BIOMECHANICS OF FOOT FUNCTION IN RELATION TO SPORTS PERFORMANCE

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ABSTRACT

The foot forms the dynamic base upon which a sprinter functions. The actions that occur within the foot are of critical importance to the task of sprint running, since they influence the functional mechanisms of the entire body and especially the lower extremity. The aim of this research was to evaluate how foot function may contribute to sprinting performance and the interaction between the mechanical properties of sprinting footwear and performance, with a focus on the role of the metatarsophalangeal joint (MPJ). Currently, little is known about the effect of footwear upon the normal biomechanical function of the MPJ during sprinting, as this joint has often been neglected in previous biomechanical studies of lower limb energetics. A series of four empirical investigations were therefore undertaken to advance the understanding in this area.

The initial study revealed the importance of two important methodological issues on the analysis of MPJ function during sprinting. Appropriate MPJ axes representation and appropriate data processing procedures are vital to ensure the accurate assessment of joint kinetics. Empirical investigations on eight trained sprinters performing maximal sprint trials, both in barefoot and sprint spike conditions determined normal patterns of foot behaviour and the role of the MPJ during sprinting. Several aspects of foot function, including kinematic, kinetic and pressure characteristics, were determined. Sprint spikes reduced MPJ range of motion and dorsiflexion velocity but increased total energy generated during the push-off phase, biomechanical measures which may be linked to sprinting performance.

To investigate whether manipulations in the mechanical properties of sprinting footwear may influence sprinting performance and MPJ function, sprint spikes with insoles of varying stiffnesses were manufactured and mechanically tested. For a group of sprinters increasing the sprint spike stiffness did not elicit an improved sprinting performance. Due to the high variability between athletes and highly individualised responses to perturbations in footwear a single-subject analysis was undertaken. This study demonstrated that individual sprinting performance may be improved by implementation of relevant shoe mechanical characteristics. Whilst varying the mechanical characteristics of sprint spikes clearly showed controlling influences over the natural motion of the MPJ, the relatively minimal effect on the resultant MPJ energetics, potentially suggests that sprint spikes do not minimise energy loss during sprinting. This research highlighted several aspects of MPJ function which could be altered by footwear in an attempt to improve sprint running performance.

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GLOSSARY

Definitions of key terms used throughout the thesis

MPJ	Metatarsophalangeal Joint *:* The articulations between the five metatarsal bones of the foot and the proximal phalanges, including the hallux and lesser toes.
Performance	A measure of the success with which a particular task is accomplished
Technique	A specific movement strategy used to accomplish a particular task, including both kinematics (e.g. joint angles) and kinetics (e.g. resultant joint moments)
Stiffness	Longitudinal bending stiffness of the shoe midsole
Toe Spring	The upward curvature of the shoe sole at forepart, generally related to the stiffness of the sole so the more rigid the sole, the greater the toe spring that has to be provided
Midsole	The layer between the inner and the outer sole of a shoe
Centre of Pressure (CoP)	Location of the point of application of the resultant ground reaction force vector
Moment	Net muscle moment (effect of all acting forces) acting on a segment at the proximal joint
Moment arm	Perpendicular distance from an axis (MPJ) to the line of action of a force (CoP)
Energy	Ability to produce mechanical work performed by a joint
Power	Rate of mechanical work done by muscles, the product of the net muscle moment and angular velocity.

Inverse Dynamic Analysis	Method using link-segment bodies for computing forces and moments based on the kinematics and inertial properties of a body
Model	Anatomical axis system, defined by skin markers, a local reference system for an individual segment
Cut-off frequency	A frequency characterising a boundary in a signal's frequency response at which the signal is attenuated
Single subject approach	Experimental assessment of differences between conditions for an individual subject
Response pattern	Individual strategy for the performance of a task resulting in a unique pattern of movement
α	Probability level at which an effect is deemed significant
Coefficient of Variation	Measure used to describe the reliability of a data set (reproducibility of a measurement), calculated as the ratio of the standard deviation to the mean
Windlass mechanism	Tightening of the plantar fascia when the toes dorsiflex during gait

CHAPTER 1: INTRODUCTION

1.1 Research Overview

In the 2008 Beijing Olympics Usain Bolt won the 100 m title with a then world record of 9.69 s. The next three places were separated by just 0.04 s. Small margins are typical in elite 100 m sprint races with the winning margin in the Athens 2004 Olympics just 0.01 s. It appears that even small performance improvements can make a large meaningful difference in race outcome, therefore anything which can give a sprinter the leading edge can make the difference between medals won and lost. Hopkins, Hawley & Burke (1999) calculated that the minimum change in 100 m performance that would be meaningful (defined as one that resulted in a change in race position) could be as low as 0.3%.

Sprinting is one of the most powerful forms of human movement. Understanding the individual biomechanical factors, the kinematic and kinetic variables that are most important to sprinting, is vital to an improved performance. By improving the understanding of the joint kinetic and muscular contributions to sprint performance the potential for assessment and adaptation of sprint performance in an applied setting will likely be improved. Ultimately, athlete-specific analyses could provide detailed biomechanical feedback to a coach in a manner that could facilitate the development of specific technical training programs designed to improve sprint performance (Bezodis, Salo & Kerwin, 2009).

In sprinting, velocity is attained by the production of energy by the lower limbs (Johnson & Buckley, 2001). Researchers have examined the power produced by the hip, knee and ankle during sprinting, however energy absorption and generation at the foot has often been overlooked. The foot forms the dynamic base upon which a runner functions. In most sports, performance studies (such as maximal sprinting studies by Bezodis, Kerwin & Salo, 2008; Hunter, Marshall & McNair, 2004c; Jacobs & Van Ingen Schenau, 1992; Johnson & Buckley, 2001) the foot is modelled as a rigid, single segment. This may possibly be driven by the tendency to ignore the role of the foot in coaching practice; furthermore the single segment approach renders lower limb joint kinetic calculations easier to perform for researchers. However, potentially vital information may be missed and biomechanical knowledge of foot function in sports performance is limited. The actions that occur at the foot-shoe surface interface are of critical importance since they influence the functional mechanisms of the entire body and especially the lower extremity. Therefore, understanding how the foot functions during sprinting, for example, how loads are transmitted throughout the anatomical structures within the foot, and how the motion at the metatarsophalangeal joint (MPJ) influences propulsion during the push-off phase of stance, is of importance when investigating how energy is produced and used during sprinting.

One research group has reported kinematic and kinetic data for a multi-segment foot in sports performance (Krell & Stefanyshyn, 2006; Stefanyshyn & Fusco, 2004; Stefanyshyn & Nigg, 1997; 1998a; 1998b). They modelled the foot as two segments: rearfoot and phalanges, separated at the metatarsophalangeal joint (MPJ). In running, sprinting and jumping it was found that the MPJ absorbed large amounts of energy and produced minimal energy at take-off, with the toes remaining in a dorsiflexed position. No significant relationship between extension of the MPJ and sprint time for 100 m Olympic athletes was found (Krell & Stefanyshyn 2006). Therefore, the MPJ appears a large dissipater of energy and this loss appears to be detrimental to sprinting performance. If this energy loss could be reduced, the result may be a positive effect on sprinting performance. Stefanyshyn & Nigg, (1998b) demonstrated an average increase in vertical jump height of 1.4 cm and directly related this to a reduced amount (5 .4 J average reduction) of energy absorbed at the MPJ. Whether any similar reduction of energy would have a meaningful effect on sprinting performance is unknown. Furthermore, although this research group have reported the kinematic and kinetic characteristics of the MPJ during sprinting, their analysis of MPJ motion is however, somewhat limited by their methodological approach.

Unlike sports performance research, the investigation into foot function in clinical biomechanics research is well developed. In walking gait analysis, multi-segment foot modelling is advanced and most researchers use a three-segment model of the foot, with hindfoot, forefoot and hallux segments (Leardini, Benedetti, Catani, Simoncini, & Giannini, 1999; Carson, Harrington, Thompson, O'Connor, & Theologis, 2001; Stebbins, Harrington, Thompson, Zavatsky, & Theologis, 2006). More complicated models have also been proposed, such as the nine-segment model with medio-lateral divisions (MacWilliams, Cowley, & Nicholson, 2003). There is a large gap between the foot models used in clinical gait analysis and those used to evaluate sports performance. An appropriate multi-segment foot model is needed for the investigation of sports performance.

Whilst there is plethora of literature on the effects of footwear on running biomechanics, there remains little research exploring the influence of footwear characteristics upon sprinting performance. In the sprinting performance studies of Stefanyshyn and Fusco (2004) and Krell and Stefanyshyn (2006), it has been assumed that the motion of the running shoe is tightly coupled with the motion of the foot. Researchers have attempted to design running shoes that minimise energy loss at the foot and hence improve sports performance. Stefanyshyn and Nigg (2000) varied the midsole bending stiffness of running shoes and found that stiffer shoes minimised the loss of MP joint energy in running and jumping, but did not increase the energy generated at take-off. Stefanyshyn and Fusco (2004) found that stiffer shoes resulted in quicker 20 m sprint times. These results highlight the need for shod foot function in sprinting to be evaluated. Investigating the behaviour of the shod foot when manipulated by shoe modifications is necessary to improve our understanding of the role the foot plays in high speed sporting movements.

1.2 Statement of Purpose

The aim of this research was to evaluate how foot function may contribute to sprinting performance and the interaction between the mechanical properties of sprinting footwear and performance, with a focus on the role of the MPJ.

1.3 Research Questions

A series of research questions were developed in order to help achieve the stated aim and provide focus for the study.

i) **What influence do methodological issues have on the analysis of MPJ function during sprinting?**

Previous simplifications of the MPJ axes and the filtering of the kinematic data may misrepresent the joint kinetic information during sprinting. Past analyses of MPJ motion have used a single lateral marker to define the axes of the joint (Stefanyshyn & Nigg 1997; 1998b; 2000). This two-dimensional approach simplifies the motion analysis of the MPJ and does not reflect the changing or oblique nature of the MPJ axis. The moment arms necessary for joint kinetic calculations will be influenced by the definition of the joint axis used by investigators. In addition, during highly dynamic activities, those kinetic calculations will be influenced by the filtering of the segmental displacement data (Bisseling & Hof, 2006). It is not known whether typical measurement procedures, comprising of low video sample rates along with low cut-off frequencies for filtering kinematic data, can adequately capture the rapid motion of the foot during ground impact in high-speed activities such as sprinting. Potentially high speed aspects of joint behaviour could be omitted from the analysis.

In order to gain an accurate perspective of the motion and role of the MPJ, appropriate methods, including MPJ modelling, data capture and processing procedures needed to be developed. An appropriate representation of the joint, together with appropriate kinematic data sampling and filtering, enabled a better understanding of MPJ function during sprinting, leading to the assessment of the effect footwear has on kinematic and kinetic characteristics of MPJ motion during sprinting.

ii) How do sprint spikes affect the behaviour of the foot in terms of MPJ function and pressure distribution?

Toon, Williams, Hopkinson and Caine (2009) demonstrated that sprint spikes compromise angular range at the MPJ during maximal sprinting, compared to barefoot sprinting, therefore potentially affecting an athlete's energy generation ability during push off which may, in turn, potentially influence sprinting performance. A better understanding of natural foot behaviour and the loads experienced under the foot whilst sprinting is needed to determine the role and function of the MPJ during sprinting. In comparison to barefoot running, if sprint spikes have a marked effect on the energy produced and absorbed at the MPJ, it is likely that footwear may directly influence sprinting velocity. The outcome of this research question then led to the assessment of whether the mechanical properties of sprint spikes influence sprinting performance.

iii) **Can increasing sprint spike stiffness lead to significantly improved sprint performance?**

Various footwear characteristics have been proposed to influence performance, although some of these claims are without scientific evidence and may even be erroneous. Previous researchers (Stefanyshyn & Fusco, 2004) have suggested that a stiffer sprint spike can improve the sprint times of a group of elite sprinters. If this is true, then the effect of footwear properties alone may have the potential to improve an athlete's best sprint time. However, the interaction between sprint spike bending stiffness and sprinting performance remains unclear. If footwear has an influence on technique the athlete is using, this may have an effect on performance.

iv) **How does the stiffness of sprint spikes affect the biomechanical characteristics and function of the MPJ for individual athletes?**

In order to provide scientific evidence to evaluate the interaction between footwear characteristics and sprinting performance, it is necessary to explore the potential biomechanical mechanisms involved. A detailed biomechanical analysis of MPJ function, including the mechanical energy contribution of the joint and loading under the foot would provide an insight into the effect of sprint spike stiffness upon foot behaviour within the shoe. It is not known whether sprint spike properties directly influence the energetics of sprinting or indeed reduce energy loss within the foot, nor is it known whether any reduction in energy loss has a direct impact on performance. Footwear can influence the dynamic force production of muscles by manipulating the intrinsic musculo-skeletal characteristics of the force-length and force-velocity relationships, allowing greater production of force from a particular muscle group (Stefanyshyn & Nigg, 2007). A single subject research design permits individual responses to be explored, thereby individual athletes' foot functional and behavioural responses to different stiffness conditions can be assessed.

1.4 Organisation of Chapters

Chapter 2 presents a review of literature relating to the aims of the thesis. Included in this chapter are reviews of research on kinematic and kinetic variables shown to be related to sprinting performance. The chapter then focuses on the role of the foot and MPJ function to athletic tasks. Studies on running shoe design; in particular the relationship between shoe bending stiffness and running / sprinting performance are explored. Finally, traditional and contemporary methodological approaches are considered, along with the benefits and weaknesses of biomechanical data collection and analysis procedures.

Chapter 3 is concerned with methodological issues that affect the interpretation of MPJ function during sprinting and addressed research question i) What influence do methodological issues have on the analysis of MPJ function during sprinting? The investigation is comprised of two phases of testing, the first of which developed methodologies and data processing techniques for the collection of kinematic and kinetic data throughout the remainder of the thesis. In the second phase, an alternative representation of the MPJ was presented. Recommendations on measurement procedures for assessing MPJ function during sprinting were made. This chapter provided an initial insight into normal foot function and MPJ energetics whilst wearing sprint spikes.

Chapter 4 compares foot function during barefoot and sprint spike conditions and further explores the normal behaviour of the foot during sprinting, both barefoot and using a standardised sprint spike for a larger group of sprinters. This addressed research question ii) How do sprint spikes affect the behaviour of the foot in terms of MPJ function and pressure distribution? The study included eight trained athletes, who were analysed on a group basis to determine the effect of sprint spikes on MPJ function. An important section of this chapter was the inclusion of pressure measurement, to identify natural loading patterns under the foot and how these are affected by wearing sprint spikes.

Chapter 5 is concerned with the influence of sprint spike stiffness on sprinting performance and MPJ kinematics. This study was designed to identify whether increasing sprint spike stiffness can improve sprinting performance for a group of trained sprinters, thereby addressing research question iii) Can increasing sprint spike stiffness lead to significantly improved sprint performance? Carbon insoles were fabricated and placed into a standardised pair of sprint spikes to create the stiffness conditions. Mechanical testing of the stiffness of the sprint spikes plus insoles was performed and results compared to commercially available sprint spikes. Sprinting velocity, along with kinematic data was collected for a series of maximal sprint trials.

Chapter 6 presents an athlete-specific analysis for two single subjects, investigating the effect of shoe bending stiffness on sprinting performance, MPJ kinematics, kinetics and pressure distribution during sprinting. This addressed research question iv) How does the stiffness of sprint spikes affect the biomechanical characteristics and function of the MPJ for individual athletes? This study was performed to enable further insight into the biomechanical measures of the foot and MPJ which are influenced by shoe stiffness. In order to provide such a detailed analysis, a demanding single-subject protocol, with the use of a repeated baseline and multiple trials, was developed. This allowed for individual strategies to be explored for two sprinters, which is not possible in group designs, whereby group statistics may demonstrate limited insight into the fourth research question.

Chapter 7 presents a discussion of the major findings of the research, along with an appraisal of the methods used and demonstration of the insight that has been gained. The research questions established in Chapter 1 are addressed and implications of the results are discussed from research and footwear development perspectives. Limitations of the research are outlined, before potential directions for future research are discussed.

CHAPTER 2: REVIEW OF LITERATURE

2.1 Introduction

Whilst there has been extensive interest in the biomechanics of sprinting, the role and function of the foot has received relatively little attention. There is a plethora of research investigating the biomechanics of running shoes, yet there is limited information on running and sprinting spikes. This chapter reviews and evaluates the relevant existing biomechanical research within the areas of the biomechanics of sprinting performance, foot function, and shoe design. In addition, research pertaining to methodological considerations, relevant to the investigations in this thesis, are critically appraised.

2.2 Biomechanics of Sprinting performance

The three main biomechanical factors influencing sprint performance are sprinting kinematics, force and power production. Research has been conducted into the different phases of the sprint. Delecluse *et al.* (1995) suggested the 100 m sprint be divided into three phases. These are: early high acceleration over the first 10 m; the continuation of acceleration up to the attainment of maximum speed (10-36 m); and maintenance of maximal speed until the end of the 100 m. It is therefore important to recognize sprinting as a multi-phase event. To understand how sprinting velocity is attained and maintained, it is necessary to determine the biomechanical parameters involved in each of the sub-phases. Understanding of the associated movement patterns (joint and limb kinematic profiles) used in sprint running is essential in gaining a full insight into the technique developments required to enhance performance (Gittoes & Wilson, 2010).

2.2.1 Sprinting kinematics

In sprint running, the athlete's average speed is the product of step length and step frequency (Hay, 1993). Figure 2.1 demonstrates the basic kinematic factors involved during running.

Figure 2.1 Basic kinematic factors in running (Hay, 1993).

Step length and frequency are interrelated and dependent on morphological characteristics, duration of the contact phase, and force production in the braking and propulsive phases. Hunter, Marshall, and McNair (2004a) demonstrated a negative interaction between step length and step frequency for maximal sprinting; that is, the athletes who used longer step lengths tended to have a lower step frequency. Vertical velocity at take-off was deemed the most prominent source of this negative interaction; whilst vertical velocity of take-off had a positive effect on step length (via a greater flight distance) it also had a negative effect on step rate (via a greater flight time). It has been suggested that some athletes may either be stride length or stride frequency reliant (Salo, Bezodis, Batterham, & Kerwin, 2011). Typical sprinting velocities and stride characteristics have been well documented during the acceleration (Hunter *et al.*, 2004a, Johnson & Buckley, 2000) and maximal velocity phases of sprinting (Bezodis, Kerwin, & Salo, 2008; Mann & Herman, 1985) with typical maximal velocities around 9 – 10·5 $m.s^{-1}$, step lengths around 2 – 2.4 m and step frequencies around 4 – 5 Hz. Contact times in absolute maximal sprinting are reported to be very low: 0.080 to 0.120 s (Bushnell & Hunter, 2008; Mero, Komi, & Gregor, 1992).

The kinematics of the stance leg are believed to be performance-determining factors in sprint running (e.g. Hunter *et al.*, 2004c) The angle of the thigh in stance and the shank at touchdown were considered critical determinants of elite athlete performance differences (Mann & Herman, 1985). Hip and knee flexion-extension velocities have been considered to be particularly influential in generating large forward propulsion required in the acceleration phase of sprinting (Hunter *et al.*, 2004c). Krell and Stefanyshyn (2006) highlighted that plantarflexion of the ankle and corresponding knee extension just before take-off are also vital to the generation of sprinting velocity. Gittoes and Wilson (2010) highlighted the role of knee-ankle coupling to movement patterns during maximal velocity sprinting and demonstrated that well trained sprinters use a more reproducible knee-ankle coupling.

2.2.2 Ground reaction forces and sprinting performance

The most critical issue in sprint running is the action of the leg before and while it is contact with the ground (Kunz & Kaufmann, 1981). The leg action during this period is related to the amplitude and patterns of the ground reaction force (GRF), which in turn determines the acceleration of the athlete's centre of mass. The anterior-posterior and vertical components of the GRF are of most interest, with these components increasing with speed (Weyand, Sternlight, Bellizzi, & Wright, 2000). In order to improve sprinting performance, sprinters should minimise the horizontal braking GRF, which acts posteriorly and usually occurs early in the stance phase, and maximise the horizontal propulsive GRF, which acts anteriorly and usually occurs later in the stance phase (Mero *et al.,* 1992).

Hunter, Marshall, and McNair (2005) also concluded that high magnitudes of propulsion are required to achieve high acceleration in sprinting. For data collected at the 16 m point, the strongest predictor of sprint velocity (accounting for 61% of the variance) was the relative horizontal propulsive impulse. GRF impulse is an informative measure as it reflects the total force applied in a certain direction over the duration of stance phase, and also because, when expressed relative to body mass, it reflects the change in velocity of the athlete.

Maximising forward propulsion seems to require optimal force application, by improving body positioning with greater forward lean (via utilizing a more posterior foot plant or a longer ground contact time), as opposed to maximising force application (Kugler & Janshen, 2010). Morin, Edouard and Samazino (2011) showed that the orientation of the total force applied onto the ground during sprint acceleration is more important to performance than its amount, although this was based on an instrumented treadmill; therefore maximal sprinting speeds were quite low, with belt speeds of 6.8 to 7.3 $m.s^{-1}$.

2.2.3 Joint kinetics and sprinting performance

The study of joint kinetics can improve the understanding of the underlying causes of a movement (Winter, 2005). Joint moments and powers can provide an estimation of the net summation of all muscle activity of the joint and determine the action of the muscle groups crossing a joint. Mechanical power is an important biomechanical parameter when analysing performance of human gait as it describes the energy flow between lower limb segments. The inverse dynamics (ID) method (Winter, 2005) is therefore important in specifying joint kinetics and muscle function (Belli, Kyrolainen, & Komi, 2002).

In sprinting, where the goal is to cover the allotted distance in the least possible time, power production in the leg muscles is crucial to high performance. Joint energy production and muscle power patterns are particularly important contributors to the efficiency of the early acceleration phase. Investigating the kinetics of sprinting provides an insight into the muscles involved in running and the loads experienced at the sprinter's joints (McClay & Manal, 1999). Resultant joint moments and muscle powers are vital biomechanical parameters. The conventional model used for an ID analysis of the lower limbs is a 3 segment rigid body model, whereby resultant 2D sagittal plane joint moments and powers are reported for the hip, knee and ankle joints (e.g. Mann, 1981; Jacobs & van Ingen Schenau, 1992; Johnson & Buckley, 2001; Hunter, Marshall & McNair, 2004c; Bezodis *et al.,* 2008). Resultant joint powers can be used to determine which muscle groups are working concentrically or eccentrically around a joint (Johnson & Buckley, 2001).

