discontinued quickly. Furthermore, the cushioning element despite being able to operate at 1000 Hz is significantly slower than the body's own mechanical alterations which are sensed and mediated proprioceptively via the CNS. Nonetheless, it is recommended that research and development work into this concept should be continued rather than being abandoned following one unsuccessful attempt, as it has been highlighted by some of the most preeminent researchers in this field of as being one of the principal avenues of research in the future of footwear biomechanics.

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Appendix A

Informed consent forms

Consent to be a research participant

Introduction

This study is conducted as part of a Ph.D project. Data collection will take place at the University of Central Lancashire in the Darwin building biomechanics laboratory (Room 018) and physiology laboratory (Room 026).

Procedure

Prior to commencement of the research, you will be asked to complete a PAR-Q form confirming that you are physically able to participate. You will be accepted for participation if you are injury free and able to run at speeds comparable to those of an average competitive distance runner. Upon being accepted for participation you will be scheduled to arrive at the Biomechanics laboratory Darwin building (018) on a designated data collection day. You will be advised to wear athletic attire and conduct an appropriate warm up.

You will be required to complete both treadmill and overground running. To acquire kinematic data retro-reflective markers will be placed on your body to define various anatomical positions. An accelerometer will be attached to lower part of your dominant leg. In order for a trial to be accepted you will be required to be within $\pm$ 5% of the required running speed. You will be allowed as much recovery time as is necessary between trials. You may also be asked to have an accelerometer mounted to your lower leg in order to measure tibial accelerations and electrodes positioned onto your muscles to measure muscle activation.

Risk

You will be required to run in your normal running style at a moderate velocity, thus risk of injury will be low. A risk assessment has been conducted by the university and thus risks associated with this study have been assessed and where possible controlled in order to keep them to a minimum.

Benefits

You may improve your ability to pace yourself appropriately at typical distance running speeds. Feedback regarding your running style may be provided upon request.

Data protection/Confidentiality

All data recorded during from the study will be saved on a laptop computer and will be used to generate means and standard deviations. However any information provided will remain confidential, and no information that could lead to the identification of any individual will be disclosed in any reports on the project, or to any other party (You will be identifiable by a number or letter only). No identifiable personal data will be published. The identifiable data (i.e. consent forms) will be stored by the supervisor in a locked filing cabinet and will not be shared with any other organisation. We may ask to take a photo of you (which will also not be identifiable) in order to give an example in the final report of the experimental set-up, if you are happy to do this then the option will be given at the end of this form.

Withdrawal

Participation in this study is completely voluntary. You may withdraw at anytime without being penalized or disadvantaged in anyway. However following completion of the data collection, data will be anonomized and thus can no longer be withdrawn.

Ethical Approval
This study has been approved by UCLan School of Psychology ethics committee.

Questions about the study
If you have questions about this study you may contact Dr Hobbs.
Email: sjhobbs1@uclan.ac.uk
Tel: 01772 893328

Or Jonathan Sinclair
Email: JSinclair1@uclan.ac.uk
Darwin Building 255
Centre for Applied Sport and Exercise Sciences
School of Psychology
University of Central Lancashire
PR1 2HE

I have read and understood the above data and give my consent to be a participant in this study.

Name: ....(please print)

Signature: ....……Date: ....

It is my understanding that ………………………….. understands the above project and gives her/his consent voluntarily

Name: ....(please print)

I give my permission for the student to include images of myself in his/her thesis.

Name...(please print)

Signature..Date:......................................

Consent to be a research participant

Introduction
This study is conducted as part of a Ph.D project. Data collection will take place at the University of Central Lancashire in the Darwin building biomechanics laboratory (Room 018).

Procedure
Prior to commencement of the research, you will be asked to complete a PAR-Q form confirming that you are physically able to participate. You will be accepted for participation if you are injury free and

able to run at speeds comparable to those of an average competitive distance runner. Upon being accepted for participation you will be scheduled to arrive at the Biomechanics laboratory (Darwin building 018) on a designated data collection day. You will be advised to athletic attire and conduct an appropriate warm up.

You will be required to complete five good trials running from one end of the lab to another striking the force platform with your dominant leg. To acquire kinematic data retro-reflective markers will be placed on your body to define various anatomical positions. An accelerometer will also be attached to the lower part of your dominant leg. In order for a trial to be accepted you will be required to cleanly strike the force platform and to be within $\pm$ 5% of the required running speed. You will be allowed as much recovery time as is necessary between trials.

Risk

You will be required to run in your normal running style at a moderate velocity, thus risk of injury will be low. A risk assessment has been conducted by the university and thus risks associated with this study have been assessed and where possible controlled in order to keep them to a minimum.

Benefits

You may improve your ability to pace yourself appropriately at typical distance running speeds. Feedback regarding your running style may be provided upon request.

Data protection/Confidentiality

All data recorded during from the study will be saved on a laptop computer and will be used to generate means and standard deviations. However any information provided will remain confidential, and no information that could lead to the identification of any individual will be disclosed in any reports on the project, or to any other party (You will be identifiable by a number or letter only). No identifiable personal data will be published. The identifiable (i.e. consent forms) data will be stored by the supervisor in a locked filing cabinet and will not be shared with any other organisation. We may ask to take a photo of you (which will also not be identifiable) in order to give an example in the final report of the experimental set-up, if you are happy to do this then the option will be given at the end of this form.

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Centre for Applied Sport and Exercise Sciences
School of Psychology
University of Central Lancashire
PR1 2HE

I have read and understood the above data and give my consent to be a participant in this study.

Name: ....(please print)

Signature: ....……Date: ....

It is my understanding that ………………………….. understands the above project and gives her/his consent voluntarily

Name: ....(please print)

Signature..Date:......................................

I give my permission for the student to include images of myself in his/her thesis.

Name..(please print)

Signature...Date:......................................

Appendix B

Ethical approval and risk assessment forms

Department of Psychology RISK ASSESSMENT FORM (Low risk, Student Version)

Use this form to risk-assess:

- ***Off-campus student activities (research, fieldwork, educational visits etc) in medium/high risk environments such as factories, farms, prisons, or remote areas.***
- ***All student activities involving medium/high risk procedures or use of specialist equipment.***

.

This form should be completed by the staff member responsible for the activity (e.g. the project supervisor), in consultation with the student and a qualified or otherwise competent person (normally a technician or Faculty HSE officer). Completed forms must be countersigned by the Head of Department or the Chair of the Department Health & Safety Committee.

Student:	**Assessment Undertaken By: (Staff member)**	**Assessment Verified By: (Technician or other competent person)**
Name: Jonathan Sinclair	Name: Dr. Sarah Jane Hobbs	Name:
Signed:	Signed:	Signed:
Date:	Date:	Date*:
Note: Risk Assessment is valid for* ***one year* *from the date given above. Risk Assessments for activities lasting longer than one year should be reviewed annually.*		
Countersigned by Head of Dept or Chair of H&S Committee:		
Date:		

Risk Assessment For:
Activity: Kinetics, and kinematics of footwear.
Location of Activity: Biomechanics Laboratory (Darwin building 18) and Physiology Laborotory (Darwin building 26).

List significant hazards here:	**List groups of people who are at risk:**	**List existing controls, or refer to safety procedures etc:**	**For risks which are not adequately controlled, list the action needed:**	**Remaining level of risk (high, medium or low):**
Dangerous of faulty facilities.	Researchers and participants.	Area should be well maintained and checked prior to the arrival of participants.		LOW
Medical conditions requiring medication.	Participants	Participants not medically able to compete will be recognised on completion of a par-Q form. Fit, participants should bring any necessary medication with them e.g. Inhalers. Participants who answer yes to any of the questions will be excluded		LOW
Inappropriate attire.	Participants	Participants will be instructed to wear appropriate clothing. Footwear will be provided by the researchers and is thus controlled for.		LOW
Fire	Researchers and participants.	Researchers and participants should be aware of where the nearest available fire exit is located. The exits should be kept clear. Researchers will be familiar with fire protocol.		LOW
Personal effects of the floor of the laboratory.	Researchers and participants	Ensure all belongings are placed by the wall, on the table or upstairs prior to the commencement of the testing.		LOW
Muscular Injury	Participants	Participants will		LOW