It is well documented that rapid extension of the hip, knee and ankle, and production of power by the joints of the lower limb, are the main source of propulsion in sprinting (Johnson & Buckley, 2001). High hip joint extensor moments, in particular, have been associated with increased propulsion (Simpson & Bates, 1990), as well as the magnitude of positive work performed at the ankle joint (Bezodis *et al.*, 2008). The hip extensor theory (Mann & Sprague, 1980) states that the hip extensors are the major determinant of thigh acceleration during stance, which contribute to the generation of the propulsive GRF.

Jacobs and Van Ingen Schenau (1992) investigated intermuscular coordination in a sprint push off and reported joint moments and muscle powers for the second ground contact out of the blocks. They stated that, to increase the horizontal velocity of the centre of gravity in the first steps of sprinting, an interaction between rotation around the foot and leg extension is necessary. The athletes performed the sprint in a stereotyped manner with the strategy to wait with leg extension and rotate around the contact foot first to conserve angular momentum. This is due to the specific requirement of dealing with the location of the ground reaction force vector, which changes continuously with respect to the body's centre of mass. Peak power at the hip, knee and ankle occurred in a proximal-to-distal sequence, reflecting the timing of the monoarticular muscles in delivering positive power. The action of the biarticular muscles contributed to the distribution of power to the adjacent joints, where the power can be applied more effectively. These findings were comparable to lower limb action in the vertical jump (Bobbert & Van Ingen Schenau, 1988); however, in sprinting, the biarticular hamstrings and rectus femoris had a more pronounced pattern of reciprocal activity. This is due to the specific constraint of controlling the direction of the ground reaction force in order to the conserve angular momentum of the system.

Jacobs, Bobbert and Van Ingen Schenau (1993) stated the results of their study in running fitted well in the global functional difference between monoarticular and biarticular muscles, that is monoarticular muscles appear to be active predominantly where they can contribute to positive work, whereas biarticular muscles appear to be responsible for the regulation of net moments about the joint (Jacobs & Van Ingen Schenau, 1992). They investigated muscle function during stretch-shortening cycles (successive occurrence of lengthening and shortening) in sub-maximal fast running in the mid-acceleration phase of sprinting, at the 20 m mark. Compared to the maximal sprint push off, the mechanical output of the ankle was higher. Explanations for this difference may lie in muscle stimulation; in the sprint push off the plantar flexors are not fully activated due to the conflict of conserving angular momentum described above. Jacobs, Bobbert and Van Ingen Schenau (1996) provided additional evidence for the hypothesis that the work done by the large monoarticular muscles spanning proximal joints is transferred to distal joints by the action of biarticular muscles during the sprint push off (second stance phase). This causes an effective conversion of body segment rotations into the desired translation of the body centre of gravity.

Johnson and Buckley (2001) determined the net muscle action of the ankle, knee and hip during the mid-acceleration phase of sprinting, analysing a stride 14 m into the sprint. Their results also showed a proximal to distal timing in the generation of peak extensor power during stance. Peak extensor moments were -377 ± 34 N·m for the hip, 231 ± 90 N·m for the knee and -328 ± 89 N·m for the ankle The ankle plantarflexors demonstrated high power generation in late stance (3066 ± 846 W), possibly the result of power generated at the hip in early stance (3242 ± 1086 W) transferred by biarticular muscles.

Hunter, Marshall, and McNair (2004c) provided further insight into how speed is generated in mid-acceleration phase sprinting. They analysed a stride 16 m into the sprint using segment-interaction analysis, and their results agreed with reports of a proximal to distal sequence. They also noted a large propulsive ground reaction force (GRF) was produced at approximately the same time as peak hip and knee extension velocities, thus knee and hip flexion-extension angular velocities have been considered to be particularly influential in generating propulsive GRFs during sprint acceleration. Hunter *et* *al.* (2005) also reported that, in the mid-acceleration phase, faster athletes produced high magnitudes of horizontal propulsion and only moderate magnitudes of relative vertical impulse, as strength reserves should be directed horizontally to achieve high acceleration.

During maximal sprinting, Bezodis *et al.* (2008) demonstrated that the hip and ankle extensors produced most work during early and late stance respectively. Video and force data was collected at the 40 m point of a 60 m maximal sprint for four well-trained sprinters. Power production for the hip, knee and ankle is demonstrated in Figure 2.2. For the hip, there was a power generating extensor action immediately after touchdown and a second peak at approximately 50% of stance. The knee did not play a substantial power producing role. The ankle underwent a power absorbing peak followed by a power generation peak during late stance. The action of the knee joint was a facilitator for the radial transfer of power from the hip to the ankle. This sequence of power production agrees with that presented by Johnson and Buckley (2001) for accelerated sprinting. However, for maximal sprinting some differences were highlighted: a reduced propulsive role of the knee joint (disagreeing with a previously reported powerful knee extension phase before take-off). For maximal sprinting, Bezodis *et al.* (2008) also demonstrated considerable ankle plantar flexor power was dissipated, possibly due to the high level of athletes used or the phase of the sprint investigated.

Figure 2.2 Power production over time during the support phase at the hip, knee and ankle, one typical maximal sprinting trial from Bezodis *et al.* (2008).

All of the aforementioned studies have analysed the contribution of the stance limb during sprint running using Inverse Dynamics Analysis (IDA). Hunter *et al.* (2004c), however, looked at the segment interaction involved, incorporating the resultant joint moment at the distal end of the segment and all the other resultant joint moments of the linked system, and thereby quantified interactive moments for the stance limb segments. They confirmed the hip extensor moment as the major determinant of increasing the angular velocity of the thigh (agreeing with Belli *et al.,* 2002), but the extensor moment at the ankle was attributed with preventing the collapse of the shank under the effects of the interactive moment due to ground reaction force. However, the as the first documented use of the segment interaction analysis method to a closed linked system, their results should be used only as a guide, until further segment interaction analyses are performed on more athletes, using a more intricate model and highly accurate joint centre locations.

Overall, it has been shown that a proximal to distal sequence of muscle power generation and transfer of power occurs throughout sprinting. The dominant action of the hip extensors during early stance has been shown to be a crucial variable to sprinting performance. The knee arguably produces relatively low values of moment and power in maximal sprinting and, therefore, may take on a compensatory role (Bezodis *et al.,* 2008), although this is not consistent within the research and depends on the phase of the sprint investigated. Together, the segment interaction between the large torques produced at the hip and knee during the first part of stance, in particular, place high demand on the hamstring muscles and may lead to the occurrence of hamstring injury (Liu, Wei, Zhong, Qing, & Fu, 2009). The ankle dissipates energy during stance but produces a powerful extension at take-off, again crucial to sprinting performance (Johnson & Buckley, 2001). Stefanysyhn and Nigg (1998b) compared the biphasic nature of the ankle joint absorbing and producing energy to a spring being compressed and allowed to extend, with the stiffness of the ankle a specialized characteristic of the activity or demand placed upon it. Ankle joint stiffness was reported to be consistent throughout the entire range of motion and corresponding changes in the resultant joint motion during sprinting. As Butler, Crowell & Davis (2003) concluded that it appears that increased stiffness is beneficial to performance, if ankle stiffness indeed influences performance, training the ankle to achieve higher stiffness could possibly lead to better performances. In current coaching practice, the action of the ankle in generating sprinting velocity is recognised and exercises such as ankle bouncing and running rebounds are advocated, as part of sprint training (Donati, 1996).

2.2.4 Musculo-skeletal geometry and sprinting performance

As sprint running induces stretch-shortening cycle activity in the muscle-tendon complex of the lower limbs, certain morphological and mechanical properties of muscle and tendon may be related to sprint performance. The capacity of a muscle-tendon system to store strain energy when stretched eccentrically depends on the size and tension of the muscle and on the length and compliance of the tendon. Longer muscle fascicles produce more power at a higher shortening velocity, increasing the rate at which force can be applied to the ground, thereby contributing to better sprinting ability (Karamanidis, Albracht, Braunstrein, Catala, Goldmann & Bruggemann, 2011). Although longer muscle fascicles have been observed in the calf musculature of human sprinters when compared to distance runners (Abe, Kumagai & Brechue, 2000), triceps surae muscle fascicle length is not strongly linked to performance (Stafilidis & Arampatzis, 2007). Long plantarflexor moment arms increase the mechanical advantage for joint moment development, but will shorten more rapidly and produce less force owing to the force-velocity property of muscle (Baxter, Novack, Van Werkhoven, Pennell & Piazza, 2011). Recently, both Lee & Piazza (2009) and Baxter *et al.* (2011) found sprinters to have shorter plantarflexor moment arms and longer toes / forefoot bones (increasing the lever arm of the GRF and raising the gear ratio) than non-sprinters. Conversely, Karamandis *et al.* (2011) found no significant differences in plantarflexor moment arms, toes lengths and midfoot lengths between elite and slower sprinters and therefore suggest differences in sprint ability in world class athletes are not a result of differences in lower leg musculoskeletal geometry.

Muscular strength and the proportion of fast-twitch fibres to slow-twitch fibres are correlated with maximal running velocity (Mero, Luhtanen, Viitasalo & Komi, 1981). As it is assumed that high ankle plantar flexor strength is beneficial to sprinting performance, this could lead to the suggestion that strength of the toe flexor muscles may be another morphological factor that could influence sprinting performance (Potthast, Niehoff, Braunstein, Goldmann, Heinrich & Bruggemann, 2005). There is no known evidence to support this notion.

2.2.5 Summary

There have been numerous biomechanical investigations of sprint running and the joint kinematics, kinetics, and ground reaction forces of the movement have been documented. The importance of joint kinetic factors to sprinting performance has been presented for the start, acceleration and maximal sprinting phases. The significance of these variables to improving sprint performance warrants further clarification. Individual level detailed biomechanical analyses and feedback is available to athletes (such as Bezodis *et al.*, 2009) which has the potential to improve sporting performance.

2.3 Foot function and the role of the MPJ during walking, running and sprinting.

Various researchers have studied mechanical energy production and dissipation in the lower extremities during athletic tasks. However, the MPJ has consistently been omitted from such studies. In running and sprinting, large forces are encountered at the MPJ as well as large rotations; therefore, the MPJ may play an important role in lower limb power production in such athletic activities. Inclusion of the MPJ in joint kinetic analysis extends the conventional 3 segment rigid body typically used for ID analysis, by recognising the foot as more than one segment. Bezodis, Salo and Trewartha (In press) demonstrated the extent to which this inclusion would affect the calculated kinetics at the other joints in the leg during sprint running. They found that by ignoring the MPJ moment, peak extensor moments at the ankle, knee and hip were 35% higher, 40% lower and 9% higher respectively (all significant) than those calculated with inclusion of the MPJ. Therefore, the resultant moments at the stance leg are inaccurate (in particular, artificially high peak ankle joint moments are calculated) when the MPJ plantarflexor moments are ignored.

2.3.1 The function of the MPJ during walking

It is recognised that during walking, dorsiflexion of the toes plays an important role in push off phase in gait. Bojsen-Moller and Lamoroux, (1979) were one of the first to research the area. They stated that compared to a rigid lever, the foot with its intermediate break at the MPJ has several advantages for smooth accomplishment of the vertical and horizontal accelerations that are necessary for the initiation of the swing phase of walking. These are: a) the resistance arm (distance from ankle to MPJ) of the foot diminishes during early stance, reducing the demands upon the triceps surae; b) the resistance arm of the foot then increases with greater horizontal speed, allowing the triceps surae to provide useful forces over a longer period of time; c) the flexor hallucis longus is stretched, reaching a higher tension and providing force for push off. There are two metatarsophalangeal axes (Figure 2.3): one transverse passing through metatarsal heads (MTH) 1 and MTH2, and an oblique axis through MTH2 to MTH5, with the oblique axis primarily used during walking (Bojsen-Moller, 1978).

Free dorsiflexion at the MPJ is essential for the function of the foot during gait, not only for the mechanics of the forefoot during push off but also for arch support and the windlass action, i.e. when the toes are dorsiflexed, a pull is exerted on the plantar aponeurosis, whereby the plantar aponeurosis under the medial longitudinal arch is wrapped around the metatarsal heads like a windlass to pull the calcaneus towards the metatarsals (Hicks, 1954; Mann & Hagy, 1979). This raises the longitudinal arches of the foot when the toes are dorsiflexed (Bolgla & Malone, 2004) and forces the transverse tarsal joint into a forced flexed position, creating a solid structural support (Dugan & Bhat, 2005).

Figure 2.3 Axes of the right foot. Taken from Bojsen-Moller and Lamoureux (1979, p474):

Diagram of right foot showing the location of the axes. During push off the two mechanical axes of the ankle joint complex (tc and st) combine into axes (C_{tr} and C_{obl}) which are parallel in the horizontal plane with the primary axes (B_{tr} and B_{obl}) at the metatarsophalangeal level. The push off can thus be performed about a set of transverse axes or a set of oblique axes. With the transverse axes the foot is a divided lever with C_{tr}-B_{tr} as the resistance arm in the first phase of the push off and C_{tr}-A_{tr} as the resistance arm in the final phase (compare with Figure 3). With the oblique axes the resistance arm C_{obl}-B_{obl} is shorter and there is no advanced axis.
tc: mechanical axis of the talocrural joint.
st: mechanical axis of the subtalar joint.

Scott and Winter (1993) demonstrated that during walking the force under the hallux increases to a peak at about 80 – 85 % stance, with loads ranging from 95 N to 260 N. The first MPJ extends to approximately 95% of stance then rapidly flexes in the last 5% of stance. However the authors did not include the other four toes as they were assumed to 'provide minimal load support during stance' (Scott & Winter, 1993, p.1093). The first MPJ moments exhibited depended largely on the load under the foot which varied considerably between subjects.

During walking, the variable ground reaction forces resulting from gravity and body segment accelerations are applied to the plantar surface of the foot and toes. The toes do not bear significant loads during contact and midstance of walking, but during propulsion the metatarsal heads and distal phalanges are the only points of contact with the ground (Rolian, Lieberman, Hamill, Scott & Werbel, 2009). Hayafune, Hayafune and Jacob (1999) reported typical pressure distributions under the normal foot during the push off phase in walking and demonstrated that the load is mainly shared between MTH1, MTH2 and the hallux, with these structures taking 64% of the total forefoot load during push off. Overall it is clear that the MPJ has an important role in the push off during walking and, therefore, free mobility of the five MPJs is important (Bosjen-Moller & Lamoreux, 1979).

2.3.2 Foot function in athletic tasks

As with walking, the push off phase is vital to effective propulsion during athletic activities, including running and sprinting. As well as supporting the body and providing traction, the toes help to control the forward motion of the centre of mass during propulsion, with the digital flexors assisting the more powerful ankle plantarflexors in generating lift, particularly during running and sprinting (Rolian *et al.,* 2009). Samazino *et al.* (2009) highlighted the spring-like behaviour of the arch of foot during stance, allowing the foot to flatten and absorb shock during midstance of running then later returning energy by elastic recoil. Bojsen-Moller (1978) stated that, for high speed locomotion, the push off phase is performed around the transverse axis of the MPJ, the so called 'high gear' of the two-speed construction of the human foot.

During barefoot running it has been shown that runners often adopt a flatter foot placement at touch-down to limit local pressures under the heel as well as an increased external vertical loading rate and earlier impact peak (De Wit, De Clerq, & Aerts, 2000). Higher tibial accelerations have also been reported for barefoot running (McNair & Marshall, 1994) as well as higher ankle joint stiffness and lower knee joint stiffness (Coyles, Lake & Lees, 2001). Finally, changes in knee kinematics (De Wit & De Clerq, 2000), foot eversion and tibial rotation (Stacoff, Nigg, Reinschmidt, van den Bogert & Lundberg, 2000), and earlier maximal tibialis anterior EMG activity (von Tscharner, Goepfert & Nigg, 2003), have been reported for barefoot conditions. Barefoot running has, in fact, received much attention in recent research with the Nature publication of Lieberman *et al.* (2010) who stated that habitually barefoot runners often land with a forefoot or midfoot strike which may protect the feet and lower limbs from some impact related injuries. Whilst there is no question that the kinematics and kinetics for barefoot running are different when compared to shod running, there is little evidence to demonstrate a reduction in impact forces, loading rates, nor injury rates in barefoot runners. There is also no known published evidence for the effects of running barefoot on simulated or real competitive performance.

De Cock, De Clercq, Willems, and Witvrouw (2005) demonstrated the temporal characteristics of foot roll-over during barefoot jogging in young adults. Foot roll-over during jogging started with heel contact. After heel off, the forefoot started to push off at the lateral metatarsals, followed by a more central push off over the second metatarsals and, finally, over the hallux. This pattern differs slightly to the pattern suggested for running with shoes, in which increasing medial forefoot loads were followed by loading almost entirely carried by MTH1 and hallux during push-off (Henning & Milan, 1995). However, compared to walking, Mann and Hagy (1979) found that the toe muscles are much more active during running and are vital in assisting with forward propulsion of the body.

There is very limited research regarding loading patterns during sprint running. Eils *et al.* (2001) demonstrated that during sprinting in football boots, the predominant loading areas were found in the forefoot (medial forefoot and hallux, central forefoot and second toe), with highest in-shoe pressures evident at the medial forefoot, and significant increases in peak pressure under the first and second toe, when compared to running in running shoes. Queen, Haynes, Hardaker, and Garrett (2007) also found high peak pressures on the medial and central forefoot and hallux during a football acceleration task. Fourchet, Kuitunen, Dingerkus and Millet (2007) demonstrated that plantar loading under the midfoot and forefoot was substantially increased wearing sprint spikes, compared to running shoes, for sprint running in young athletes, with higher loads under MTH1 and MTH2, in particular. Whilst, there is some evidence for high loading in the medial forefoot during sprinting, there is currently no known pressure data to describe the typical loading patterns of trained sprinters wearing sprint spikes.

Overall, it is clear that the MPJ has a major role during the push-off phase in running (Mann & Hagy, 1979). This has also been found for jumping activities. Dozzi, Winter, and Ishac (1989) investigated the MPJ power required during jumping in ballet and found that the muscles near the MPJ provide great power relative to the joint cross-sectional. The percentage of combined total work done at the ankle and MPJ was 44% by the MPJ flexors versus 56% by the ankle plantarflexors.

2.3.3 The function of the MPJ during running, sprinting and jumping

Stefanyshn and Nigg (1997, 1998a) and Krell and Stefanyshyn (2006) are the main contributors to the research field of MPJ kinematics and energetics in athletic activities and have investigated the role of the MPJ during running, sprinting and jumping activities.

Their first article (Stefanyshyn & Nigg, 1997) determined the mechanical energy contribution of the MPJ to the activities of running and sprinting, using the joint energy method. A sagittal plane analysis was performed on five runners and five sprinters who wore their own training shoes. Kinematic (four camera motion analysis system, 200 Hz) and kinetic (Kistler Force Platform, 1000 Hz) data was collected at the 15 m mark, where the athletes were still accelerating. In the running trials, speeds were 4 ± 0.4 ms^{-1} and MPJ peak plantarflexor moments ranged from 40 to 80 $N.m^{-1}$. The MPJ absorbed, on average, 20.9 J ($\pm$ 6.6 J) total energy during stance and produced 0.3 J ($\pm$ 0.2 J) during the take-off phase. In sprinting, mean speed was 7.6 ms^{-1} (range

7.1 - 8.4 ms^{-1}) with stance times between 0.10 s and 0.14 s, and peak MPJ plantarflexor moments were ranged from 70 to 120 N.m^{-1} with quite large intra-individual variation (Figure 2.4). The MPJ absorbed, on average, a total of 47.8 J (± 16.6 J) energy during stance, and produced only a small amount of energy (0 J to 2 J) during the take-off phase, although the mean total energy production throughout the whole stance phase was 6.0 J (± 3.1 J). Overall, the MPJ was responsible for 32% of the energy absorbed in sprinting.

Figure 2.4 Mean (6 trials) MPJ moments and powers for five sprinting subjects (7.6 m·s^{-1}), taken from Stefanyshyn and Nigg (1997).

It appears that the MPJ is a dissipater of large amounts of energy and produces no or little energy during the take-off phase, both in running and sprinting. Stefanyshn and Nigg (1997) stated that the toes remained in a dorsiflexed position during the take-off phase and did not produce any push off. They concluded that athletes absorb energy at the joint as they roll onto the balls of their feet, however, fail to produce any plantarflexion or push off with the toes during the take-off phase. Stefanyshyn and Nigg (1997) also speculated that the inclusion of the toe spring in sprint spikes may force the MPJ to remain in a dorsiflexed position at take-off.

Stefanyshn and Nigg (1998a) determined the mechanical energy contributions of the hip, knee, ankle and MPJ to running long jumps and vertical jumps, movements that are common in athletics activities. Five male basketball players and four male long jumpers performed running jumps with run ups up to 15 m. The movement patterns were similar to those previously found in the literature for standing jumps: a proximal to distal sequence of power production with the ankle absorbing and generating the most energy, and the hip extensors producing power during early stance. Similar to running and sprinting, the MPJ was a large energy dissipater, responsible for 15-16% of the energy absorbed by the lower extremity in the running jumps. In the running long jump the MPJ absorbed on average a total of 43.6 J (± 12.4 J) and produced only 1.8 J (± 1.1 J) during the take-off phase. One limitation to their studies mentioned thus far, was the influence of biarticular muscles was not included, therefore it was unknown how much work performed at the MPJ was as a result of energy transfer from the ankle by muscles such as the flexor digitiorum longus or extensor digitorum longus.

Krell and Stefanyshyn (2006) investigated the extension angle and angular velocity of the MPJ in 76 100 m Olympic athletes. They performed a sagittal plane analysis at the 60 m point, using two video cameras sampling at 120 fields per second, during the 2000 Sydney Olympic Games. They found faster females exhibited higher posterior sole angles at touchdown and smaller posterior sole angles at take-off. Posterior sole angle was defined as the angle between the ground (horizontal) and the inferior surface of the rearfoot in the sagittal plane. The faster males exhibited higher MPJ extension velocities. It was hypothesized that faster sprinters would have lower maximal MPJ extension than slower sprinters; however, they found no evidence in support of this theory. Firstly, maximum MPJ extension alone does not provide sufficient information for prediction of energy absorption. Energy absorption is determined by both joint moment and the movement the joint undergoes. This result indicates that kinetic information is required of the MPJ to predict energy absorption at that joint. Therefore, the lack of kinetic data in this article somewhat limits its applications. Secondly, energy absorption at the MPJ alone does not provide sufficient information for predicting sprint running performance. The MPJ represents only one joint which contributes to the movement of the lower limb during the stance phase of sprint running. The data collected from the 100 m Olympic sprinters was also reported by Stefanyshyn, Krell & Chow (2002) where values of 36.5° and 37.7° peak MPJ extension were reported for males and females respectively. In this publication, there was no relation found between peak MPJ extension and 100 m sprint times.