		run at only moderate velocities. Following completion of a suitable warm up.		
Slipping, tripping or falling during trials.	Participants	The laboratory floor will be kept clear in the area in which the participants will be running. It is also important to ensure that the floor is kept dry.		LOW
Injury or discomfort attributable to fatigue.	Participants	Participants will run for only 10-20m at moderate velocities and be allocated as much rest between trials as they need.		VERY LOW
Damaged equipment	Researchers and participants	All necessary equipment should have recently passed an appropriate safety inspection. PAT tested electrical equipment only, will be used. Any trailing cables etc will be taped down.		LOW
Medical Emergency	Researchers and participants	A first aid kit should be readily available as should a phone.		LOW
Tripping/stumbling on wires	Researchers and participants	All cables are under tables or kept as far away from the activity as possible..		LOW

SCHOOL OF PSYCHOLOGY ETHICS COMMITTEE
ETHICS FORM FOR
STAFF, MPhil/PhD & MSc RESEARCH PROJECTS

All researchers MUST obtain ethical approval BEFORE collecting any data.

Research Team

Researcher name(s) & email
Jonathan Sinclair
JSinclair1@uclan.ac.uk
Researcher type:

1. Supervisor name(s) & email (if applicable)
Dr Sarah Jane Hobbs
SJHobbs1@uclan.ac.uk

Project details (please see attached guidance notes)

2. What is the project title?
The influence of footwear on the kinetics, kinematics and metabolic aspects of distance running. ?
3. What is the likely duration of project?
Three years
4. Please provide a brief summary of the project aims (Max 250 words)

The aim of the project is to analyze the influence of footwear on the kinetics, kinematics and metabolic costs of distance running.

1.) Previous studies analyzing the kinematics of distance running have been conducted using the treadmill. Ronando and Squadrone (2000) suggest that the treadmill may be a more appropriate method for gait analysis. The convenience of the treadmill means that it is an appealing instrument for the analysis of human gait. It provides a well standardised and reproducible running environment, where speed and gradient can be easily defined and maintained, and the required order to calibration volume for capturing kinematic data is considerably reduced (Schache et al. 2001).
2.) However previous research concerning the kinematics of over-ground vs. treadmill running suggests that kinematic variations exist between the two methods. It must be acknowledged that little of this research has concerned out of plane variations between over ground and treadmill running, nor has it focused on the loading conditions being applied to the lower extremity that may facilitate the development of overuse injuries. The concept of this aspect of this project is to provide a 3-D kinematic and kinetic analysis of both conditions using the 3-D camera system and a shank mounted accelerometer, with the aim of determining whether a different shoe is necessary for the two running conditions.
3.) The vast majority of the research conducted to date on the biomechanics of running has involved male subjects. Research on female runners has been extremely limited. Consequently, very little information is available on the biomechanics of female distance runners. This paucity of scientific data combined with the likelihood of increased female participation in distance running suggest that research directed toward a better understanding of the biomechanical aspects would be of both theoretical as well as practical significance. A special issue are the specific demands of athletic footwear for women as compared to men's shoes. The majority of running shoes are designed with the same last rather than being specially adapted specially for females. The purpose of this aspect of the project is to provide both a kinetic and 3-D kinematic comparison of male and female runners using the 3-D camera system, acceleromter and force platform. Furthermore, various measurements of your feet and lower limbs will be obtained in order to assess foot shape differences between males and

females. The aim of this study is to determine whether or not current footwear models are adequate of whether female specific footwear needs to be developed more extensively.

4.) In recent years the concept of barefoot running has experienced a resurgence in footwear biomechanics literature. It is often hypothesized that barefoot running is associated with a substantially lower prevalence of chronic injuries, but well-designed studies of the effects of barefoot and shod running are lacking. In recent times shoes manufacturers have responded to this demand by creating footwear designed to simulate barefoot running. Shoes designed for this purpose are generally lightweight and intended to mimic the natural pressure patterns associated with barefoot running, thus allowing the feet to move through their natural (un-shod) ROM. The aim of this aspect of the project is to provide both a kinetic and 3-D kinematic comparison of running in cushioned running shoes, those designed to simulate barefoot running and barefoot running itself, using the 3-D camera system, in an attempt to determine whether i. barefoot running can reduce the hypothesized causes of overuse injuries in runners and ii. to what extent barefoot simulation shoes can mimic the kinetics, kinematics and natural pressure patterns associated with barefoot running.

5.) Impact forces act between the ground and the foot of a person during walking and running. Ground reaction forces typically have an impact peak after heel-strike due to the collision of the foot with the ground. The impact force occurs within 50 ms after first contact and causes shockwaves to travel through both the soft-tissues and skeletal components of the body. These impact forces are input signals into the locomotor system. The proposed concept of muscle tuning (Nigg, 1997) suggests the body reacts to different inputs to control the soft-tissue vibrations. Muscle activity in the lower extremities responds to the excitation frequency of the impact shock at heel strike. In addition to this the economy of locomotion is hypothesized to be the primary performance determinant in distance running. Although the surface EMG technique has been viewed as having a great deal of potential to explain several running-related biomechanical questions, little is known regarding its interaction with metabolic responses during distance running. The purpose of this experiment Is to measure changes in the muscle activity and soft- tissue vibrations which occur during. Running in different footwear conditions (including barefoot) in order to evaluate the muscle tuning hypothesis and to assess the influence of the variations in footwear conditions on oxygen consumption during heel–toe running. This aspect of the overall project will be conducted using a treadmill, oxygen consumption will be analyzed using the online digital spirometer. In addition to this muscular activity, transient accelerations and 3-D kinematics will also be obtained through the QTM software.

6.) It is likely that pilot investigations allied to the four main investigations (and using the same techniques and participants pool) will be required.

5. Please provide a brief summary of the project methods (Max 250 words).

A minimum of 10 healthy participants primarily between the ages of 18 and 30 will be recruited from the university student population for each aspect of the project. All of the data collection will take place in either the biomechanics lab (DB 018) or physiology lab (DB 026). On arrival to the Laboratory the participants will be required to fill in a Par-Q form to be checked for suitability to take part in the study. The participants will then undertake a 15 minute led warm-up in suitable Sports kit.

1.) To allow the quantification of impact forces the force platform and shank mounted accelerometer will be used. 3-D kinematics will be examined using the 3-D camera system. Timing gates will be used to control running velocity.

2.) Participants will complete this aspect in three different conditions, barefoot, Nike free running shoes and conventional running shoes. Participants will run across the force platform with a shank mounted accelerometer also attached in order to compare impact forces amongst the three footwear conditions. The 3-D camera system will be used to examine 3-D kinematics of the lower extremities.

3.) This aspect will be conducted using the treadmill and completed in both barefoot and shod conditions. Oxygen consumption and 3-D kinematics will be analyzed using the online spirometer and mobile 3-D camera system. In addition a shank mounted accelerometer will be used to measure the impact input to the musculo skeletal system and surface EMG will be used to quantify the lower extremity muscle activity in response to this impact.

6. Does the research involve contact with any other organisation or group (e.g. schools, companies, charities, hospitals, sports clubs)? If **yes**, please give details.

No

7. Is the research to be funded externally? If **yes**, please give details.

No

8. Will ethical approval for the proposed research be sought from any other body (e.g. collaborating departments, Home Office, health authority, education authority)? If **yes**, please give details.

No

9. Has a Risk Assessment form been completed?

Yes (See enclosed)

10. Has permission been obtained to use any copyright materials (e.g. personality tests)? Please also indicate whether particular qualifications or training are needed to administer the tests, and if so, whether the researcher is appropriately qualified.

None

11. Who do you propose to use as participants and do they belong to a group unable to provide informed consent?

In the main students from the university, lecturers and other members of staff may be recruited if necessary. All will be able to give informed consent.

12. Please indicate exactly how participants will be recruited for the project.

A minimum of 10 for each aspect.

13. How exactly will consent be given (e.g., verbal or written)?

Written consent via an informed consent form.(See enclosed)

14. What information will be provided at recruitment and briefing to ensure that consent is informed?

Participants will receive an information sheet.

15. Please indicate what information will be provided to participants at debrief.

Participants will be given the aims of the study prior to commencement as there is no deception at any point of the study. In addition participants will be given the hypothesis upon debriefing if required. They will also be allowed to see the data collected from their trials, and the results of the study (once the analysis is complete) upon request. A biomechanical interpretation/explanation will be provided at the participants request.