Stefanyshyn and Nigg (1997, 1998a) and Krell and Stefanyshyn (2006) considered the foot as two segments: phalanges and rearfoot. A pilot study was first performed to help decide how to distribute force plate data to the two segments. For a single subject, pressure data and force data were collected simultaneously, allowing two resultant forces to be obtained. They then compared MPJ moments, calculated using force platform data alone, against using pressure data plus the horizontal forces from force platform. The difference between the two methods was 3% in running and 7% in sprinting Stefanyshyn and Nigg (1997), for long jumping the difference was 5% Stefanyshyn and Nigg (1998a). Hence, for all their studies, a single ground reaction force was used to determine the net joint moments at the MPJ. The MPJ moment was assumed to be negligible until the GRF acted distal to the joint, assuming the inertial effect of the phalanges was negligible. Using the single force vector method chosen by Stefanyshyn and Nigg (1997; 1998a) may lead to underestimation of joint moments at the MPJ. Instead, the use of pressure insoles would allow the division of the vertical ground reaction force at both the phalanges and rearfoot segments, and provide more detailed information specific to each region of contact and possibly more accurate estimates of MPJ kinetics.

In all three articles only sagittal plane information was obtained. There is relatively little information about the movement of the foot in the other planes in sprinting and other highly dynamic activities. In the first and second studies kinematic data was captured at 200 Hz, and in the third 120 Hz was used. Considering the short contact times associated with sprinting, a higher data capture rate (500 Hz or 1000 Hz) would help to improve the accuracy of the data and likely better characterise MPJ motion. Furthermore, another limitation of the Krell and Stefanyshyn (2006) study was that the shoe complex was digitized and used to represent movement of the foot. The authors felt that the movement of the foot and the movement of the foot-shoe complex were tightly coupled, as the athletes were wearing spikes. Although this seems a reasonable assumption, this remains an issue to be confirmed in quantifying foot motion of the shod foot.

In the first two studies, joint markers were placed on the shoe at locations of the heel, MTH5, and the distal end of the toe box. The marker at MTH5 represented the MPJ and divided the foot into two segments: phalanges and rearfoot. For purposes of the investigations the five MPJs were considered as a single joint rotating about a transverse axis. In Stefanyshyn and Nigg (1997, 1998a) the sensitivity of marker placement was calculated. A 1 cm shift of the MTH5 marker resulted in 27% decrease in energy absorbed in running, 23% decrease in sprinting, 12-28% in the running vertical jumps, and 18-27% for the running long jumps. This highlights the need for extremely accurate marker placement which becomes more difficult when placing markers on the surface of the shoe.

2.3.4 Summary

The importance of inclusion of joint moments at the MPJ when calculating joint kinetics at the ankle and knee during sprinting has been recently highlighted by Bezodis *et al.* (In press). The research by Stefanyshn and Nigg (1997, 1998a) and Krell and Stefanyshyn (2006) has also demonstrated the significance of including the MPJ, suggesting that the MP joint is a large energy absorber in dynamic activities and this energy loss appears detrimental. If the energy loss was lessened, or indeed if any energy could be produced during push off, the result may be a positive effect on performance. It may be that, in comparison to the relative stiffness of the foot, athletic shoes are too compliant, to have a marked influence upon the energetics of the MPJ.

2.4 Studies on shoe design and running

Investigations into barefoot running, barefoot training, and barefoot shoes feature regularly within recent biomechanics research as it has experienced a resurge in interest. Future research is needed to establish injury rates among barefoot runners as there remains no epidemiological evidence that barefoot runners have fewer running related injuries (Nigg, 2009).

The biomechanics of sports shoes and the effect of different footwear have been dominant in sports science journals throughout the last three decades and have had a significant and substantial effect on sport shoe development. In 2005, Nigg, Stefanyshyn, Cole and Boyer commented on the past, present and future of footwear research. Their last suggestion for future research to be undertaken between 2005 and 2015 was to identify the reasons for the effect of certain shoe constructions on performance. This section will not provide a full review of past footwear studies but instead focus on the effect of footwear on sports performance. Fundamental questions in research have recurred. Are running shoes with harder or softer soles better? How can energy return be maximised?

2.4.1 Running shoes, impact forces, and injuries

Running is one of the most common and popular forms of exercise around the world. For the average recreational runner, the overall yearly incidence rate for running injuries varies between 37 and 56% (van Mechelen, 1992). As a result, there is a plethora of literature regarding running injuries and footwear. To review all these studies is beyond the scope of this literature review. Several studies have investigated the reduction of impact force in running shoes, in comparison to barefoot running, with increased loading rates and greater impact forces reported when running barefoot. The results of studies concerning the relationship between impact forces and increased injury risk have been equivocal. Some investigators claim that greater GRFs during running may be associated with increased injury risk (Gottschall & Kram, 2005; Hreljac, Marshall & Hume, 2000; Messier *et al.*, 1995; Milner *et al.*, 2006). Others have not found a positive correlation (Bennell *et al.*, 2004; Crossley *et al.*, 1999) with Duffey *et al.* (2000) even reporting decreased injury risk. However, there still exists a common assumption that impact forces, especially in heel-toe running, are the reason for specific injuries and need to be damped. Hence, for example, the concept of cushioning was introduced to reduce impact loading. Nigg, Denoth, Luethi & Stacoff, (1983) did not support this association and the muscle tuning paradigm was developed (Nigg, 2010). Other biomechanical factors that have been associated with the development of running injuries include excessive foot eversion and excessive tibial rotation.

Previous literature has indicated that loading characteristics can be altered by the cushioning properties and the density of the shoe's midsole (Bartlett, 1999). Some studies have reported no differences in impact forces between a hard and soft midsole (Clarke, Frederick & Hamill, 1983; Nigg, Bahlsen, Luethi & Stokes, 1987). Others have found that a softer midsole produced lower ground reaction forces (Devita & Bates, 1988), yet some have reported higher ground reaction forces (Kaelin, Denoth, Stacoff & Stussi, 1985; Snel, Delleman, Heerkens & van Ingen Schenau, 1985). In addition, previous research has shown that running in a shoe with a harder midsole can decrease initial impact forces (De Koning & Nigg, 1994; De Wit, De Clerq & Lenoir, 1995; Kaelin *et al.*, 1985). Previous literature has also focused on various aspects of running shoe design such as examining the differences based on shoe expense (Clinghan *et al.*, 2008), as well as changes in lower extremity mechanics resulting from differences in midsole construction (De Wit *et al.,* 1995). It is clearly evident from the footwear research that individuals respond differently to varying longitudinal bending stiffness. Runners desire different cushioning properties depending on their own preferences and varying running conditions, bodyweights, running styles, and speed (Michel, Kleindenst & Krabbe, 2005). Pressure distribution also varies for different types of footwear (Hennig & Milani, 1995). These findings have implications for matching the footwear characteristics to individual natural foot function (Morio, Lake, Gueguen, Rao & Baly, 2009). Furthermore, Stacoff *et al.* (2001) demonstrated that tibiocalcaneal kinematics of running may be individually unique and that shoe sole modifications may not be able to change them substantially.

2.4.2 Work and Energy

The question of how sports shoes can influence energy strategies and sport performance has been investigated for running shoes. Nigg, Stefansyhyn and Denoth (2000) suggest that an athlete has three major strategies to improve the work-balance during locomotion, these being: a) to store and return energy, b) to minimize the loss of energy, and c) to optimize muscle function.

Currently, there are two main methods to store and return energy within a sports shoe: one is to use the stiffness of the shoe sole (high longitudinal bending stiffness) as a spring, and the second is to deform the sole material by compressing the sole and then return this energy when the material expands (Nigg, 2010). To do this energy return must be substantial enough and be returned at the right time, frequency and location (Nigg & Segesser, 1992). For sprinting, energy return should be delivered during the last 50 ms of contact, thus the loaded natural frequency of the shoe sole for sprinting should be approximately 5 Hz (Nigg, 2010). It is very technically difficult to return energy from a shoe sole at the right time, frequency and location; furthermore, current scientific techniques are not sufficiently developed to make a reasonable estimate of the return of stored energy in the ankle joint / human leg during locomotion (Nigg, 2010).

There are two less-considered energy-based possibilities to improve performance: minimizing energy loss, and optimizing the system. The concept of minimizing energy loss has been demonstrated in the influence of midsole bending stiffness on performance (Stefanyshyn & Nigg, 1997; Stefanyshyn & Nigg, 1998; Stefanyshyn & Fusco, 2004) which is discussed in the next section. The energy absorbed and generated is approximately the same for the hip, knee, and ankle during sprinting, but the metatarsophalangeal joint is a net energy loser as the MPJ flexes during contact. If it did not bend it would not lose energy; therefore increasing the stiffness may reduce the energy lost. This is not a concept that is often used in high performance sport tuning; it may however be an effective possibility to improve performance (Nigg, 2010). Conversely, Oleson, Adler and Goldsmith (2005) stated that typical variations in the bending stiffness of running shoes are unlikely to have a significant effect on performance. They argued that total forefoot stiffness is dominated by that of the human foot, although forefoot stiffness was characterised for running and not sprinting. Shorten (1993) also suggests that differences in the energy dissipated by well-designed shoes are predicted to be small and unlikely to have a direct effect on the energetics of the body as a whole (less than 1%), although again this was based upon the study of running, not sprinting.

2.4.3 Training shoes and barefoot technology

There has been some, albeit limited, research into training shoes worn by runners (mainly distance runners). Logan *et al.* (2010) reported ground reaction differences between running shoes, racing flats, and distance spikes in long distance runners. Although there were few statistically significant differences due to high variability in the kinetic measures (males and females also responded differently), both the racing flats and spikes increased the loading rates, peak vertical forces, and peak braking forces when compared to running shoes. Wiegerninck *et al.* (2009) also demonstrated a significant difference between training shoes and racing flats in terms of peak pressure, maximum force (greater in the racing flats) and contact area (smaller in the racing flats).

'Barefoot shoes' have been recently developed by several manufacturers, based on different conceptual ideas (Nigg, 2009). The Adidas 'Feet You Wear' shoes mimic the shape of human foot to reduce foot eversion / pronation by reducing the levers at the heel and have been shown to reduce the frequency of injuries amongst basketball players (Meeuwisse, Selmer & Hagel, 2003). The Nike Free shoes were designed to mimic the kinematics of barefoot running, increasing foot muscle activity, resulting in a 20% increase in toe flexor strength and a 7% decrease in the range of motion at the MPJ during running (Potthast *et al.*, 2005). A 29% reduction in lower limb injuries was also reported ten months after a five month test period of wearing the shoes (Bruggemann, Goldmann & Potthast, 2008), although the mechanism for this reduction is not known. The MBT (Masai Barefoot Technology) concept shoes have a rounded bottom profile, mimicking the feeling of barefoot movement on soft ground. The unstable sole was designed as a training device for the feet and leg muscles. Romkes, Rudmann & Brunner (2006) reported that the muscle activity of the gastrocnemius and tibialis muscles increased, with co-contraction of these muscles providing the additional stability required during walking gait with the MBT shoes. Nigg, Emery & Hiemstra (2006) also reported an increase in balance time when using the MBT shoes over a period of three months and a 25% decrease in subjective knee joint pain for arthritic subjects. It appears that 'barefoot shoes' may provide some benefit to the athlete. Barefoot training, or related strategies that strengthen the small muscles crossing the ankle joint, would be beneficial to athletes; however, at present, there is no evidence of the potential benefit to elite performers or indeed sprinters.

2.4.4 Sprint shoe design

There has been little published in the literature about the design of sprint spikes. Currently sprint shoe outsoles are injection moulded and researchers who have adjusted the stiffness of sprint spikes have done so with carbon insoles. However Toon, Hopkinson & Caine (2007) customised the mechanical properties of sprint spikes using selective laser sintering nylon-12 sole units.

Mechanical tests and subject tests have been used to evaluate the cushioning properties of sports footwear (Lake, 2000). Mechanical tests have been used to measure the bending stiffness of running shoes, whereby, three-point bending tests have been used in accordance with the ASTM standard for flexibility tests of running shoes (ASTM, 1994) (Kleindenst *et al.,* 2005; Oleson *et al.,* 2005; Roy & Stefanyshyn, 2006). For sprint spikes, Stefanyshyn and Fusco (2004) quantified the bending stiffness of the shoe insoles with a three-point bending test, but did not measure the shoe stiffness of the athletes' standard sprint spike shoes, thus the mechanical characteristics of the shoe complex were unknown. Toon *et al.* (2011) reported the primary mechanical performance measurements of eleven pairs of commercially available sprint spikes. Bending stiffness was quantified using a three point bending test, following procedures of Oleson *et al.* (2005), whereby the shoe is held in place by a rigid rearfoot last and vertical forces are applied to the forefoot. Bending occurred about the region associated with the MPJ, at a distance of approximately 70% of the total shoe length from the heel counter.

The current knowledge base for comfort of running shoes is small and for sprint spikes is non-existent. None of the studies involving sprint spikes reported any subjective measures of comfort, stiffness, or performance. It has been suggested that comfort should be at the centre of sport shoe development (Nigg, 2011). Comfort is very much a subject-specific and ever-changing perception that is influenced by mechanical, neurological and psychological factors. It is currently known that direct assessment of comfort of sports shoes is highly unreliable (Mundermann, Stefanyshyn & Nigg, 2001), although assessment through comparative measures such as Visual Analogue Scales and ranking scales have shown higher reliability (Mills, Blanch & Vicenzino, 2010; Mundermann, Nigg, Stefanyshyn & Humble, 2002).

2.4.5 Shoe bending stiffness, running and jumping performance

Stefanyshyn and Nigg (2000) investigated the effect of changing midsole hardness on running (heel-toe running) and vertical jump performance. Three different running shoe conditions were used to test their hypothesis: a control shoe (stiffness of 0.04 $N.m.deg^{-1}$), the same shoe with a stiff midsole (0.25 $N.m.deg^{-1}$), and with a very stiff midsole (0.38 $N.m.deg^{-1}$). Commercially available running shoes (Adidas Tech Road) were used and carbon fibre plates were inserted into a pocket of the shoe midsole, formed by removing 5 mm thick ethyl acetate material from the entire length of the midsole. Three 1 mm carbon plates were inserted for the stiff shoe and five 1mm carbon plates for the very stiff shoe. The carbon fibres were aligned with the anterior-posterior axis and the carbon plates were secured together with athletics tape.

Kinetic data (1000 Hz) and kinematic data (200 Hz, 4 cameras) were obtained for a total of 18 running (4.0 ± 0.4 ms^{-1}) and 18 jumping trials per subject. Each subject performed six trials per shoe condition over a 3 week period. The foot was modelled as in Stefanyshyn and Nigg (1997): the MPJ axis of rotation was located at the 5th metatarsal head and the data was analysed in 2D (sagittal plane). Energy absorption and generation at the hip, knee, and ankle did not differ between the three shoe conditions for both running and vertical jumping.

During vertical jumping, the stiffer shoes also decreased the amount of energy absorbed at the MP joint compared with normal shoes but did not increase the energy created at take-off. The stiffer shoes resulted in a significantly increased vertical jump height (by 1.7 cm) compared to the normal shoes.

In running, the energy absorbed at the MPJ while wearing either the stiff shoe

(19.6 J) or the very stiff shoe (17.7 J) was significantly less than while wearing the control shoe (27.6 J). There were no significant differences in the amount of energy generated at the MPJ between the different shoe conditions: energy generated was 0.4 J for the control shoe, 0.7 J for the stiff shoe and 0.7 J for the very stiff shoe. Even with the stiffer shoes, Stefanyshyn and Nigg (2000) reported that the MPJ remained dorsiflexed at take-off, suggesting the concept of energy return by shoes is not supported. However, increasing stiffness did result in a reduction in the loss of energy, suggesting this concept is more applicable. Hence, subsequent research has focused on how to reduce this loss of energy which could lead to improved performance.

Over the past 20 years, long jump and high jump spikes have progressed to have relatively stiff midsoles (Stefanyshyn & Nigg, 2000). The data from the investigation by Stefanyshyn and Nigg (2000) may help explain this natural progression to stiff shoes for performance jumping applications. In contrast, shoe manufacturers seem to be moving toward running shoes that are more flexible at the MPJ by either increasing the flexibility of the materials or modifying the structure of the midsole (e.g. incorporating flexion grooves). The authors speculated that this may not be beneficial with respect to performance and, rather, may be driven by comfort aspects of the recreational runner. However, running performance would need to be directly quantified as a function of shoe stiffness to either support or refute this speculation.

Roy and Stefanyshyn (2006) focussed on the effect of shoe midsole bending on running economy in heel-toe running. Nine heel-toe distance runners were tested for three running shoe conditions: an unmodified control shoe (stiffness 18 N·mm), and two test shoes with carbon plate insoles (bending stiffness 38 N·mm and 45 N·mm). The subjects underwent treadmill running economy tests lasting six minutes (two tests in each shoe, shoe conditions were randomised and trials were blind) and oxygen consumption was measured. They also performed 20 running trials in each shoe in the biomechanical tests where EMG (sampling at 2400 Hz) of five lower extremity muscles were collected along with kinematic (240 Hz) and kinetic (2400 Hz) data. The foot was modelled as the forefoot and rearfoot, as Stefanyshyn and Nigg (1997). However, Roy and Stefansyhyn (2006) used the midpoint of the 1st and 5th metatarsals as the joint axis of rotation, unlike Stefanyshyn and Nigg (1997, 2000) who used solely the 5th metatarsal.

The stiffer shoes resulted in decreased oxygen consumption (1% metabolic saving). There were also significant differences in ankle joint moments, supporting an influence on the length and velocity of stretch of the ankle plantarflexors. There were no differences in MP joint moments, possibly due to the differences in the way this joint was analysed, but possibly also suggesting that the principle of minimising energy loss at this joint may not apply for endurance activities, as there was no difference in EMG activity between different shoes. Therefore, the mechanisms that can be attributed to the improvement in running economy are not fully understood.

Kleindienst, Michel and Krabbe (2005) investigated the effect of varying midsole hardness on the MPJ, based on kinematic and kinetic during heel-toe running. The possible dependency of gender, bodyweight and running velocity was also investigated in order to provide guidelines for sport shoe construction with reference to injury prevention. They used three shoe types: Adidas Manhattan running shoes, varying in midsole hardness (40 Shore C, 55 Shore C and 70 Shore C).

A total of 28 subjects performed running trials at both 3.0 (± 0.2) ms^{-1} and 4.5 (± 0.2) ms^{-1}. Kinematic (200 Hz) and kinetic data (1000 Hz) were collected for five trials per condition. The MPJ centre of rotation was represented by midway between the 1st and 5th metatarsal. They presented only the MPJ angle and moment, for males and females separately.

They found that, for females, the MPJ was significantly more dorsiflexed in the soft shoe than the stiffest shoe; however, only when running at 4.5 (± 0.2) ms^{-1}. For males, the MPJ was significantly more dorsiflexed in the soft shoe compared to the hard, as well as to the moderate shoe, at both running speeds. At the running speed of 4.5 $m \cdot s^{-1}$ the hard shoe modification revealed significant higher MPJ moments than the moderate and the soft shoe for both genders. They stated their findings could be indications for a gender specific reaction pattern caused by varying bending stiffness of the midsole based on kinematics.

2.4.6 Shoe bending stiffness and sprinting performance

Stefanyshyn and Fusco (2004) investigated the effect of increased bending stiffness on sprint performance and whether simple anthropometric factors could be used to predict shoe bending stiffness for optimal performance. A total of 34 athletes, comprising university athletes and Cuban national athletes, all specialist sprinters / jumpers / hurdlers or combined eventers took part. The athletes completed eight sprint trials (two per condition) with four different stiffness conditions: their own sprinting spikes, their own shoes plus 42 $N.mm^{-1}$, 90 $N.mm^{-1}$, and 120 $N.mm^{-1}$ carbon plates. The carbon plates were 1 mm thick and inserted under the sock liners. However, the stiffness of the subjects' own shoes was not measured. Their sprinting performance over 20 m (20-40 m from the start) was recorded using single beam timing lights placed at chest height. No kinematic / kinetic data were recorded.

The results demonstrated that increasing the midsole bending stiffness significantly improved performance. Performance improved by 0.69% whilst wearing the 42 $N.mm^{-1}$ plates ($p = 0.07$), but did not further improve with increasing stiffness. This was deemed a significant effect considering the small differences between elite performers (Hopkins, Hawley, & Burke, 1999). There were large individual differences between athletes, with some sprinters performing better with flexible plates and others performing better with stiff plates, highlighting that there is large inter-individual variation in how athletes respond to manipulations in footwear stiffness. One limitation of the study was that athletes wore their own sprint spikes and the overall stiffness of their footwear plus insoles was, therefore, unknown. These differences in each athlete's own footwear may have influenced the results; although Stefanyshyn and Fusco (2004) speculated that the magnitude of these errors would not have had a large effect on the results. Furthermore, the sprint tests were performed on an outdoor running track where the wind speed differed between trials. In order to limit the influence of wind, only trials that were collected at wind speeds between +1.0 and -1.0 $m.s^{-1}$ were used in the analysis. Even within these tight constraints, differences in wind speed could still influence the results (Linthorne,1994).

Unlike Kleindienst *et al.* (2005), Stefanyshyn and Fusco (2004) found no anthropometrical effects; optimal stiffness for performance was not affected by height, weight or gender. Stefanyshyn and Fusco (2004) speculated that increasing the stiffness would move the point of application of the GRF anteriorly. This would increase the moment arm and resultant moments seen at the ankle, so the ankle plantarflexors would have to produce additional force to counteract the increased moment. They, therefore, suggest that optimal stiffness may depend on individual force-length and force-velocity characteristics. This needs studying experimentally, as Stefanyshyn and Nigg (2000) found increasing bending stiffness to increase only MP joint moments and not ankle joint moments for running. This possible relationship needs to be established in more detail, examining whether force produced by the ankle plantarflexors changes with varying midsole hardness.

No researchers have directly studied the effect of sprint spike stiffness on foot function, nor provided any evidence for the potential biomechanical mechanisms that may contribute to a possible improved sprinting performance. Ding, Sterzing, Liu and Cheung (2011) examined the interaction of individual athlete's MPJ stiffness with different sprint spike stiffness but found no influence of either for sprint acceleration performance or for biomechanical variables. However, biomechanical data was only collected for the first two foot strikes out of blocks, only two sprint spike stiffness conditions were used, and young athletes (aged 14.8 ± 1.7 years) with relatively low sprinting experience (3.5 ± 1.8 years) were used, therefore little useful information was gained. Toon, Williams, Hopkinson & Caine (2009) demonstrated some kinematic differences between barefoot sprinting and sprinting in sprint spikes at 10 m and 50 m, with the sprint spikes reducing MPJ range of motion and angular velocity significantly. However, the relationship between these kinematic factors and sprinting performance is unknown; therefore, the implications of these findings are not understood and kinetic measures are needed to provide further insight.