16. Please give details of any proposed rewards or incentives to be offered to encourage participation.

None

17. Is any deception involved? (If **yes**, please give details and explain why deception is necessary.)

No

18. Does the procedure involve **any** possible distress, discomfort or harm to participants? If so, what measures are in place to reduce it?

The participants will all be physically active and be performing movements at low intensity and will therefore be used to the type of exercise stress involved.

19. What mechanism is there for participants to withdraw from the investigation and how is this communicated to participants?

Participation will be completely voluntary. Participants will be free to withdraw at any time during the testing without reason, or fear of being penalized in any way. However following completion of the data collection, data will be annonomized and thus can no longer be withdrawn.

20. How are confidentiality and/or anonymity to be maintained?

The recorded data will be stored on a laptop computer and will be used to generate means and standard deviations. However the information provided will remain strictly confidential, participants will be identifiable by a number only. Identifiable data will not be shared with any other organisation.

Identifiable data e.g PAR-Q, informed consent form or medical questionnaire will be stored by the first supervisor in a locked cupboard.

Additional information

21. Please give details of any other ethical issues that have been considered.

Signed ……………………………..……………..……………………………………...…

(Signing this form certifies that you agree to carry out your research in the manner specified. If you want to deviate from the approved method at any time, you should seek further ethical approval for the change.)

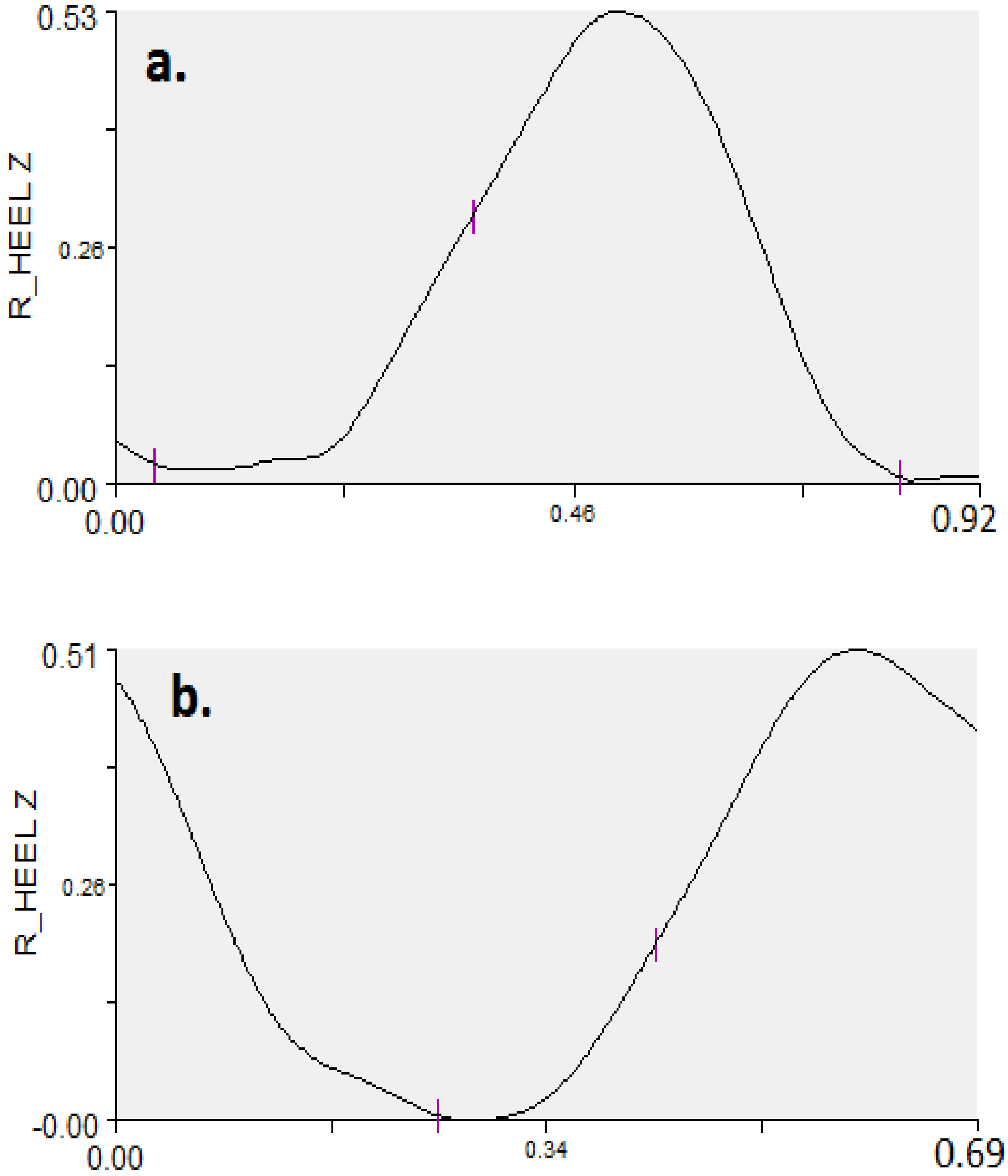
Date ……………………………………………………………………………………….