Toon, Hopkinson and Caine (2008) and Toon, Vinet, Pain and Caine (2011) have also presented conflicting data on the effect of sprint spike bending stiffness on jumping performance. Toon *et al.* (2008) identified no effect of stiffness on concentric jump performance, and performance in the bounce drop jump was reduced by stiff sprint spikes, in comparison to the control shoe (Vibram, Fivefingers). However, in 2011, the results for a single sprinter demonstrated that a medium stiffness shoe was best for squat jumps, and the maximum stiffness shoe was best for bounce drop jumps. There were no differences in MPJ moments, powers or energies, although the stiffer shoes actually reduced the joint moments (but not significantly). In both studies, bounce drop jump and concentric jump performance were assumed to provide an accurate representation of maximal speed sprinting and starting performance, respectively.

2.4.7 Summary

There are some fundamental limitations and gaps in the research in this area to date. Different authors use different values of stiffness, making it harder to compare the studies. The studies have looked at MPJ and ankle joint motion in the sagittal plane only, and few studies have incorporated kinetic measures. Although more research has been done on heel-toe running, the effect of shoe midsole bending stiffness on sprinting has only been determined with timing gates to quantify sprinting performance. Further investigation, reporting 3D kinematics, kinetics, and pressure data, is needed to help explain why sprinting and jumping performance improve with increased bending stiffness; and to determine if the loss of energy at the MPJ is reduced, and, if so, the magnitude of this reduction. This research could be applied to other dynamic activities, such as hurdling, running jumps, and jump landings, to see whether increased shoe bending stiffness improves performance. Further data on the energetics of sprinting and jumping with a stiffened MPJ are required to fully understand how to tune shoe stiffness appropriately to individuals to improve performance. Finally, the speculation that specific characteristics, such as that the force length and force velocity properties of calf muscles of individual athletes may be related to the shoe stiffness required for maximal performance, still remains unanswered. Overall, researching the concept of reducing energy loss at the MPJ by manipulating shoe stiffness in order to improve performance appears a very promising area for the future of footwear design.

2.5 Methodological considerations – foot models

2.5.1 Multi-segment foot models in clinical research

Traditional running gait analysis considers the foot as a single rigid segment, but this approach fails to provide meaningful information of the kinematic behaviour of the foot (Okita, Meyers, Challis & Sharkey, 2009). In clinical gait analysis of walking, various foot models have been proposed with varying number of segments and degrees of freedom. Over the last decade, these models have been validated, accepted and widely adopted in clinical practice. Most investigators appear to use three segments, one of which is the hallux (Baker & Robb, 2006). However, the area is still relatively new and developing; normative data is appearing in the literature and researchers are far from deciding on a standard model for clinical purposes. There seems to be common consensus that the foot can be modelled as a small number of segments with Euler angles used to express the angular relationships between the segments. The following section summarises some of the recent clinical research into multi-segment foot modelling.

Morlock and Nigg (1991) demonstrated that the internal forces calculated within the foot depend entirely on the formulation of the model used. They suggested using a foot model that resembles the functional anatomy of the foot more closely, such as a 6 degree of freedom model. This enables the quantification of the transmission of GRFs through the foot, an important factor in understanding the functional behaviour of the foot. The research of Scott and Winter (1993) was one of the first to devise a multi-segment gait foot model. This comprised eight segments and eight monocentric, single degree of freedom hinge joints, but only considered the hallux as they suggested that the other four toes provide minimal load support during walking stance.

Leardini, Benedetti, Catani, Simoncini and Giannini (1999) presented a five segment, six degrees of freedom model which included a shank, calcaneus bone, midfoot segment, 1st MT segment and, finally, proximal phalanx of the hallux. Using multiple marker plates (stating that the rigid clusters of markers can embrace the underlying bones better than individual skin-mounted markers) they collected normative kinematic data for adults. They identified joint rotations for the tibia-fibula to calcaneus, calcaneus to midfoot, 1st metatarsal to midfoot, and hallux to 1st metatarsal.

Carson, Harrington, Thompson, O'Connor and Theologis (2001) presented a three segment foot model for the evaluation of barefoot walking kinematics with a hindfoot, forefoot and hallux segments, plus a tibial segment. Stick markers were used for the hallux segment (Figure 2.5) and four inter-segment pairs were examined: tibia with respect to floor; hindfoot with respect to tibia, forefoot with respect to hindfoot; and hallux with respect to forefoot. Also named the Oxford foot model, this is perhaps the most widely used and referenced model in clinical biomechanics. This was later adapted and validated for children by Stebbins, Harrington, Thompson, Zavatsky and Theologis (2006).

Figure 2.5 Markers used during reference position for Oxford foot model and hallux (HX) segment description (Carson *et al.,* 2001) HX: Sagittal plane perpendicular to the floor (from static calibration) and through the stick markers (HLXP, HLXD) aligned with its long axis; transverse plane parallel to the floor (from static calibration).

MacWilliams, Cowley and Nicholson (2003) used a detailed nine segment model to produce normative kinematic and, encouragingly, kinetic foot data for adolescents. They combined pressure and GRF data and provided valuable baseline data. The foot was also divided medial-laterally with medial and lateral forefoot and toes segments as well as a hallux (hallux marker triad used), and joint moments demonstrated large medio-lateral differences. Each of the eight articulations demonstrated flexion / extension, actions which are masked when using a single segment model. Furthermore, they indicated that single segment models overestimate ankle joint powers in gait. The results demonstrated that as the tarsalmetatarsal joints generated power, the metatarsophalangeal joints absorbed power simultaneously; also that the medial rays of the foot carry greater load, flex more and generate more power than the lateral rays. However, Buczek, Walker Rainbow, Coney and Sanders (2005) are critical of the model of MacWilliams *et al.* (2003), stating that the moments and powers need to be viewed with caution, as they ignore force between mediolaterally adjacent segments. Therefore the assumptions of the model require more critical evaluation before routine clinical implementation can be considered.

Simon *at al.* (2006) presented a different kinematic measurement method; the Heidelberg foot measurement method (HFMM) which does not define different segments, but a new way of describing joint angles, defining projection angles between anatomical landmarks or reference points (similar to those used in radiology) and by assuming hinge joints with one degree of freedom. Although this method may be an effective means of describing foot deformities, Baker and Robb (2006) suggested segment based models are more suited to explain biomechanical mechanisms.

There have since been additional multi-segment kinematic models presented in the literature, with and without toe / hallux segments (e.g. Jenkyn & Carol, 2007, Okita *et al.,* 2009) but, overall, there remains no consensus on segment or joint definition. Okita *et al.* (2009) added further evidence that the medial and lateral components of the forefoot behave differently with observed variation in the profiles of the 1st and 5th MT heads, indicating relative motion between these and a departure from the rigid body assumption. However, they had no toes / hallux segment; therefore, the effect on any motion at the MPJ is unknown.

It is clear that modern clinical gait analysis is now capable of capturing the motion of small multiple markers on the foot as well as those on the rest of the body during locomotion. In clinical biomechanics, many models have been suggested for obtaining baseline data of normal foot function. They have shed some interesting light onto modelling the foot and have highlighted problems with multi-segment models.

2.5.2 Foot models in sports research and modelling of the MPJ

In sprinting studies, traditionally the foot is modelled as a single rigid segment. For 2D sagittal plane analysis, typically points on the ankle and distal end of the foot are manually digitised to define the foot (e.g. Johnson & Buckley, 2001), for 3D analysis the foot segment is defined from markers on the head of the second metatarsal to the ankle joint centre, (e.g. Hunter *et al.,* 2004c) although additional markers are placed on the heel and medial side of the big toe. This is clearly oversimplifying the foot segment. With just one centre of rotation for dorsi / plantarflexion and internal / external rotation, the information obtained about foot motion is inadequate. Multi-segment foot models that include the forefoot / toes have been developed, during approximately the last ten years, for studying athletic tasks, in particular running, sprinting and jumping. However, these are relatively sparse and greatly vary in complexity and accuracy for ascertaining the role of the forefoot / toes for propulsion.

The simplest foot models presented in sports biomechanics research have been 2D analysis, based on the digitising of a single marker to represent the division between the forefoot and toes. Krell and Stefanyshyn used the 1st metatarsal head to model the MPJ as a single hinge joint. Toon *et al.* (2009) digitised both the medial first MPJ and lateral fifth MPJ separately and then aggregated (averaged) the medial and lateral aspects.

Stefanyshyn and Nigg (1997, 1998b, 2000) used a two segment foot (phalanges and rearfoot), with markers on the toe and 5th metatarsal head to represent a toe segment and a marker on the heel for the rearfoot segment. The MPJ centre was considered to be at the location of the 5th metatarsal head and the five MP joints were considered as a single joint rotating about a transverse axis. This assumes that the MPJ can be accurately represented by the motion of a single marker on the lateral portion of the metatarsal heads. Motion of the foot segments was measured by tracking markers placed on the external surface of the shoe upper. Therefore any relative internal movement between the foot and shoe was not accounted for.

Oleson *et al.* (2005) used a two segment foot to ascertain the stiffness of the forefoot in running barefoot, with a forefoot segment defined by MTH1 and MTH5 and 2nd toe markers and the rearfoot defined by proximal, distal, and posterior calcaneus markers (Figure 2.6). The MPJ was defined as the line segment joining MTH1 and MTH5. Whilst this model offers a more anatomically correct oblique representation of the MPJ axis, the motion of the forefoot is quantified relative to the rearfoot and not a midfoot segment. Therefore any motion within the rearfoot segment may adversely influence the resultant MPJ kinematics and kinetics.

Figure 2.6 Markers locations taken from Oleson *et al.* (2005), p. 1887.

A similar approach was taken by Kleindienst *et al.* (2005) and Rolian *et al.* (2009). Although Kleindienst *et al.* (2005) also used three markers per rearfoot and forefoot segment, they chose the midpoint of the MTH1 and MTH5 to represent the MPJ centre of rotation. Rolian *et al.* (2009) defined the phalanges using markers on MTH1, MTH5 (transverse MPJ axis, single hinge like axis), and the distal phalanges of the first and third toes. A perpendicular line from the transverse MPJ axis to the COP was used to estimate the moment arms and moment at the MPJ during walking and running.

In sports performance biomechanics research, modelling of the MPJ has not advanced further from the oblique axis representation described above, used for barefoot running / studies on running. Only single foot segment or 2D single marker approaches for the MPJ have been implemented for sprinting studies. Bosjen-Moller (1978) proposed that two axes for the MPJ are utilised: a transverse axis through MTH1-2 and an oblique axis through MTH2 to MTH5. However, the oblique definition of the MPJ axis using MTH1 and MTH5 using a three- dimensional marker capture system appears to be a beneficial step forward from 2D approaches based on single medial or lateral markers placed on top of footwear.

2.5.3 Skin marker models

One large source of error with modelling the foot as a multi-segmented structure is motion artefact due to skin and soft tissue (or footwear) motion in relation to the underlying bones (Lundberg, 1996; Reinschmidt *et al.*, 1997). Clinical foot models have varying degrees of evidence supporting their reliability but few present evidence of how accurately the segments reflect the kinematics of the underlying bones.

Two research groups have compared foot kinematics using bone and surface mounted markers. Nester *et al.* (2007) compared kinematics derived from a four segment model - heel, navicular/ cuboid, medial forefoot and lateral forefoot- over three separate data collection sessions, with potentially different standing positions. Overall, the match between the kinematic data between skin, plate and bone protocols was reasonable or good, and the authors stated that is unlikely that no one particular rigid body or marker attachment method is always preferable over another. They implied that, although rigid segment foot models will not be able to accurately capture the precise kinematics of individual foot structure, they will continue to be of value in indicating general trends in the effects of interventions, such as footwear.

Okita *et al.* (2009) also compared a virtual clinical marker set derived from bone-mounted markers versus skin markers for a three segment foot model (shank, hindfoot, and forefoot). The comparison of the segment and joint angles revealed only small differences as a result of soft tissue artefact, although the forefoot segment violated the rigid body assumption with relative motion between the first and fifth metatarsals. Similarly to Nester *et al.* (2007) the authors were more concerned with rotations of the rearfoot than motion of the forefoot and also did not look at MPJ motion.

Stefanyshyn and Nigg (1997) demonstrated the sensitivity of MPJ kinetics to marker location, with a 1 cm shift in the MP marker in the anterior-posterior direction resulting in an average decrease of 27% and 23% energy absorbed at the MPJ for runners and sprinters respectively, although their MPJ definition was based solely on a single lateral marker on MTH5. All of the researchers that have investigated MPJ motion during shod athletic tasks have placed markers directly onto the shoes' uppers as close to the underlying foot anatomy as possible. Bishop, Paul, Uden and Tewis (2011) reported the reliability of landmark palpation through the shoe uppers versus barefoot, using X-ray, and reported absolute errors of 0 - 3.9 mm on the hindfoot and forefoot segments and up to 10.1 mm on the hallux, although the effect on the resulting kinematics is unknown. Other researchers have suggested cutting windows in shoes or using custom-made shoes with holes to give a better indication of the actual position of the foot bones (Eslami, Begon, Farahpour & Allard, 2007; Reinschmidt, Stacoff & Stussi, 1992).

2.5.4 Summary

There remains a large gap between the models used in sports performance research and the new advanced foot models for clinical gait analysis. This gap clearly needs bridging to gain further insight into foot function in the more dynamic activities related to sports performance. In order to study dynamic forefoot function during sprinting, a multi-segment foot model is needed with an anatomically appropriate definition of the MPJ axis and relative motion quantified between appropriate forefoot and midfoot segments. When using markers placed upon the skin or on top of footwear, an appropriate marker attachment system is needed to ensure repeatability of marker placement and to minimise any unnecessary relative movement between the marker and underlying bone.

2.6 Data collection and processing in sprint biomechanics investigations

For biomechanists to be able to answer research questions with confidence, to make comparisons between different conditions, and to make conclusions based on the implications of a research study, that study must be well designed and controlled. Appropriate study design, experimental set up, data collection and processing procedures are vital to collect high quality data and to ensure threats to internal and external validity are minimised. In order to provide meaningful data, raw data must be processed. For high speed dynamic activities, such as sprinting, the correct smoothing of movement transients is of particular importance. In order to adequately capture marker based motion, the motion not only needs to be tracked accurately but that motion needs to be adequately sampled and processed.

2.6.1 Equipment used to measure sprinting biomechanics and foot function

In order to gain a detailed insight into the biomechanics of sprinting, the function of the foot and the role of footwear, then kinematic, kinetic and plantar pressure data need to be combined. This will enable detailed quantification of the motion of the MPJ, the loads experienced by the joint and the role of the joint, along with descriptions of how loads are distributed underneath the foot during sprinting. The choice of the appropriate equipment and data collection procedures is an important issue.

Motion analysis

Two-dimensional video analysis has been used in the literature to analyse the motion of the foot during sprinting (Krell & Stefanyshyn, 2006; Stefanyshyn & Fusco, 2004). The main advantage to manual video analysis is that it offers the opportunity to collect data in an externally valid situation, or during competitive situations, without the use of markers. However, the digitising process is time consuming and will inevitably introduce some systematic and random errors to the co-ordinate data (Payton, 2008). Bezodis *et al.* (2008) reported joint power root mean square error values, ranging from 2.9% (knee) to 8.4% (hip), attributable to the digitising process.

Automatic video analysis systems are now the norm in sports biomechanics research. Although, in most situations, data collection is confined to a laboratory setting, the advantages in terms of image quality, marker tracking, and high speed data collection usually outweigh the reduction in external validity.

High video frame rates are necessary to capture high speed activities and impacts. Biomechanical data related to impacts involving large accelerations can be prone to error due to inadequate sampling rates (Knudson & Bahamonde, 2001). Therefore, the collection of kinematic data at high frame rates should minimize the error associated with the derivatives of displacement data, although this is mostly dependent on the signal-noise ratio in the raw displacement data. Studies that have documented sprinting kinematics have typically captured at rates between 100 and 500 Hz, with most researchers investigating lower limb / foot function during sprinting choosing to capture at around 120 – 200 Hz (Bezodis *et al.,* 2008; Gittoes & Wilson, 2010, Krell & Stefanyshyn, 2006; Stefanyshn & Nigg, 1997; 1998; 2000; Toon *et al.*, 2009). It is unclear whether these sample rates are adequate for capturing and describing movement characteristics of foot contact during sprinting which typically lasts 80-120 ms (Bushnell & Hunter, 2008; Mero, Komi & Gregor, 1993). It is logical to assume that displacement data obtained in these studies are unable to provide enough data points to adequately describe the curves for these short duration, high frequency movement characteristics. For example, if the sample rate is 100 Hz, for a 100 ms stance phase, only 10 data points would be obtained, clearly insufficient for high frequency movements. This is similar to Numone, Lake, Georgakis, & Stergioulas (2006) who reported that typical sample rates and filtering procedures were inadequate to capture the lower limb impact phase kinematics of the soccer kick.

Force transducers

Force platforms are commonly used in biomechanics studies of running and sprinting due to their relative ease of implementation and high accuracy and resolution. Several researchers have used force platforms to determine the ground reaction forces, impulses, powers, as well as joint kinetics, during the different phases of sprinting.

For accurate kinetic data to be collected, contact with the force plate must occur within the boundaries of the plate, so that the measured force is not affected by force being applied to the surrounding surface. The need for contact within the plate boundaries can lead to rejected trials if foot contacts overlap the boundaries of the plate (Johnson & Buckley, 2001), as this increases the number of trials required to allow collection of sufficient data for analysis. One problem associated with the use of force plates to collect sprint data is the size of the force plate surface relative to athletes' step length. As a result, multiple force plates have been used (Exell, Kerwin, Irwin & Gittoes, 2011). Abendroth-Smith (1996) noted the detrimental effects on the kinetic data when athletes target the force plates to increase frequency of acceptable contacts.

The centre of pressure is an important component required for the calculation of moment arms and resultant joint moments using inverse dynamics and has also been used as a measure of balance and foot function. It is well established that the accuracy of the centre of pressure is compromised towards the edges of the plate (Bobbert & Schemardt,1990) and for small forces at the beginning and end of stance (Nigg & Herzog, 2007), and that a small change in the point of application can lead to substantial errors in resultant joint moments. Consequently correction algorithms have been proposed (Bobbert & Schemardt, 1990; Schmiedmayer & Kastner, 2000).

Plantar Pressure measurement

Pressure mats and insoles can also be used to calculate the centroid of pressure at the foot/ground and foot/shoe interface, respectively. Measurement of the plantar pressure, that is, the distribution of force over the sole of the foot, provides detailed information specific to each region of contact and provides a good indication of how the load is transferred during ground contact. Plantar pressure measurement systems are now able to record plantar loading transitions at high data acquisition rates and, therefore, have the potential to predict rapid movement characteristics of the foot and lower leg (Robinson & Lake, 2005).

Platform systems are restricted to use in a laboratory setting and have been used extensively in the literature for barefoot measurements. Insole systems are better suited for evaluating footwear and footwear modifications. Several researchers have published pressure data for running and athletic tasks, although, to date, no comprehensive research study has been published on foot pressure patterns in sprinting. This may be due to the tendency of pressure insoles to deform when rapidly bent. The validity of these devices can be affected by several factors including sensor accuracy, repeatability, size, number, arrangement, and sampling rate (Chesnin, Selby-Silverstein & Besser, 2000); however, these only measure normal, not shear forces. Typically, the plantar pressure recorded is divided into regions based on key landmarks of the foot, using visual inspection, although the accuracy of this masking process has been questioned in the literature, especially in the presence of abnormal foot contact (Miller, 2010), and has led to alternative mapping techniques (Pataky *et al.*, 2008).

2.6.2 Data smoothing

The importance of using an appropriately high sampling rate when sampling time series data has previously been discussed. Whilst sampling theorem states that a signal should be sampled at a rate that is greater than twice the highest frequency component in the signal, a more general guideline is to sample at a rate ten times greater than the anticipated highest frequency in the signal (Challis, 2008).

Undesired random errors (noise) contaminate all converted coordinate data and must be smoothed. Minimal noise within raw displacement data must be removed as noise is amplified when derivatives are calculated (Winter, 1990) and small errors in displacement data will have a dramatic effect upon the velocity and acceleration. As these are key inputs for IDA, the resultant joint moments and powers will consequently be exceptionally contaminated with noise. It is essential that noise content of the signal is minimized prior to data differentiation, as differentiation amplifies the signal and so can lead to considerable inaccuracies in the derivatives, and for inverse dynamics the second derivatives are required. With image based motion analysis it is assumed that the movement signal occupies the low frequencies, with noise more apparent at higher frequencies. Therefore, to reduce the influence of noise, the data is low-pass filtered. There are various methods for low-pass filtering data, which smooth the data and reduce the noise, whilst leaving the true signal relatively unaffected. These include: polynomial smoothing (e.g. Pezzack, Norman & Winter, 1977), spline functions which do not require equally spaced data (e.g. the Quintic spline used by Vaughan, 1982) and digital filters. Fourier analysis (e.g. Hatze, 1981) also detects and removes noise content of the signal by truncation.

Digital Filtering

Digital filtering is aimed at the selective rejection, or attenuation, of certain frequencies. Low-pass filters can be used to remove high-frequency noise whilst passing the lower signals below a specified cut-off frequency (Winter, 1990). Low-pass Butterworth filters are used commonly in biomechanics research owing to their simplicity and acceptable performance (Erer, 2007, Pezzack, Norman & Winter, 1977); more specifically the fourth-order, zero-lag, low-pass digital filter. Critically damped digital filters have been suggested as an alternative to the Butterworth filter, as they remove the undershooting and overshooting (Robertson & Dowling, 2003); however, for data that need to be double-differentiated the Butterworth remains the better choice.

The selection of cut-off frequency (FC) is very important when filtering the data. Some authors have selected the degree of filtering by visual inspection of the curves, although the repeatability and objectivity of this approach has been questioned (Challis, 1999; Derrick, 2004). Others have used previously published methods for selecting cut-off frequencies, ignoring the variations in the quality of the data. Winter (1990) suggested the cut-off frequency could be determined by conducting either a harmonic or residual analysis. The power of each harmonic component of the data is examined in harmonic analysis and a decision made as to how much power of the raw data should be accepted as the signal of interest. In residual analysis, the raw data is filtered at different cut-off frequencies, and the residuals between the filtered and raw data are determined. An intercept is made which can be used as a guide to choose the appropriate cut-off frequency, with the compromise that the amount of signal distortion and amount of noise passed through the filter are equal. This procedure has been used commonly in biomechanics research.

Yu, Gabriel, Noble and An (1999) developed regression equations to estimate the mean optimal cut-off frequency for a given sampling frequency. Sampling frequency accounted for 91% of the total variance of the optimum cut-off frequency. The suggested that cut-off frequencies derived from residual analysis alone were too low, especially when the sampling frequency is high and could, therefore, result in over-smoothing and should be evaluated cautiously. Challis (1999) also presented a procedure for the automatic determination of cut-off frequency by representing the differences between filtered and unfiltered data to approximate white noise, then finding the cut-off frequency which gave the smallest difference between estimated and true signal values. This procedure performed similarly to the generalized cross-validated Quintic spline (Woltring, 1986). Therefore, there are many objective and repeatable methods for determining the optimal cut-off frequency and it is the responsibility of the researcher to choose the most appropriate for their data. Separate cut-off frequencies may be required for each individual landmark in each direction. van den Bogert and de Koning (1996) also highlight the need for different cut-off frequencies for inverse dynamics of different lower extremity joints.

Butterworth filters operate on the assumption that the signals to be processed are stationary (Woltring, 1995) and cannot accommodate changes in signal power over time – for example, during a gait cycle. Peak accelerations caused by impacts have been demonstrated to be noticeably underestimated when the displacement data are low-pass filtered using a low cut-off frequency. Therefore, to get a better estimation, the cut-off frequency needs to be increased which may result in oscillations in the rest of the acceleration outputs (Giakis, Stergioulas & Vourdas, 2000). New filtering techniques and adaptations, such as singular spectrum analysis, Winger filtering have been introduced (Alonso, Del Castillo & Pintado, 2005; Erer, 2007; Georgakis, Stergioulas & Giakas, 2002; Giakis *et al.*, 2000; Ismail & Asfour, 1999) which appear promising but are not yet widely used in biomechanics research.

Typical cut-off frequencies for filtering kinematic data in sprinting biomechanics research are around 8 – 20 Hz, similar to the 6 – 18 Hz reported for kicking research (Numone *et al.*, 2006). These may be too low to adequately describe the movement characteristics of the lower limb and foot during the stance phase of sprinting, and there is little evidence in the research, such as power or harmonic analyses, to justify this choice.

The magnitude of impact forces during locomotion has been quantified using force platforms while researchers have used accelerometers to monitor the shock experienced by the lower limbs during locomotion (Lafortune & Hennig, 1991). Lake and Greenhalgh (2007) reported high frequency components in the 15 – 55 Hz frequency range for shank angular velocity, measured from accelerometry. This agrees with Digby, Lake and Lees (2005) who measured tibial rotation during running impacts, highlighting that, typically, authors will filter out these components of the signal using a low-pass cut-off frequency of 15 Hz or less.

Inverse dynamic analyses require both kinematic and kinetic inputs which originate from two different systems. Typically kinetic data from force platforms are low pass filtered with a cut-off frequency of around 100 Hz;

however, it has been suggested that the same cut off frequency be used for both kinematic and kinetic data to obtain accurate joint moments for fast transients such as impacts (Bisseling & Hof, 2006). They found filtering accelerations at 20 Hz suppressed high frequency components and when this precaution (filtering both kinematic and kinetic data the same) is not taken, the impact peaks in joint moments upon landing can be considered artefact.

Overall, there are numerous options for reducing the noise levels in collected data; the choice of a suitable method must be based on the research question and variables of interest. The effects of filter-induced errors on resultant joint moments, for example, are difficult to assess, as true intersegmental forces are never known. However, the biomechanist needs to be aware of how the smoothing processes and choice of cut-off frequency affect data so that an appropriate method can be applied without distorting the true signal. For sprinting, in order to obtain the most accurate resultant joint moment time histories during stance, as well as applying the same cut-off frequency to both kinematic and kinetic data, it has been suggested that the cut off frequency should be as high as possible (Bezodis *et al.,* 2011).

2.6.3 Errors in Inverse dynamic analyses

Inverse dynamics is a fundamental and commonly used computational procedure for the analysis of human movement. With inputs consisting of kinematic data, ground reaction forces, and centre of pressure and anthropometric information, resultant torques are then calculated at the various joints (Winter, 1983). These provide insight into the functions of muscle and the coordination of human movement and have been used to examine a wide range of sporting activities as well as extensively in sprinting. The value of such an analysis was stated by Winter (1980) who claimed one of the most valuable biomechanical variables to have for the assessment of any human movement is the time history of the moments of force at each joint.

Errors can stem from a variety of sources in segmental parameters: noise in surface marker movement and ground reaction force measurements (discussed previously); inaccuracies in locating joint centres (inappropriate joint models); inaccuracies in estimating the centre of pressure location; inaccuracies in inertial properties; and, finally, inaccuracies caused by skin movement artefact (also previously discussed) (Riemer, Hsiao-Wecksler & Zhang, 2007). From these various sources, error is typically due to inaccuracies in the coordinates of anatomically located markers, the approximations of accelerations, as well as force plate uncertainties, but can be reduced using appropriate signal processing techniques (Rao *et al.*, 2006).

Riemer *et al.* (2007) reported relatively large uncertainties in joint torques in walking gait which were mainly influenced by inaccuracies in segment angles, associated with skin motion artefact. Therefore, they suggested the development of a corrective method to compensate for skin movement artefact, such as the cluster method and global optimization methods. The use of bone pins to remove the problem of skin motion artefact has previously been discussed. However, this does not prevent errors in inverse dynamics as the soft tissue mass is a significant part of the segment, and its accelerations remain unknown if only bone movement is measured (Van den Bogert, 1994). The errors caused by non-rigidity can be minimized, however, by a careful selection of marker locations to measure movements and accelerations that represent a suitable compromise between bone and soft tissue (Van den Bogert, 1994).

Challis and Kerwin (1996) stated that for an elbow flexion movement, the estimated joint moments were most sensitive to uncertainties in the derivatives of the position data, errors in the inertial properties were small, and variation in the elbow joint centre also had little relative influence over the resultant joint moments. The assumption of rigidity was not included in their quantification of uncertainties in resultant joint moments. The influence of body segment parameter values is more controversial during gait (Rao *et al.*, 2006), although the magnitude of the effect of segment parameter error on gait kinetics was reported by Pearsall & Costigan (1999) to be generally less than 1% of body weight during walking. It is, therefore, also expected that the influence of body segment parameter values for the foot, in particular the toes, would have no or minimal effect on inverse dynamic errors. Researchers have used various new techniques, such as DEXA, to obtain individualised and more accurate body segment parameters (e.g. Durkin, Dowling & Andrews, 2002), but they have not included foot segments.

Inaccuracies in the lever arm calculation (from either inaccurate centre of pressure or centre of joint location) are one source of error in inverse dynamic calculations. Firstly, foot placement plays a role in determining the GRF lever arm during the push off of walking/running/sprinting (Erdemir & Piazza, 2002). Bosjen-Moller (1978) stated that the use of the transverse axis passing through the first and second MTHs was assumed to produce a higher GRF lever arm in sprinting, whereas the oblique axis limited the GRF lever arm. Shorten, Eden & Himmelsbach (1989) implied modulation of the lever arm for different walking speeds, although Viale, Belli, Lacour & Freychat (1997) found that foot position did not affect the lever arm in running.

The lever arm for sprinting has not been researched. In determination of the correct lever arm, an accurate measurement of the centre of pressure is vital. The calculation of the centre of pressure from force platforms (dividing forces by forces) is sensitive to noise when forces are small, and, therefore, typically inaccurate at the beginning and end of stance (Nigg & Herzog, 2007) as well as towards the edges of the plate, particularly outside the central region bounded by the four force transducers (Bobbert & Schamhardt, 1990). In fact, errors up to ± 30 mm have been reported by the manufacturers (Kistler, 1993), although the magnitude of the error depends on the load distribution (Middleton, Sinclair & Patton, 1999). Bobbert and Schamhardt, (1990) reported average errors of 3.5 mm in the x (short axis) and 6.3 mm in the y (long axis) direction. It is also reported that errors of ± 10 mm in the point of force application cause, on average, 14% changes in maximum joint torque in the lower extremities (McCaw & DeVita, 1995). These errors do not arise from cross-talk between transducers, non-linearity of individual transducers, but rather from transducers not being loaded exactly at their centres due to bending of the top plate (Bobbert & Schamhardt, 1990). Numerous correction formulae have been presented in the literature, including Schmiedmayer and Kastner (2000) for single point loading.

Stefanyhsyn and Nigg (1997) also used plantar pressure sensing insoles to allow the distribution of the GRF to the two foot segments, rearfoot and phalanges, in the vertical direction but not the horizontal direction. The pressure insoles allowed the division at the phalanges and rearfoot. The MP moment was then calculated using the vertical force measured on the phalanges by the pressure insole (100 Hz) and a percentage of the total horizontal force (measured by the force plate) which was equal to the vertical force For example, if 10% of the total vertical force acted on the phalanges, it was assumed that 10% of total horizontal force also acted on the phalanges. The moment about the MP joint, due to the vertical force on the phalanges, was determined by summing the individual moments created by the forces applied to each sensor. There was no description of how the moment arm was calculated, assumed to be from the pressure insoles' centroid of pressure. The difference in maximal MPJ moment was 3% during running and 7% during sprinting. Therefore, the method chosen was to use a single ground reaction force to determine net moments at the MPJ.

2.6.4 Experimental design and variability

Reliability of sprinting biomechanical variables

Reliability refers to the repeatability of a measurement. Better reliability implies better precision of single measurements and better tracking of changes in measurements in research or practical setting (Hopkins, 2000). Studies examining the reliability of sprint running have been relatively limited, despite the importance of the knowledge of the degree of variability for experimental work in indicating which measures can confidently detect small changes in an athlete's performance. Investigations have indicated that many basic temporal-spatial and ground reaction force variables of sprinting show relatively low intra-individual variability, and, therefore, a large number of trials may not be required to obtain stable and reliable data (Bradshaw, Maulder & Keogh, 2007; Hunter, Marshall & McNair, 2004b; Mero & Komi, 1986) However, these studies did not provide conclusive evidence and were based upon sprint acceleration. Salo & Grimshaw (1998) reported mean coefficient of variations of less than 10% for 23 and 21 kinematic variables, for females and males respectively, out of 28 variables measured, for sprint hurdling.

Hunter *et al.* (2004b) reported for all 33 sprinting kinematic and GRF variables reported, reliability improved notably when the average score of multiple trials was used. When using the average score of five trials the ability to detect a change is improved considerably, although after three maximal sprint trials fatigue might add additional variability to the measurements. Korhonen *et al.* (2010) collected maximal speed running data for two 30 m and two 60 m sprint trials from 18 young highly-trained male sprinters. Firstly, they showed that athletes achieved 99-100% of their maximal sprinting speed in the 30 m to 40 m distance used for force platform measurements (9.5 ± 0. 4 $m \cdot s^{-1}$). All temporal-spatial, vertical and resultant GRF variables were deemed reliable to identify small changes in an athlete's performance with Coefficients of variation (CV) of less than 6%; however, the vertical loading and horizontal GRF variables demonstrated larger variability (CV>10%). There was a variable specific symmetry between legs, suggesting that measuring just one side of the body, with the assumption that similar results would be obtained for the contra lateral side, may not be fully justified. Targeting-, velocity- and fatigue- induced variability, which may arise when a single force platform with multiple maximal sprint trials is used, was reduced by a small number of trials and a unique long force platform system, (Abendroth-Smith, 1996; Hunter *et al.*, 2004b). In a previous study with a single force plate method seven to eight sprints were needed to obtain four to five successful ground contacts for one side (Hunter *et al.,* 2004b).

Despite some recent studies on joint kinetics during sprinting, including individual analyses (Bezodis *et al.*, 2008; Bezodis *et al.,* 2009), there have been no studies that have reported reliability for joint kinetic measurements (moments and powers) during sprinting, possibly due to the high number of participants and trials needed for reasonable precision for estimates of reliability (Hopkins, 2000). However, for running, high intra-subject (DeVita & Skelly, 1990) and inter-subject (Simpson, 1988) variability of lower extremity joint moments have been reported, suggesting that individual adaptations in running technique may exist.

Experimental design

The appropriateness of the study design and data collection set-up must be considered, in terms of internal and external validity. There are numerous threats to the internal and external validity of biomechanics research (Knudson, 2009) and the biomechanist often has less control over the environment. Usually data collected in the lab are more accurate and reliable, but the validity can be substantially reduced (Schwameder, 2008). Often it is not possible to collect biomechanical data from a competitive setting; therefore, training or lab sessions during the competitive season are used as an alternative, with the mode / specificity of the task and calibre of athletes important in ensuring study reliability. In general, sports biomechanics research is often performed with relatively low numbers of subjects and trials. Some sprinting studies use as few as two successful trials (Bezodis *et al.,* 2009; Morin *et al.*, 2011). Hopkins, Hawley & Burke (1999) argue that the majority of previous studies in sports performance enhancement have been deficient on one or more of the following counts: the sample size was too small for adequate precision of the estimate of performance enhancement; the performance test had questionable validity; the athletes were not of a sufficient calibre; and their behaviours were not representative of training or competition. They do, however, state that a performance enhancement as small as 0.3 to 0.7 of the coefficient of variation is important for the best athletes. However, to detect the smallest worthwhile effect large sample sizes are needed, often beyond the resources of most sports scientists. Moreover, the reporting of accuracy, reliability, error estimation, and validity are relatively sparse in biomechanics research papers, although these aspects are very important to prevent the researcher, coaches and athletes from misleading or misinterpreting collected data (Schwameder, 2008). The researcher must consider that the data collected is accurate and reliable and useful for addressing the specific research question.

2.6.5 Single subject analyses

Some researchers have been unable to detect significant differences in conditions based on the high variability of biomechanical measures in running within groups (for example Logan *et al.*, 2010). High inter-subject variability may reflect the adoption of different unique strategies or response patterns to accomplish one common task (Bates, James & Dufek, 2004). In group designs, this form of variability violates the homogeneity of variation assumption, compromising the validity of the data and cannot be accommodated statistically (Bates *et al.*, 2004). The result of the group analysis is often support for the null hypothesis which may not be a correct conclusion for some or all of the individual subjects.

Group based analyses can identify general trends and mean differences in an attempt to generalise to a wider population. Group statistics have the advantage that the error, as well as variability among individuals, could be averaged out of the results if the data set was large enough. Consequently, the emphasis of group experimental designs focussed on the mean, with the application of average findings and average individuals to make comparisons among and between groups of individuals (Bates, 1996). In some experimental data, however, it has been shown that the behaviour of the average performer or subject is not, in fact, representative of any of the individual subjects' performances (Dufek, Bates, Stergiou & James, 1995). Alternatively, substantial biomechanical differences may be masked as a result of grouping subjects. For example, Dufek *et al.* (1995) revealed a number of different strategies utilised by performers in response to perturbations during impact activities.

An alternative, or a supplement, to group analysis exists with single subject analysis. Single subject analysis has been used to examine individual athletes' sprinting technique, providing detailed biomechanical analysis and feedback to aid coaches and athletes (Bezodis *et al.*, 2009). It has, however, primarily been used and advocated for the investigation of individual responses / strategies, when there are different solutions to the same task by individual subjects. The basic rationale for single subject evaluation is that individuals are unique, and will demonstrate variability in human movement due to mechanical, morphological and environmental constraints (Bates *et al.*, 2004). Individuals will select different strategies for the performance of a motor task, based on the perceptions and experiences of the performer. Experimental evidence in support of individual strategies can be found in the literature, such as Lees and Bouracier (1994), who found evidence of a 'movement pattern fixation' amongst experienced and inexperienced runners during a longitudinal evaluation, explaining that the subjects may be able to select a method of running from a pool of solutions which suit the particular requirements on the particular day of testing. Demonstrating differing individual responses has implications for the testing of sports equipment and footwear. In the case of manipulating a footwear condition, for example, the subject's response will depend on his/her recognition/perception of the perturbation, which in turn will be a function of past experiences (Bates, 1996).

These types of strategy response have been observed for impact forces (Nigg *et al.*, 1987) and landing (Dufek & Bates, 1990; Dufek *et al.*, 1995). Dufek and Bates (1991) advocate the use of single subject analyses for identifying shoe differences and assessed dynamic performance characteristics of four different shoe models (two basketball, one volleyball and one running shoe) during landings, using GRF data. Data was analysed using both group (repeated measures ANOVA) and single subject analyses (Model Statistic). The group analysis revealed a preferential performance rank order but no consistent trends between the impact forces for all subjects. As such, the authors did not generalise to a 'best shoe'. Single subject analyses demonstrated that individuals elicited unique rank orders, although some shoes appeared better for more individuals than others. The authors concluded that even when a high number of trials were used (n=25), as well as both a group and within-subject analysis procedure, whether or not the observed differences are biomechanically meaningful remains unknown. Bates and Stergiou (1996) also combined both group and single subject statistical analyses when investigating the effects of shoe hardness (three stiffness conditions) on impact force and maximum knee flexion angle during running. Once again, the group ANOVAs provided relatively little insight into the research question with just two significant differences found. The individual analyses demonstrated that ten subjects exhibited significant differences amongst all three conditions. The results supported the response strategy continuum with 'some demonstrating a greater Newtonian or mechanical component and others showing greater neuromuscular accommodation' (Bates *et al.*, 2004, p.23).

Criticisms of the single subject approach have focussed on the statistical assumptions as well as lack of generalisation of results. As with group analyses, violation of the normality distribution should be checked and acknowledged. Trial independence for single subject measurements has also been challenged, although it has been argued that there should be no more dependence for a single variable among trials produced by a given individual than for trials generated by different subjects (Bates, 1996; Bates *et al.*, 2004). Therefore, the appropriate approach for single subject analysis is to assume the trial values to be independent and use the corresponding independent test. A wide range of analysis techniques are available for single subject analyses, including non-parametric techniques, 'bootstrap' or 'randomization' procedure, the commonly used Model Statistic Procedure (single subject t-test approach), 'Fischerian' techniques and multiple regression techniques (Bates *et al.*, 2004). As with group analyses, the importance of the correct statistical method, as well as some attention to any natural trends in data, is paramount to true interpretation of the data (Reboussin & Morgan, 1996).

The results from a single subject design provide little useful information on the general utility of an intervention or condition without the addition of assumptions for which the data was collected (Reboussin & Morgan, 1996). Multiple single subject analyses, however, can be performed and, whilst difficult to analyse, will provide at least some data to suggest the consistency of any intervention effect. To assess the effect of an intervention there are different study designs available, such as the simple AB, the more complex ABAB, and multiple replicates. As the number of trial repetition increases, the ability of the researcher to evaluate assumption and potential influence of the intervention increases. Using an ABAB approach, condition A is thought of as a control or baseline and condition B is some active condition or intervention added by the researcher. Repeated observations are made, with changes in the experimental condition taken as evidence of the effect of the intervention. Indeed, the importance of the use of a repeated baseline in biomechanical studies has been highlighted and supported with evidence in the literature from running / jumping obstacles (Stergiou & Scott, 2005). The authors observed differences between kinematic and kinetic baseline measurements during running, concluding that the control condition may be stable over time; although the repeated baseline has not been deemed necessary for walking studies (Revill *et al.*, 2008).

2.6.6 Summary

Whilst group based analysis may provide a good starting point to investigate the effect of a condition on a group of sprinters, it is important to consider that sprinters adopt different sprinting techniques, especially in terms of power generation and foot contact patterns. In particular, it is also likely that individual sprinters may respond to different footwear based upon past experience and individual preferences. Therefore, data should be analysed on an appropriate level, to be able to demonstrate and analyse individual strategies. This section has also demonstrated the importance of equipment considerations, along with data collection and processing procedures. When resultant joint kinetics are required to assess a specific research question, such as the role of the MPJ during sprinting, the accuracy of the kinematic and kinetic input data is paramount.

2.7 Chapter summary

This chapter has reviewed relevant literature in order to determine the current state of the body of knowledge of the biomechanics of foot function in relation to sprinting performance, and to highlight key findings from previous research. The foot is often neglected in sprinting research. For example, researchers who have combined both kinematic and kinetic data to calculate the internal joint kinetics to understand the sources of power production during sprinting have simply modelled the foot as a single segment, ignoring any influence of the MPJ to the energetics of sprinting. Areas of limited knowledge and ideas for future research have been identified in this review. This includes the need for a simple multi-segment foot model for the analysis of foot function during athletic tasks. The present knowledge base is somewhat limited by oversimplification of the modelling of the joint in sports performance research. Furthermore, the need for obtaining both kinematic and, in particular, kinetic data for the MPJ, is vital to ascertain the function of this joint in sprinting. The relationship between MPJ kinematics (e.g. angular range of motion) or energetics (e.g. power production during push-off) and sprinting performance remains unknown. There is a large gap in the current literature for the quantification of typical plantar pressure patterns during sprinting, starting with loading profiles at barefoot sprinting at the foot / ground interface. Finally, there is also a need to investigate the potential effect of shoe construction characteristics on the energetics about the MPJ. Although some researchers have stated that sprint performance may be improved by increasing the stiffness of sprint spiked footwear, there is no substantial evidence of any changes in foot behaviour, nor the mechanism responsible for this improvement as very little kinetic data for the MPJ during sprinting is available.

The review of biomechanical data collection and analyses procedures revealed that current data processing procedures employed by the majority of researchers may not, in fact, be accurate enough to describe high speed transients that are experienced during fast movements, such as foot contact during sprinting. Finally, due to the inherent nature of biomechanical research designs and natural variability in human responses, in order to analyse small biomechanical or performance changes due to footwear conditions, alternative experimental analyses procedures may provide more insight than traditional group based analyses.

CHAPTER 3: MEASUREMENT PROCEDURES AFFECT THE INTERPRETATION OF METATARSOPHALANGEAL JOINT FUNCTION DURING SPRINTING

Overview: Accurate measurements of MPJ joint motion and energy contributions during sprinting are essential for investigating the role that footwear has on sprinting performance. This study demonstrates that MPJ kinetic calculations are highly sensitive to errors in both the anatomical representation of the MPJ line and the processing of the kinematic and ground reaction force data. Based on the results of this study, it is recommended that the MPJ axis be represented as an oblique axis from MTH1 to MTH5. Both a high sample rate and low-pass filtering cut-off frequency for kinematic and kinetic data, are necessary for accurate assessments of high speed impacts and movements, as evident in sprinting.