Supervisor signature (MSc projects only)……………………………………………

(**Note to supervisors**: Signing this form certifies that, in your opinion, the project specified here is ethical under Departmental and BPS guidelines. Do not sign if you are unsure, or if the student has not attached final versions of the research materials they are planning to use.)

Appendix C

Representative heel marker motion in the vertical axis during treadmill (a) and overground (b) running.



Appendix D

Results provided by Labosport regarding the shock attenuating properties of the treadmill and lab surfaces

LABOSPORT

Technical Report

Assessment of an Woodway™ Treadmil sports hall flooring and three types of trainers

in accordance with

EN 14808 & EN 14809

For

University of Central Lancashire

Summary

The Woodway™ Treadmill and indoor sports floor at the University of Central Lancashire has been tested in accordance with EN 14808 Surfaces for sports areas - Determination of shock absorption & EN 14809 Surfaces for sports areas - Determination of vertical deformation.

Unit 3, Heanor Gate Road, Heanor, Derbyshire, England, DE75 7RJ
Tel. +44 (0)1773 765007 Fax +44 (0)1773 765009
www.labosport.co.uk

1 Client

University of Central Lancashire
Darwin Building
Preston
Lancashire
PR1 2HE

2 Test details

Shock absorption EN 14808 and vertical deformation EN 14809 tests were carried out on a Woodway Treadmill, indoor sports floor and three brands of trainers and also on a indoor sports floor installed in Darwin 18 laboratory at the university of Central Lancashire.

3 Test programme

The Treadmill and indoor sports floor were tested for shock absorption and vertical deformation using the procedures detailed in EN 14808 Surfaces for sports areas – Determination of shock absorption & EN 14809 Surfaces for sports areas – Determination of vertical deformation.

4 Results

The results of the test programme are detailed below.

Woodway™ Treadmill

Property	Units	Result		
		Pos 1	Pos 2	Pos 3
Shock absorption	%	54	52	51
Vertical deformation	mm	2.76	3.12	3.08

Indoor Sports floor

Property	Units	Result		
		Pos 1 (on force plate)	Pos 2	Pos 3
Shock absorption	%	23	17	17
Vertical deformation	mm	0.33	0.22	0.27

Report details

Report prepared by	
Name	J R Blackburn - Laboratory Manager

Appendix E

Vo_2 % Differences between footwear

Nike Free	Saucony	Absolute difference	Mean of both	Individual % Difference
43.14	43.70	-0.56	43.42	-1.29
46.45	46.35	0.10	46.40	0.22
42.22	41.29	0.93	41.76	2.23
38.48	38.43	0.05	38.46	0.13
40.26	40.45	-0.19	40.36	-0.47
41.17	41.01	0.16	41.09	0.39
43.95	43.25	0.70	43.60	1.61
43.46	42.35	1.11	42.91	2.50
43.31	43.18	0.13	43.25	0.30
43.20	43.14	0.06	43.17	0.14
43.69	45.00	-1.31	44.35	-2.42
43.69	44.43	-0.74	44.06	-1.68

Nike Free	Saucony
42.75	42.72

MEAN OVERALL
42.73

Vo2 diff between	0.04
Diff / Overall mean	0.00
% Diff	**0.09**

Appendix F

Peer reviewed publications relating to this thesis