3.1 Introduction

The metatarsophalangeal joint (MPJ) has been shown to be a large dissipater of energy during the stance phase of sprinting. Stefanyshyn and Nigg (1997) presented kinematic and kinetic characteristics of the metarsophalangeal joint (MPJ) motion during sprinting for five male sprinters wearing their own sprint spikes. However, there remains a lack of biomechanical data evaluating the function of this joint during sprinting. Measurement procedures presented in the research (Oleson *et al.*, 2005; Stefanysyn & Nigg 1997; 1998b; 2000; Toon *et al.*, 2009) have likely oversimplified the motion analysis of the MPJ and therefore, there still is a need to accurately analyse the function and role of the MPJ. Firstly, assuming a perpendicular axis (as used by Stefanyshyn & Nigg, 1997, 1998b, 2000) to represent the five MPJs may lead to differences in resultant joint kinetics, in comparison to an oblique axis definition (as Oleson *et al.,* 2005) or dual axis approach (as a two-gear system was suggested by Bojsen-Moller, 1978). Secondly, low kinematic sampling rates (100-200 Hz) along with filtering data with low cut-off frequencies (F_c = 8-20 Hz) may not be adequate to capture the rapid motion of the foot during ground impact during high-speed activities such as sprinting. If high frequency components of the motion are present, then filtering the data with a low cut-off frequency may distort the curves through over smoothing and MPJ peak angular velocities would be underestimated. Consequently, higher filter cut-off frequencies must be used, otherwise calculations of joint kinetics during fast movement transients may be inaccurate (Bisseling & Hof, 2006).

3.2 Objectives

The research question to be addressed in this study is:

What influence do methodological issues have on the analysis of MPJ function during sprinting?

This study was designed to explore whether previous simplifications of the MPJ joint axes and the filtering of the kinematic data may misrepresent joint kinetics during sprinting. It was expected that the moment arms necessary for joint kinetic calculations will be influenced by the definition of the joint axis used by investigators. Stefanyshyn and Nigg (1997, 1998b, 2000) defined the MPJ by a single lateral marker on MTH5 and assumed the joint to lie perpendicular to the sagittal plane. However, the more anterior location of the medial metatarsal heads would shorten the lever arm of the ground reaction force and, therefore, reduce the calculated MPJ moment. The simplified, perpendicular approach may therefore lead to an inaccurate assessment of the amount of work performed at the MPJ and the role of this joint to the energetics of the task of sprinting. A more anatomically appropriate joint representation, such as an oblique or dual axis definition, therefore may be more suitable for calculating MPJ kinetics. Hypothesis 1 of this study states that a more anatomically appropriate joint axis definition would significantly reduce the resultant joint kinetics, in comparison to a perpendicular approach. Hypothesis 2 states that filtering with low cut-off frequencies would significantly underestimate MPJ motion transients (segmental displacement and velocity data). This is based upon suggestions that for other types impacts, where high frequency movement transients are present, for example the impact phase in running and ball impact during kicking, typical low sampling rates and cut-off frequencies may have been insufficient to adequately capture motion.

Specifically, the objectives of the study were:

1) To determine the effect of two key methodological issues; MPJ axis definition; and data processing techniques, on the assessment of MPJ function during sprinting.

2) To develop an appropriate methodology and MPJ axis definition to accurately assess the mechanical energy contribution of the MPJ to sprint running

3) To determine typical MPJ behaviour and function during sprinting for a group of sprinters, including the kinematic, kinetic and energy characteristics of the MPJ and compare to previous researchers

3.3 Methods

3.3.1 Participants

Four competitive athletes volunteered to participate in this study; three female sprinters / hurdlers (mean age 22.1 ± 3.9 years, mean height 163.6 ± 6.0 cm, mean mass 63.7 ± 4.6 kg and mean 100 m best 12.4 ± 0.4 s) and one male decathlete (age 26.8 years, height 180 cm, mass 82 kg, 100 m best 11.1 s, decathlon best score 7500 points). Each subject also underwent a DEXA scan of the foot, for imaging purposes to identify individual's foot anatomy and orientation of the MPJ line. Informed written consent was obtained from all participants. The study was approved by the University's Ethics Committee, Approval number 0753 (see Appendix G).

3.3.2 Protocol

Four maximal sprinting trials were collected on each sprinter, wearing their own sprint spikes. The sprints were performed on a 55 m indoor runway with an indoor synthetic track surface and the athletes accelerated for 20 m before data were collected and were probably still accelerating during data collection. A customized starting mark was used to aid the athlete in striking the force plate without the need to alter their stride pattern prior to force plate contact. Timing gates were located 2.5 m either side of the force platform, therefore recording sprint times over 5 m as the athletes crossed the force platform.

Kinematic data were collected using a 6 camera Qualisys system (Pro-reflex MCU 1000 cameras, Qualisys Inc., Sweden) sampling at 1000 Hz. Kinetic data were simultaneously collected using a force platform (Kistler model 9287B, Kistler, Switzerland), also sampling at 1000 Hz. Stance phase of the left foot in contact with the force platform was chosen. Trials that landed towards the edges of the plate were discounted due to the high centre of pressure inaccuracies (Kistler, 1993, reported inaccuracies of Δ ax and Δ ay < ± 15 mm around the load cells).

Laboratory tests for the spatial synchronisation and calibration of the motion capture system and force plate were completed prior to data collection using the Caltester device and software (C-Motion Inc., USA), documenting the typical error for the centre of pressure (see Appendix A).The location of the force plate was determined and the transducer origin values were amended within the Qualisys Track Manager Software (Qualisys, Sweden) to include the height of the track surface above the surface of the force platform (15 mm).

From previous single subject pilot work, a three segment foot model (forefoot, midfoot and rearfoot) was developed. The forefoot segment was defined similarly to Oleson *et al.* (2005), who used markers on the 1st and 5th metatarsal heads (MTH1 and MTH5) and the 2nd toe. 12 mm Diameter reflective markers were placed on the medial and lateral malleoli (later removed for sprint trials), the posterior, medial and lateral heel, the bases of the 1st and 5th metatarsals, the heads of the 1st, 2nd and 5th metatarsals (MTH1, MTH2 and MTH5) and finally on the head of the second toe (distal end of the toe box). Markers were placed on the sprint shoe for shod conditions, on the medial and lateral sides of the metatarsal heads and bases and for MTH2 and the second toe superior to underlying landmark, as shown in Figure 3.1. The five MP joints were considered as a single joint, rotating about an oblique transverse axis defined by the MTH1 and MTH5. The forefoot was defined by markers on the second toe, MTH1, and MTH5 and the MTH2 was used for a tracking marker only for this segment. The midfoot was defined by MTH1, MTH5, metatarsal base 1 and 5. The local coordinate system of the segments was aligned to laboratory coordinate system (with X – medio-lateral, Y – antero-posterior and Z – axial / vertical axes as demonstrated in Figure 3.2). The MPJ angle was defined as a Visual 3D joint angle between two segments: the midfoot segment and forefoot (reference) segments with normalization relative to standing calibration and cardan sequence X-Y-Z.

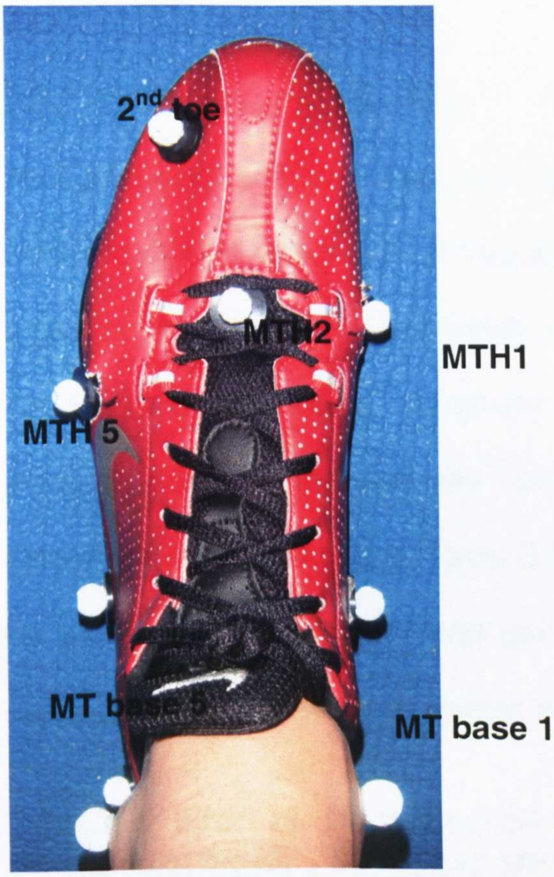


Figure 3.1. Image of the left foot demonstrating marker location and axes of the MPJ. For the first part of this study markers were placed onto the shoe upper.

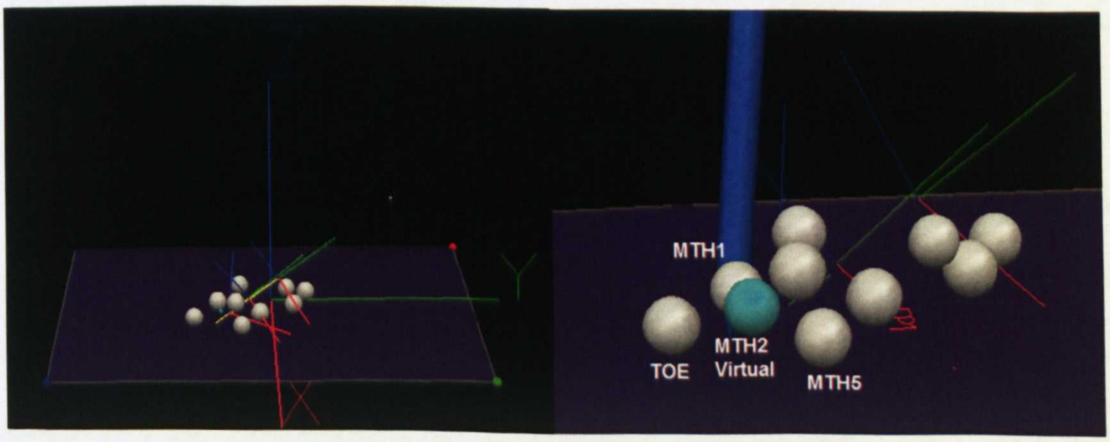


Figure 3.2 Screen shots of joint marker and axes location, from Visual 3D.

As in Stefanyshyn & Nigg (1997), the inertial effect of the phalanges was considered to be negligible. The focus of this study was to investigate only MPJ kinematics and kinetics, the other segments were not used in this investigation.

3.3.3 *Pilot work – choice of Fc*

Joint positional and force data were smoothed using a fourth-order low pass Butterworth filter with a cut-off frequency (F_c) of 100 Hz. This F_c was chosen based upon pilot work during which a range of cut off frequencies from 50 to 100 Hz were investigated. Kinematic curves were visually inspected and joint moments, powers and energies using F_c s = 50, 60, 70 and 100 Hz were calculated and are shown in Table 3.1 for one typical trial. Even using F_c = 70 Hz joint energy absorption and generation were underestimated indicating signal power loss in the joint motion data at high frequencies.

Table 3.1. Effect of different cut-off frequency on the energy absorbed and generated at the MPJ throughout stance. Data sampled at 1000 Hz.

Cut off frequency (Hz)	Energy absorbed throughout stance (J)	Energy generated throughout stance (J)
50	-26.06	2.89
60	-26.07	2.87
70	-26.12	2.92
100	-26.28	3.06

Figure 3.3 demonstrates the MPJ power throughout stance for one typical sprint trial. High frequency components of the signal occurred at touchdown, due to initial oscillations in angular velocity as the foot absorbed the load. These were suppressed when a cut off frequency lower than 100 Hz was used. The 1000 Hz signal was also filtered a different way: F_c = 100 Hz for the first 30% stance (in order to retain the high frequency signal at touchdown)

and then the remainder of stance was filtered using F_c = 50 Hz. Figure 3.4 demonstrates the MPJ power for one typical shod trial filtered using this approach. However, there was little difference in the energy generated and absorbed compared to using F_c = 100 Hz for the entire stance phase. Therefore the noise introduced by using F_c = 100 Hz did not affect the overall energy and F_c =100 Hz frequency was selected for use in the study.

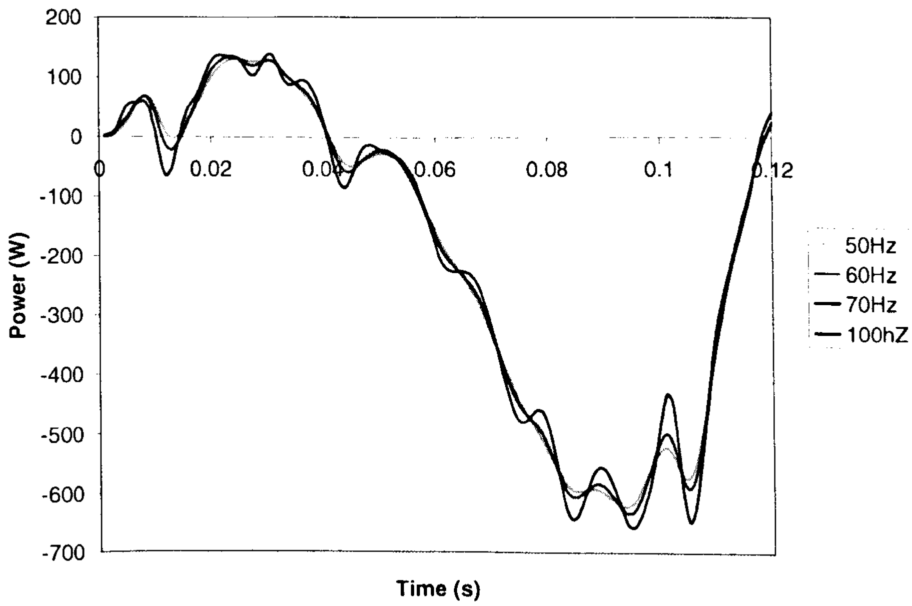


Figure 3.3. The effect of cut-off frequency on MPJ power. One typical shod trial sampled at 1000 Hz.

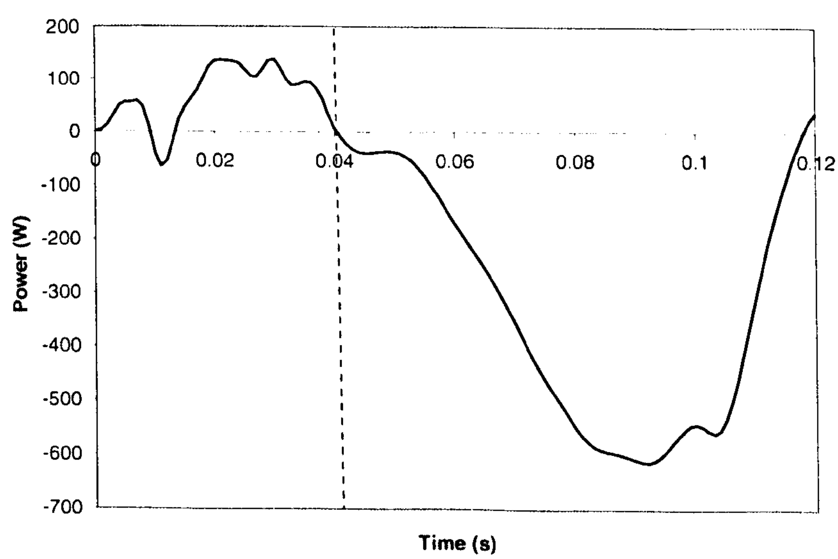


Figure 3.4 MPJ Power for one typical shod trial sampled at 1000 Hz, F_c = 100 Hz for the first 40ms of stance and 50 Hz thereafter.

During pilot work, spectral analysis was also performed in order to provide further justification of the choice of 100 Hz Fc to be used later in the study. For one typical barefoot trial (sampled at 1000 Hz) an analysis of the frequency spectrum (FFT) was performed on the MPJ angle, using 256 points and a spectral resolution of 3.96 (Figure 3.5). In order to closer examine the high frequency components of the signal, the data was also filtered using a high pass filter with F_c = 10 Hz. This attenuated the lower frequency signal (6-8 Hz) present (characterised by low frequency waveform of the MPJ curve) but retained the clearly predominant frequency around 15 Hz. There is clear signal power up to 30 Hz after which the signal power falls off, although there is some small signal power up to 80 Hz. This provides further evidence for selecting a 100 Hz cut off frequency in order to retain the high frequency components of the MPJ angle data which are evident at touchdown and then at the maximal degree and rate of MPJ flexion.

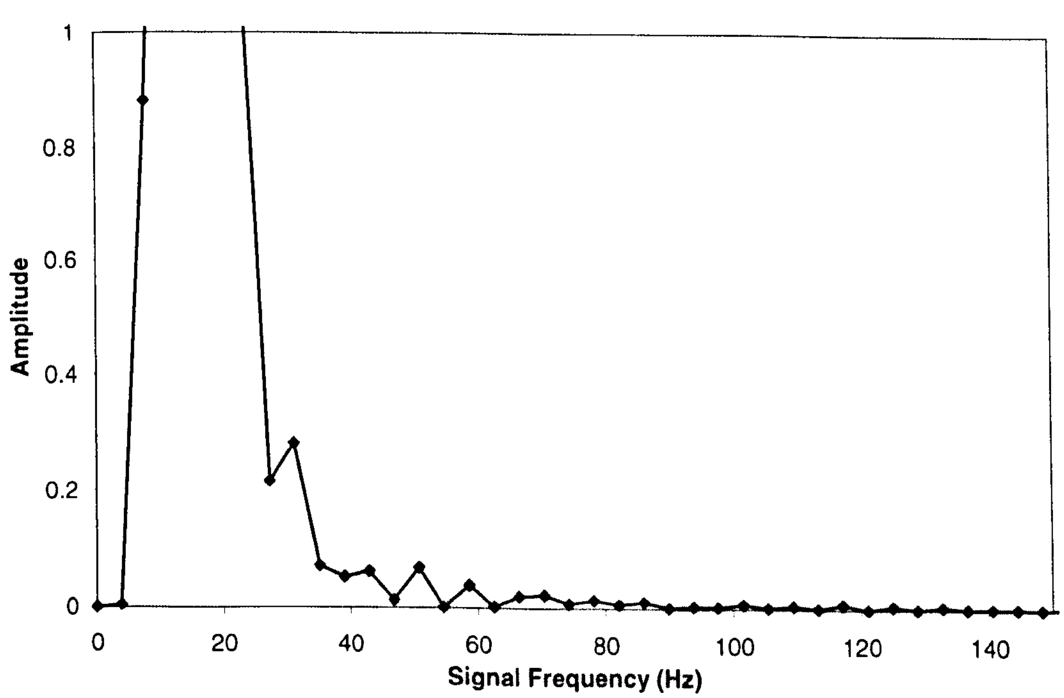


Figure 3.5. Spectral analysis of the MPJ angle for a typical barefoot trial. Data was filtered using a high pass filter with a 10 Hz cut-off so that the high frequency components could be highlighted.

Using the same cut-off frequency for kinematic and kinetic data when investigating high speed movements / impacts is recommended by Bisseling and Hof (2006), thus Fc = 100 Hz was used for both kinematic and kinetic data.

3.3.4 Comparison of Fc

To compare the previously chosen F_c = 100 Hz data to typical processing procedures in the research the data (thereby addressing *Objective 1 of the study*), data were also filtered using Fc = 8 Hz for kinematic data and 100Hz cut-off frequency for determining joint kinetic data. The F_c = 100 Hz was chosen based upon pilot work mentioned above. F_c = 8 Hz was chosen as a typical processing approach as it was used by Stefanyshyn & Nigg, (1997; 1998b; 2000), the largest contributors to research into the role of the MPJ during sprinting. Other authors in sprinting have also used low F_c's, such as: Hunter *et al.* (2004a; 2004c) used F_c = 7 – 12 Hz; Gittoes & Wilson used F_c = 15 Hz, despite more recently other authors have used some higher Fc's (such as F_c = 24 Hz used by Bezodis *et al.*, In press).

During additional pilot work, data were also resampled to 200 Hz then filtered at F_c = 8 Hz (exactly replicating the sample rate and processing of Stefanyshyn & Nigg, 1997, 1998b) however, besides the number of data points, there was little difference in the resultant curves from data sampled at 1000 Hz and filtered at Fc = 8 Hz, as it was the low cut-off frequency that dramatically affected the shape of the curves. Therefore it was decided that data would not be resampled for the study.

3.3.5 *Centre of pressure*

The coordinates of the centre of pressure (CoP) are typically inaccurate for small forces at the beginning and end of stance (Nigg, 2007), so to minimise errors in the CoP data, CoP thresholds of 100 N and 50 N were used at the start and end of ground contact respectively. Beyond these thresholds the CoP was distorted and in a position outside of the forefoot, due to low loading on the force platform. This was confirmed with the Caltester laboratory test results, a threshold of >100 N was needed at the initial ground contact phase, to avoid high errors in CoP coordinates at the start of the movement. For every trial, the CoP coordinates were plotted and visually inspected. At the end of the movement, towards take-off, CoP data was eliminated for the few frames where there was clear distortion in the accuracy, associated with larger shear forces relative to increasingly smaller vertical forces (see Appendix A).

3.3.6 *Kinetic analysis*

Joint moments were calculated according to Winter (1983) and the analysis assumed the resultant forces and moments at the MPJ were zero until the GRF acted distal to the joint. This was based on the assumption that the inertial effect of the phalanges was negligible (Stefanyshyn & Nigg, 1997). The MPJ moment therefore resulted from the ground reaction forces acting distally to the MPJ line and was calculated using the equation:

$$Fz * dx + Fy * dy,$$

Where Fz = vertical ground reaction force, Fy = anterior-posterior ground reaction force, dx = distance of the horizontal moment arm (CoP to MPJ axis), dy = distance of the vertical moment arm (CoP to MPJ axis). The horizontal (dx) moment arm calculated as the perpendicular distance from the x and y CoP coordinates to the MPJ line, a straight line through the x and y coordinates of MTH1 and MTH5. MPJ plantarflexor moments (defined as positive) therefore resulted from the ground reaction forces acting distally to the MPJ line.

Joint moments, powers and energies were calculated as computed by Stefanyshyn and Nigg (1997). MPJ plantarflexor moments were defined as positive. Joint power was calculated as the product of the net joint moment and angular velocity. Positive power occurs when the angular velocity of the joint is in the same direction as the moment, thus positive power occurs during a concentric contraction and negative power during an eccentric contraction. Energy was calculated by trapezoidal integration of the joint power curve.

The joint axis was then modelled a second way, to replicate the joint axis definition of Stefanyshyn and Nigg (1997). For this the MPJ was modelled using an axis perpendicular to the sagittal plane based upon MTH5 marker (Figure 3.6).

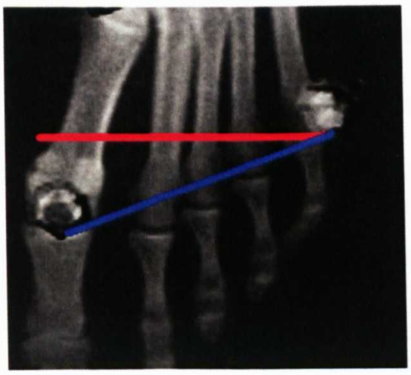


Figure 3.6. DEXA scan of the two joint axis definitions used for modelling the MPJ: oblique joint line from MTH1-5; and perpendicular joint line based on MTH5. Foot flat and lightly loaded.

3.3.7 *Alternative representation of the MPJ axis*

In order to further progress the axis representation of the MPJ, data from four additional participants was used in phase 2 of testing. Two female sprinters / combined eventers (aged 27 and 19 years, height 179 and 161 cm, mass 68 and 56 ± 5.6 kg, 100 m best 12.2 and 12.7 s) and two male combined eventers (both aged 20 years, height 188 and 189 cm, mass 84 and 76 kg and 100 m best 11.3 and 11.4 s) participated. All four participants underwent DEXA scans of the foot with the MPJ both flat and flexed (see Figure 3.7) in order to optimise the location of the markers relative to the underlying bones. The MPJ was modelled using a dual axis definition, with a transverse axes between MTH 1 and 2 and an oblique axis between MTH2 and MTH5. Figure 3.8 demonstrates all three MPJ joint axes definitions. To enable this dual axis joint definition, a virtual marker was used for MTH2. This was done by capturing the location of a pointer wand in the standing calibration trial and also performing a pointer compression trial. A virtual marker on MTH2 was then created in Visual 3D using known offset locations from three surrounding markers (1st MT base, MTH1 and MTH5). This landmark was tracked throughout the whole movement to enable the coordinates of MTH2 to be obtained, for the dual joint axes definition. Furthermore in this additional testing phase, holes were cut out in the sprint shoes and the markers were placed on the skin (Figure 3.8). The marker set and locations were identical to the first phase of testing, except for the use of the virtual marker on MTH2. All three MPJ axes definitions are demonstrated in Figure 3.8. Joint kinetics were calculated relative to the MTH1-2-5 joint axis. CoP data from the force platform was used to define which of the two joint axis were being used (MTH 1-2 or 2-5), when the medio-lateral coordinate of the CoP progressed from the lateral side to the medial side of the foot, beyond the MTH2, the joint axis was switched from MTH2-5 to MTH1–2. From inspection of the location of the CoP and the MTHs, it was evident that there was minimal in-toeing or out-toeing on the force platform therefore it was deemed acceptable to use the CoP as criteria for switching from the lateral to medial axis.

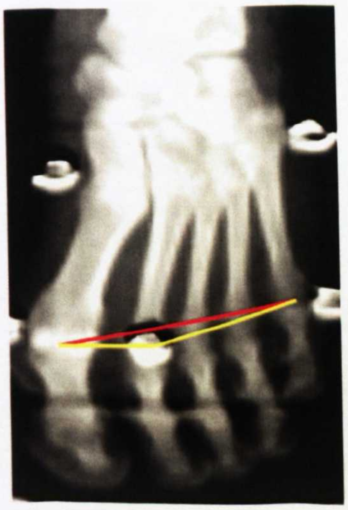


Figure 3.7. DEXA scan of additional male subject, with MPJ flexed to approximately 60 degrees. Lines represent two MPJ definitions: oblique axis (red line) and dual axis (yellow line).

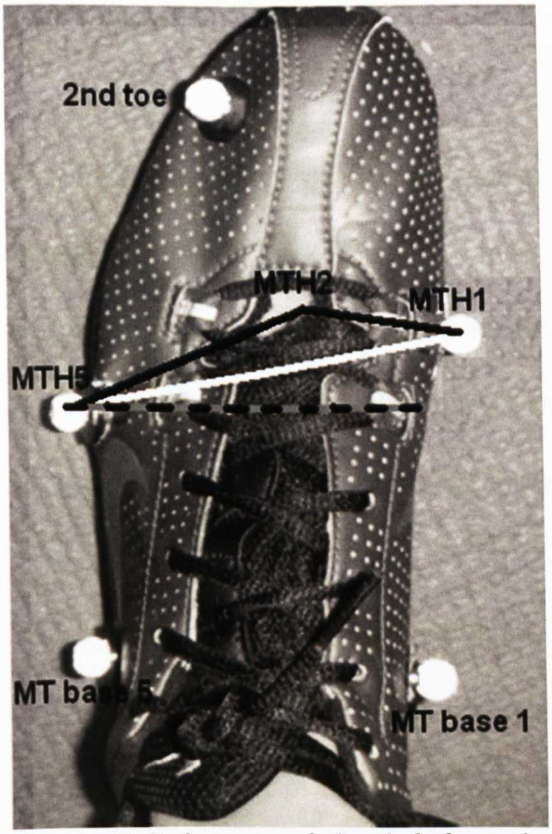


Figure 3.8. Image of the left foot demonstrating marker location and axes of the MPJ. Holes were cut out on the shoes to locate MTH1 and MTH5 and markers were on the skin overlying the medial and lateral sides of the metatarsal heads. The hole over MTH2 was used to create a virtual marker. Markers on the second toe and bases of metatarsals 1 and 5 were placed on the shoe. The dashed line represents the perpendicular axis, the white line represents the oblique axis and the black line represents the dual axis representation.

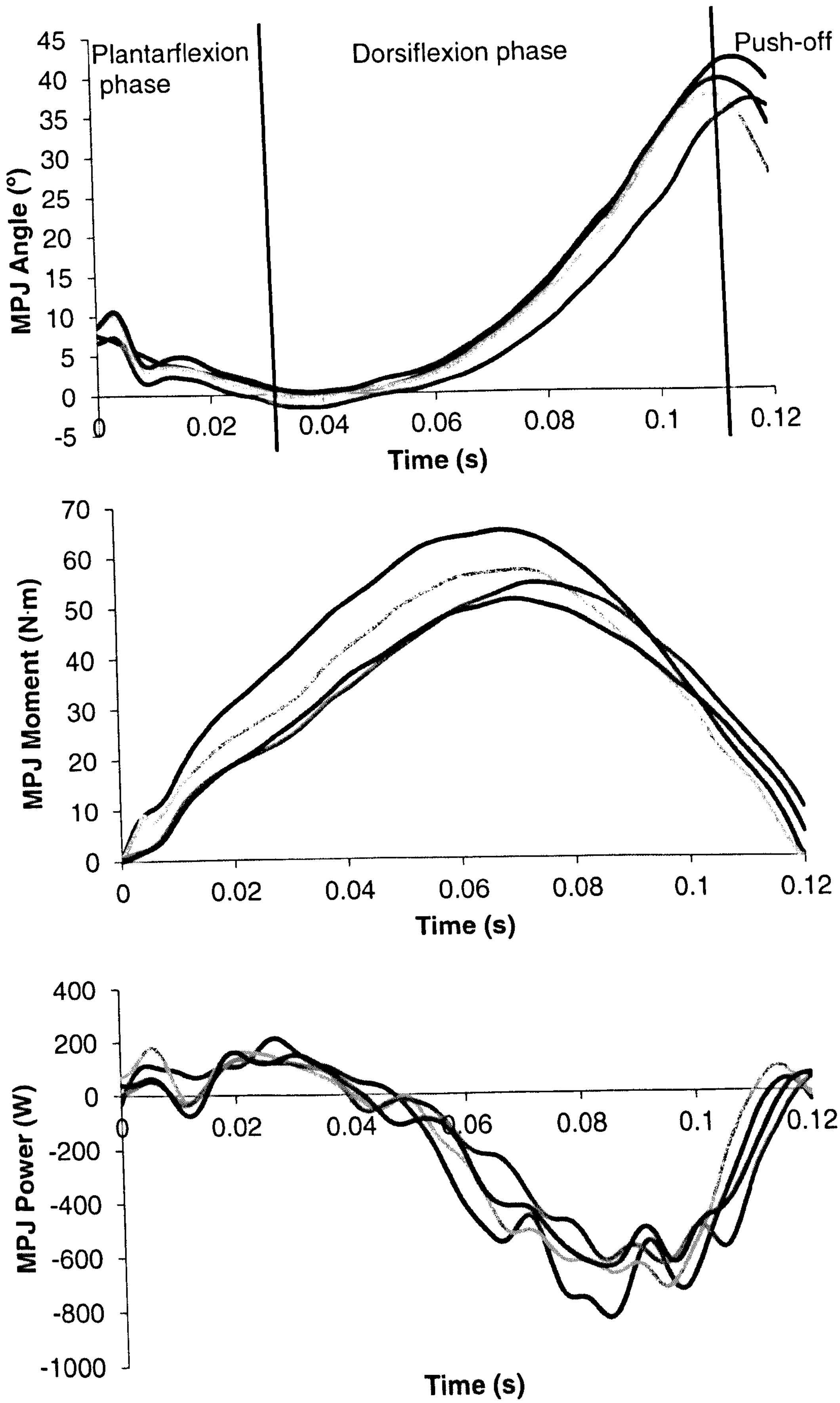
3.3.8 Statistical analysis

Where data were normally distributed, paired samples t-tests were performed to compare mean differences in MPJ kinematic and kinetic variables between different MPJ representations and processing approaches. A total of 7 out of 40 conditions analysed were not normally distributed, therefore Wilcoxon non-parametric tests were performed on this data. For both, the level of significance was set at $\alpha = 0.05$.

3.4 Results

The mean sprinting speeds in testing phase 1 were 7.2 ± 0.3 m/s for the three female sprinters and 8.5 ± 0.1 m/s for the male. In phase 2, mean speeds were 7.2 ± 0.1 m/s for the two females and 8.6 ± 0.1 m/s for the two males. These sprinting speeds were similar to those recorded by Stefanyshyn and Nigg (1997) at 15 m which ranged from 7.1 to 8.4 m/s.

The motion of the MPJ during ground contact was as follows and MPJ angle, moment, power and energy for one female participant is shown in Figure 3.9. Initial foot contact is executed with the forefoot. Immediately after touchdown the heel is lowered towards the floor and the MPJ plantarflexes typically during the first 40 ms of stance (plantarflexion phase). From 40 ms to 110 ms the heel rapidly rises and the MPJ dorsiflexes (dorsiflexion phase). The mean maximum degree of MPJ flexion was 35.6° (± 3.8°). Finally, the MPJ plantarflexes during the last 10 ms of stance (push-off phase) as the foot pushes off, however plantarflexion continues after take-off. The MPJ moment was plantarflexor throughout stance (mean peak moment 58.2m ± 11.1 N·m). For all four participants, positive power was produced shortly after touchdown, during the landing phase (190.6 ± 66.1 W) and energy was generated during the first 40ms (2.6 J ± 1.4 J). However, overall the MPJ was a large energy absorber, as the heel lifted and the MPJ flexed, 22.9 J (± 8.3 J) energy was lost at the MPJ. All four subjects produced power during the final push-off phase (111.8 ± 45.9 W) as the MPJ plantarflexed, however the MPJ did not fully extend until after take-off and little energy was generated (0.4 ± 0.4 J).


Plantarflexion phase
Dorsiflexion phase
Push-off
MPJ Angle (°)
45
40
35
30
25
20
15
10
5
0
-5
0
0.02
0.04
0.06
0.08
0.1
0.12
Time (s)
MPJ Moment (N·m)
70
60
50
40
30
20
10
0
0
0.02
0.04
0.06
0.08
0.1
0.12
Time (s)
MPJ Power (W)
400
200
0
-200
-400
-600
-800
-1000
0
0.02
0.04
0.06
0.08
0.1
0.12
Time (s)

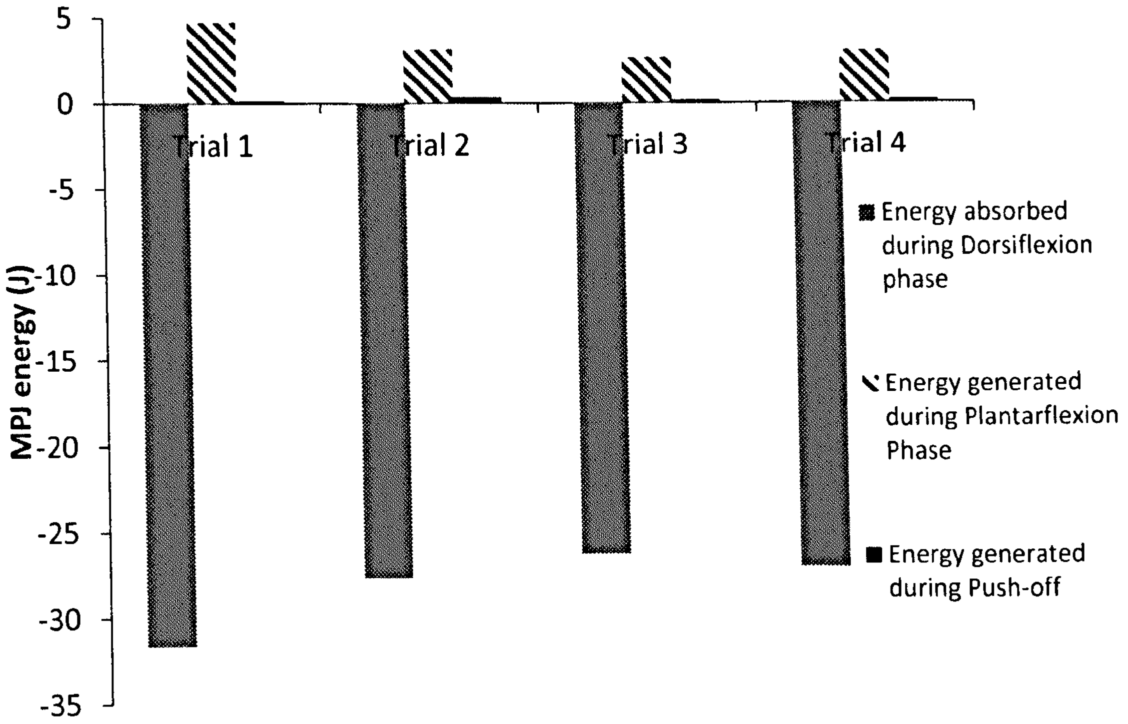


Figure 3.9 MPJ angle, moment power for one female participant, four sprinting trials.

3.4.1 Cut-off frequency

It is evident from Table 3.2 that using a typical cut-off frequency of F_c = 8 Hz significantly underestimated both MPJ angular range of motion and angular velocity compared to F_c =100 Hz. MPJ angular range of motion throughout stance was underestimated by approximately 15 degrees. Figure 3.10 clearly demonstrates the extent and rate of the MPJ flexion underestimation using Fc = 8 Hz. The F_c = 100 Hz data shows flexion of the MP joint at impact, followed by rapid extension with damping oscillation which was not present with the F_c = 8 Hz. Rapid MPJ motion and power production just after touchdown, were not present with the F_c = 8 Hz. Overall, energy absorption at the MPJ was underestimated by approximately 40% when smoothing at 8 Hz compared to 100 Hz (Table 3.3). A small amount of energy generated during push-off was only revealed when high frequency angular motion was included in the analysis (using the higher filter cut-off frequency).

Table 3.2 The effect of cut-off frequency on MPJ kinematics (n=4). * denotes a significant difference ($p < 0.05$) between F_c = 100 Hz and F_c = 8 Hz.

Joint axis Kinematics Fc	Oblique MTH1-5 100 Hz	Oblique MTH1-5 8 Hz
MPJ Range of motion (°)	36.3 (± 5.1)	21.5 (± 3.8)*
MPJ peak dorsiflexion velocity (°/s)	-1144.9 (±-707.7)	-438.6 (±-183.6)*

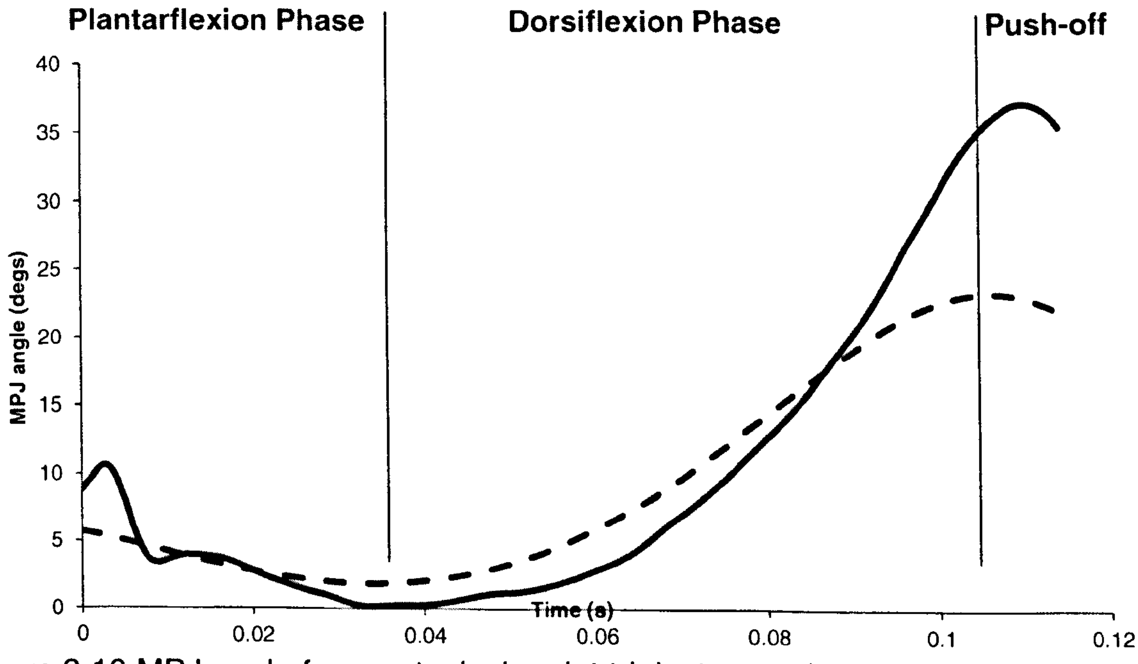


Figure 3.10 MPJ angle for one typical sprint trial, stance phase only, positional data filtered at two different cut off frequencies: F_c = 100 Hz (black line) and F_c = 8 Hz (dashed line). Vertical lines separate the plantarflexion phase, dorsiflexion phase and push-off.

3.4.2 Axis definition

The perpendicular axis definition resulted in greater values (approximately two times higher) for all joint kinetic variables. An example is illustrated in Figure 3.11. The higher moment and power for the perpendicular axis was due to an increased moment arm of the ground reaction force about the MPJ. For the four subjects, resultant joint moments were higher by on average approximately 81% or 47.7 N·m (± 21.2 N·m) and energy absorption was higher by on average by 89% or 12.4 J (± 8.0 J) for the perpendicular compared to the oblique joint axis (both filtered at 8 Hz). All joint kinetic data presented in Table 3.3 were significantly greater when using a perpendicular axis definition, in comparison to an oblique axis definition, with the exception of energy generated during push-off.

Table 3.3. Mean (± S.D.) MPJ moment, power and energy for four subjects (Phase 1). Comparison the oblique and perpendicular axis definitions as described in the text. * denotes a significant difference ($p < 0.05$) between the 2D axis definition, in comparison to the oblique axis definition. F_c = 100Hz.

Joint axis	Oblique	Perpendicular
Peak MPJ plantar flexor moment (N.m)	58.2 (± 11.1)	110.7 (± 18.7)*
Peak Positive Power (W) generated during MPJ plantarflexion	190.6 (± 66.1)	629.6(± 301.0)*
Peak Negative Power (W) generated during MPJ dorsiflexion	-758.3 (± 295.1)	-1391.0 (± 808.9)*
Total Energy generated (J) during MPJ plantarflexion	2.6 (± 1.4)	9.0 (± 6.1)*
Total Energy absorbed (J) during MPJ dorsiflexion	-22.9 (± 8.3)	-43.0 (± 20.2)*
Total energy generated (J) during push-off	0.4 (± 0.4)	0.5 (± 0.5)

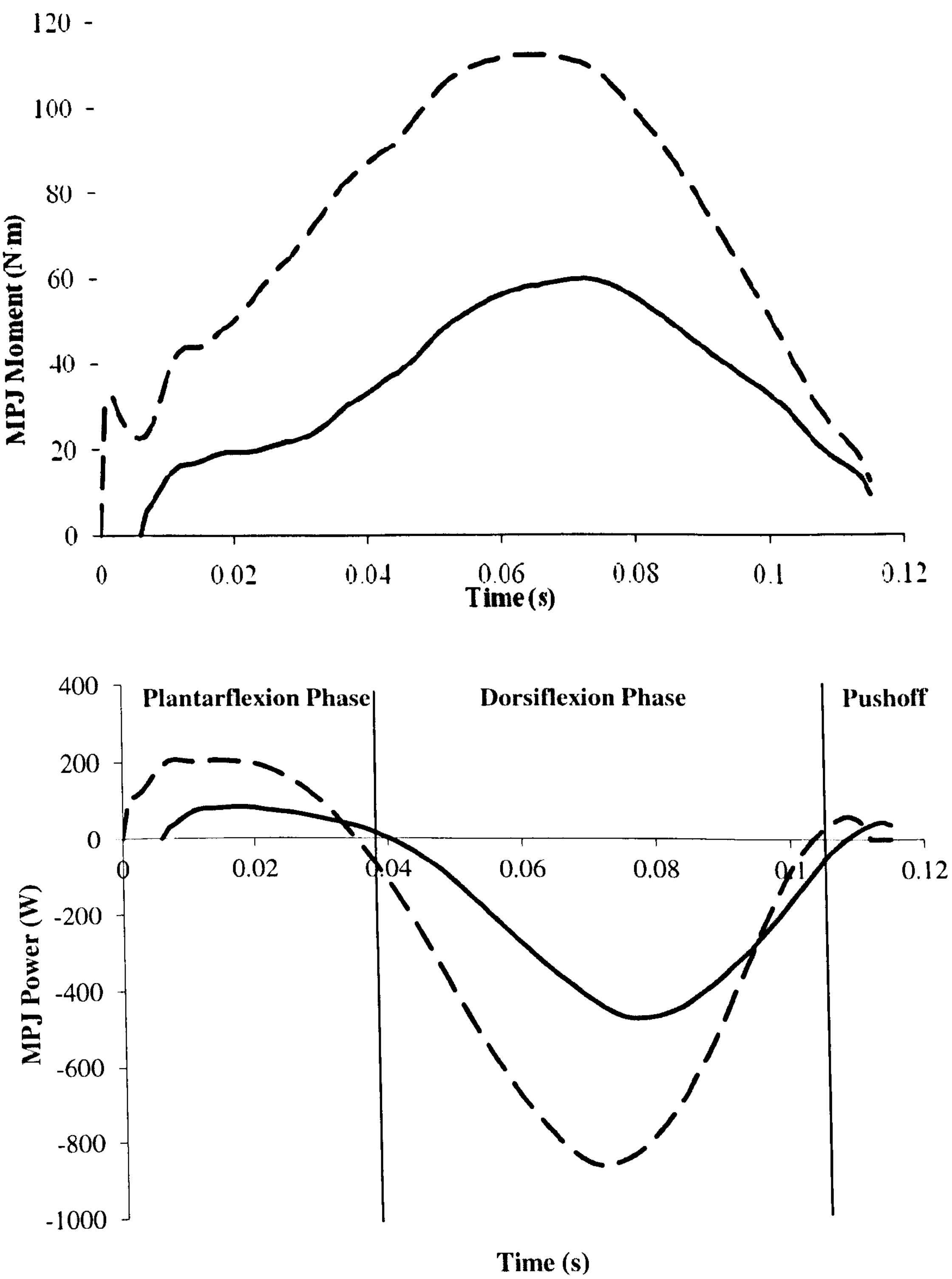


Figure 3.11 MPJ moment (upper graph) and power (lower graph) using two definitions of the MPJ axis for one typical trial. Solid curve represents the oblique axis; the dashed line represents the perpendicular axis. Positional data filtered at $F_c = 8$ Hz.

The MPJ kinetics were also calculated for a dual axis for four subjects in the second phase of testing. Three subjects only used axis MTH1-2 as the centre of pressure was medial of the MTH2 marker throughout stance. Overall, the mean joint moment, energy absorbed and energy generated, both during MPJ plantarflexion and push-off were significantly higher in the oblique axis than the dual axis (Table 3.4). This was due to the larger moment arm for the oblique axis during push off. Overall, for the four additional subjects, peak joint moments were higher by approximately 38% or 18 N·m and total energy absorbed was higher by approximately 33% or 7.5 J for the oblique axis definition compared to the dual axis definition.

Table 3.4. Mean (± S.D.) MPJ moment and energy for second set of subjects (n=4). Comparison of oblique joint axis to dual MPJ axis, data filtered using F_c = 100 Hz. * denotes a significant difference ($p < 0.05$) between the dual and oblique axis definitions.

	Oblique axis joint definition MTH1-5	**Dual axis joint definition MTH 1-2 and MTH 2-5**
Peak MPJ plantar flexor moment (N.m)	**65.3 (± 12.1)**	**47.3 (± 7.4)***
Total Energy generated (J) during MPJ plantarflexion	**1.3 (± 1.9)**	**0.5 (± 0.9)***
Total Energy absorbed (J) during MPJ dorsiflexion	**- 30.2 (± 7.7)**	**- 22.7 (± 5.2)***
Total energy generated (J) during push-off	**1.3 (± 0.2)**	**1.0 (± 0.2)***

3.5 Discussion

The research question addressed in this study was *What influence do methodological issues have on the analysis of MPJ function during sprinting?* Results demonstrate that using a perpendicular joint axis, based only on a single lateral marker, MPJ kinetics were substantially overestimated compared to other axes definitions. As the MPJ axis definition progressed towards one which was more anatomically appropriate (oblique then dual axis approaches) the kinetic variables further decreased due to smaller moment arms about the joint. This suggests that previous researchers have oversimplified the modelling of the MPJ. Furthermore, typical sampling and filtering procedures underestimate MPJ motion and suppress high frequency transients. This study has demonstrated methodological considerations that warrant attention by researchers when investigating the function of the foot during high speed activities.

3.5.1 Effect of MPJ axis definition on joint kinetics

Modelling the joint using a perpendicular axis increased the distance from the MPJ axis to the centre of pressure and overestimated joint kinetics. Therefore Hypothesis 1 is accepted: a more anatomically appropriate joint axis definition would significantly reduce the resultant joint kinetics, in comparison to a perpendicular approach Peak MPJ moment increased by approximately 81% compared to the oblique joint axis which resulted in a shorter moment arm. Resultant moments and kinetics were therefore substantially increased when using a perpendicular axis based on the lateral marker, not an oblique axis as suggested in this study. If a perpendicular analysis based upon a single marker is to be used, it is recommended that a marker on MTH2 would provide more accurate moment arms than a marker on the MTH5.

Bojsen-Moller (1978) points out that the MPJ has two axes: a transverse axis through MTH1–2 and an oblique axis through MTH2-5. Comparing the resultant joint kinetics from the oblique axis to the dual axis, the peak joint moment and total energy absorbed during stance both significantly differed on average by 38%. Overall, the moment arm had a great effect on the resultant MPJ kinetics and this was dependent on the joint axis definition. Although, with current technology, there is no way of calculating completely accurate joint moment arms, the perpendicular approach severely overestimated the MPJ moment. Although the oblique axis also resulted in higher values than the dual axis, the difference between these two axes definitions was smaller.

3.5.2 Effect of sampling rate and cut-off frequency

High cut-off frequencies for processing both position data and ground reaction force data result in better assessment of joint moments during fast transients like the impact phase (Bisseling & Hof, 2006). It has been demonstrated that using a low cut-off frequency, typically used in previous research, not only distorts vital data after landing but also severely underestimates the rate of peak flexion of the joint, evident in the severe underestimation of the MPJ power. Therefore hypothesis 2 is accepted: filtering with low cut-off frequencies significantly underestimates MPJ segmental displacement data. In this study data was oversampled at 1000 Hz, resulting in clear differentiation between high frequency signal and predominant noise from automated marker tracking procedures. This then permitted the high frequency noise to be eliminated by using a high cut-off frequency of 100 Hz.

Just after touchdown there are rapid oscillations of the MPJ angle. These were evident in previous 2000 Hz high speed video pilot testing (Appendix B). These movement transients are high frequency components with signal power of 60-100 Hz, demonstrating the need for a high cut-off frequency during the first 40 ms of stance. After this phase, the frequency of the signal is lower, with highest frequencies up to approximately 30 Hz. The attenuation of high frequency movement characteristics by typical sampling and processing techniques has also been demonstrated for the kinematics of the foot, ankle and shank during the impact phase of the instep kick in soccer (Nunome, Lake, Georgakis & Stergioulas, 2006). However, further work is needed using invasive procedures such as bone pins to support this finding for foot motion during sprinting. Furthermore, the use of bone mounted accelerometers, in combination with marker kinematics and external force measurements would allow a direct measurement of the linear acceleration and the angular velocity, and only one differentiation is required for the angular acceleration (Van den Bogert, 1994).

Power production during push-off was only evident when high frequency movement characteristics were retained which is particularly important as generating power has potential performance implications. Whether the small amount of power produced at the MPJ during push-off can be considered meaningful, or indeed has a direct effect of the energetics of the body as a whole, is yet to be determined. It has been shown that extension of the hallux and toes in walking is an important mechanism in walking (Scott & Winter, 1993) and for sprinting this mechanism may contribute to the ankle plantar flexors producing power in the important take-off phase (Johnson & Buckley, 2001).

Overall, using typical filtering cut-off frequencies greatly undervalued the energy both absorbed and produced at the MPJ throughout sprinting. This needs to be taken into account when comparing the kinetics and energetics of the MPJ motion during high speed activities.

3.5.3 Objectives

Objective 1: To determine the effect of two key methodological issues; MPJ axis definition; and data processing techniques, on the assessment of MPJ function during sprinting.

This study has demonstrated that MPJ kinetics are sensitive to errors in both the modelling of the MPJ line and the processing of the kinematic and ground reaction force data. As previous modelling definitions overestimate joint moments and powers and current processing approaches exclude high frequency components and underestimate peak powers absorbed in stance and produced during push-off, these errors are counteractive in the kinetic calculations. However, the underestimation due to the exclusion of high frequency components did not fully compensate for the overestimation due to axis definition, highlighting the importance of the modelling approach on the resultant kinetics. In order to have confidence in moment arm lengths and joint moments, the researcher should be aware that appropriate joint axis definitions should be used, with at least representing the MPJ axis as an oblique axis. For future studies in this thesis, an oblique MPJ axis definition will be used along with a cut off frequency of 100 Hz.

Objective 2: To develop an appropriate methodology and MPJ axis definition to accurately assess the mechanical energy contribution of the MPJ to sprint running

This study used a simple three segment foot model, similar to Oleson, Adler and Goldsmith (2005), suitable for most sporting activities to determine the mechanical energy contribution of the MPJ. The oblique joint axis was chosen to represent the MPJ orientation for future studies. To improve the accuracy of marker placement, in particular on the MTH's, it is recommended that joint markers are placed directly onto the skin, through holes cut out in the footwear, as suggested by Eslami *et al.,* (2007), and Reinschmidt *et al.*, (1992) to provide a better indication of the actual position of the foot bones. Furthermore, for the second phase of testing, a virtual marker was used for MTH2, which is recommended to avoid excessive motion artefact at this marker location and this will be used throughout the next chapters. It is also recommended to not aggregate medial and lateral aspects of the MPJ, as it is expected that the functions of the medial and lateral forefoot differ and high speed video observations suggest this.

Objective 3: To determine typical MPJ behaviour and function during sprinting for a group of sprinters, including the kinematic, kinetic and energy characteristics of the MPJ and compare to previous researchers.

Despite methodological and data processing differences, sprinting velocities and MPJ kinematic and kinetic values were similar to previous researchers. Toon et al., (2009) reported values between 26 and 50 degrees for maximum MPJ flexion during stance (sprinting wearing standardised sprint spike), the mean range of motion for the four sprinters in this first phase of testing was 36.3 ± 5.1°. Stefanyshyn & Nigg (1997a) reported peak joint moments ranging from 40 – 80 N·m for five male sprinters. The mean peak joint moment for the four subjects in this study using the recommended data processing approach was 58.2 (± 11.1) N·m. Overall the MPJ was indeed an energy absorber, with little energy produced during push-off. Although this warrants further exploration with a larger group of experienced sprinters, this energy loss appears extremely wasteful. If this energy loss could be reduced, by appropriate footwear characteristics, such as sole stiffness, the result may be a positive effect on sprinting performance.

In the first testing phase, the motion of foot segments was estimated by markers over the sprint shoes, which were tightly fastened. This was improved for the second phase of testing by cutting out holes for direct attachment of the markers onto the foot phalanges was assumed to be negligible (as Stefanyshyn & Nigg, 1997) and therefore excluded from the inverse dynamic calculations; the error of this assumption has not been tested.

3.6 Study Conclusion

In answer to the research question posed: *What influence do methodological issues have on the analysis of MPJ function during sprinting?* this study has demonstrated that MPJ kinetics are sensitive to errors in both the modelling of the MPJ line and the processing of the kinematic and ground reaction force data. Hypotheses 1 and 2 were accepted; a more anatomically appropriate joint axis definition significantly reduced the resultant joint kinetics and filtering with low cut-off frequencies significantly underestimated MPJ motion transients.

As previous modelling definitions overestimate joint moments and powers and current processing approaches exclude high frequency components and underestimate peak powers absorbed in stance and produced during push-off, these errors are counteractive in the kinetic calculations. However, the underestimation due to the exclusion of high frequency components did not fully compensate for the overestimation due to axis definition, highlighting the importance of the modelling approach on the resultant kinetics. In order to achieve accurate moment arm lengths and joint moments, the researcher should be aware that appropriate joint axis definitions should be used, with at least representing the MPJ axis as an oblique axis.

CHAPTER 4: FOOT FUNCTION IN SPRINTING: A COMPARISON OF BAREFOOT AND SPRINT SPIKE CONDITIONS.

Overview: This study reported typical foot function during barefoot sprinting and when wearing standardised sprint spikes. The results suggest substantial changes in foot function and performance related parameters due to footwear conditions. Sprint spikes appear to have a controlling affect over the normal behaviour of the foot by limiting the range of motion about the MPJ and reducing peak MPJ dorsiflexion velocity. This does not appear detrimental to sprinting performance as participants achieved significantly higher sprinting velocities wearing sprint spikes than sprinting barefoot. Sprint spikes appear to improve MPJ kinetics, by increasing total energy generated during the push-off phase.

4.1 Introduction

Athletes alter their running styles in adaptation to different surfaces and shoes. The causes of these changes are not well understood (DeWit *et al.*, 2000). However, by quantifying the specific differences in kinematic and kinetic characteristics and pressure distribution patterns, between manipulated shoe conditions, these uncertainties can be addressed. In sprinting, insight into how the foot naturally functions in barefoot running, versus running in sprint spikes, will enhance understanding of the role of the MPJ in relation to the energetics of sprinting. The study of barefoot sprinting will provide a baseline comparison and also demonstrate natural foot function, i.e. typical kinematics, kinetics and loading patterns, in the absence of any effects of footwear. The key differences between normal foot function in barefoot and sprint spike conditions need to be determined to provide insight into the development of sprint spikes for improving performance. Early research by Bosjen-Moller (1978) demonstrated that natural foot function, specifically the free selection of MPJ axes or gears, is compromised by footwear. Furthermore, Toon *et al.* (2009) demonstrated that sprint spikes compromise angular range and angular velocity at the MPJ during maximal sprinting, compared to barefoot sprinting. Increased bending stiffness could therefore limit MPJ motion during the push-off phase and Toon *et al.* (2009) suggested this may potentially minimise an athlete's ability to generate any energy before take-off. This energy production may also be reduced by footwear flexion stiffness properties and could potentially be a factor affecting sprinting performance. Conversely, it is possible that the stiffer sole properties along with design of the toe spring may elicit a spring-like energy return from the sprint spikes. It has also been suggested that increased shoe bending stiffness may increase the MPJ moment, through an increased moment arm, therefore more work may need to be performed to overcome the increased stiffness.

Pressure distribution patterns have been reported for barefoot jogging (De Cock *et al.*, 2005) but there has been no comprehensive study of typical pressure patterns during sprinting. Details of loading patterns and transitions of the centre of pressure that occur during high speed foot contacts will provide evidence for the significance of typical foot behaviour and function to maximal sprinting performance.

4.2 Objectives

The research question to be addressed in this current study is:
How do sprint spikes affect the behaviour of the foot in terms of MPJ function and pressure distribution compared to barefoot sprinting?

The hypotheses developed for this study were as follows:- Hypothesis 1 states sprint spikes would reduce the range of motion at the MPJ as well as the MPJ dorsiflexion velocity. Hypothesis 2 states that sprint spikes would increase the resultant joint moment by increasing the length of the moment arm. Hypothesis 3 states that sprint spikes would reduce the energy absorbed at the joint during MPJ dorsiflexion. Hypothesis 4 states that sprint spikes would increase the amount of energy produced during push-off, due to the spring like properties of the sprint spikes. Finally, hypothesis 5 states that loading would occur more on the forefoot and toes in the spike condition.

Specifically the three objectives of the study were:

1. To characterise normal barefoot behaviour during sprinting.

2. To determine the effect of standardised sprint spikes on MPJ kinematics and kinetics during maximal sprinting.

This objective will address hypotheses 1,2, 3 and 4.

3. To report typical pressure profiles for sprinting, both barefoot and wearing sprint spikes.

This objective will address hypothesis 5.

4.3 Methods

4.3.1 Participants

Eight competitive athletes (club / regional level) volunteered for this study; three female (mean age 22.0 ± 4.8 years, mean height 172.3 ± 9.9 cm, mean mass 64.0 ± 6.9 kg) and five male (mean age 22.7 ± 3.5 years, mean height 186 ± 4.7 cm, mean mass 77.2 ± 3.5 kg). All athletes were trained sprinters who specialised in sprints (including 400 m) / jumps / combined events. The three females had an average 200 m personal best of 25.8 ± 0.8 s. The males had average personal bests of 11.4 ± 0.0 for 100 m (n = 2) and 50.7 ± 0.8 fpr 400 m (n =3). Ethical approval for the study was granted and informed written consent was obtained from all subjects, as per Chapter 3.

4.3.2 Protocol

Each subject underwent two DEXA scans of the foot, for imaging purposes to identify individual's foot anatomy and orientation of the MPJ line. The first scan was performed with the foot in a flat position (transverse view) with a scaling object to obtain foot segment parameter information used in the foot segment model (length of hallux). This scan was used as an aid for placing lead covered reflective markers onto the 1st, 2nd and 5th metatarsal heads (MTH1,2,5) and 1st and 5th metatarsals bases barefoot. Once these five markers were placed onto the foot a second DEXA scan was taken, this time the MPJ was flexed against a triangle support object with an angle of approximately 60 degrees (similar to the maximum flexion angle of the MPJ recorded in barefoot sprinting). This scan was used to optimise the location of the markers relative to the underlying bones and the marker positions were marked on the barefoot then the markers were removed.

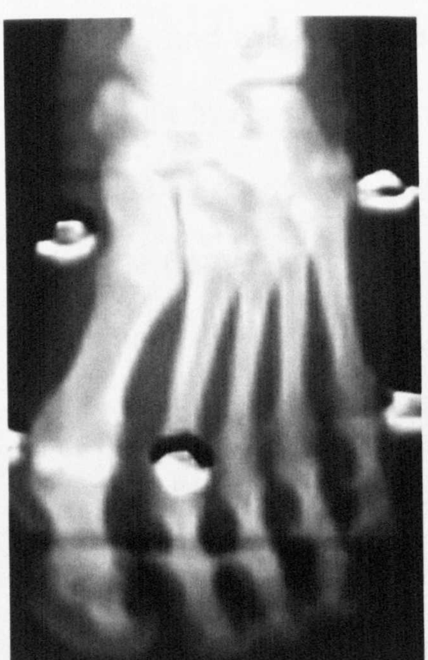


Figure 4.1 Example DEXA scan for one male sprinter, MPJ flexed to approximately 60 degrees. markers are placed on top of MTH2, and to the side of MTH 1 and 5, and metatarsal bases 1 and 5.

Eight maximal sprinting trials were collected on each sprinter, four barefoot and four wearing sprint spikes. Each subject wore the same sprint spikes (different sizes): Nike Zoom Mazcat. The sprints were performed on a 55 m indoor runway with an indoor synthetic track surface and the athletes accelerated for approximately 20 m before data collection (20 - 25 m section surrounding force platform) and were still accelerating during data collection.

Kinematic data were collected using an 8 camera Qualisys system (Pro-reflex MCU 1000 cameras, Qualisys Inc., Sweden) sampling at 1000 Hz. Kinetic data was simultaneously collected using a force platform (Kistler, model 9287B, Kistler, Switzerland), also sampling at 1000 Hz. Pressure data was also captured using a Footscan pressure mat (RS Scan International plate 1 m x 0.4 m x 0.02 m, RS Scan Lab Ltd, Ipswich) sampling at 250 Hz, placed over the force platform and covered with a non-slip matting for barefoot conditions and a 6 mm track covering for spike conditions. Stance phase of the left foot in contact with the force platform was chosen. A starting check mark was used so that the subject hit the force platform without altering technique (targeting). Trials that landed towards the edges of the plate were discounted (approximately two trials were discounted from a total of ten) due to the high centre of pressure inaccuracies (Δ ax and Δ ay < ± 15mm) around the load cells (Kistler, 1993, Appendix A).

A three segment foot model was used, as described in Chapter 3. In the previous study, a marker was placed directly on top of MTH2. As there were problems with this marker moving relative to the underlying MTH, due to the shoe bending especially when the MPJ was fully flexed, a virtual marker was created at the MTH2 for this study. This was done by capturing the location of a pointer wand in the standing calibration trial and also a pointer compression trial. A virtual marker on MTH2 was then created in visual 3D using known offset locations from three surrounding markers.

Markers were placed on the skin for barefoot (plantar surface) using the marked locations from the DEXA scan. For the spike conditions, holes were cut out in the spikes for markers MTH1, 2 (virtual marker) and 5, with the markers placed onto the skin. The remaining markers were placed on top the sprint spike, which was tightly fastened. The five MP joints were considered as a single segment, rotating about an oblique transverse axis defined by the 1st and 5th MT heads. This joint definition was selected based on the joint modelling work from the previous study.

4.3.3 Data processing

A three dimensional analysis was performed. Positional and kinetic data were both smoothed using a fourth order low pass Butterworth filter at a cut-off frequency of 100 Hz. The sample rate and cut-off frequencies were chosen based on work from the previous study. The centre of pressure co-ordinate data from the force platform was processed as per the previous study. Joint moments (including moment arms), powers and energies at the MPJ were also calculated as described in Chapter 3.

4.3.4 Pressure analysis

The maximum localized force, peak pressure and time to peak pressure were calculated for nine areas of the left foot. For each trial, nine anatomical pressure sub-areas were semi-automatically identified within RS Scan Footscan Gait software (version 7.97), using a left standard last- based upon shoe size on the peak pressure footprint. Sub-areas were medial heel, lateral heel, midfoot, metatarsals 1 to 5, hallux and the lesser toes. The footprints were visually inspected and when needed, the areas were manually adjusted using pixel zone definition, whereby individual pixels were re-assigned to other zones or deleted. An example of the zone definition can be seen for a barefoot trial in Figure 4.2.

4.3.5 Statistical analysis

Every barefoot and sprint spike trial was included in the statistical analysis (four per participant). Shapiro Wilks tests for normality of data were conducted. As data was normally distributed, paired samples t-tests were performed to compare mean differences in MPJ kinematic and kinetic variables between barefoot and sprint spike conditions. The level of significance was set at $\alpha = 0.05$.

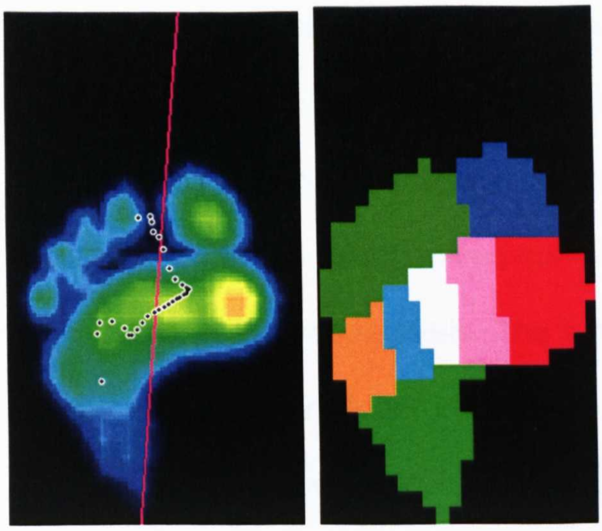


Figure 4.2 Typical barefoot peak pressure footprint and division of zones for analysis: hallux, lesser toes, metatarsals 1-5 and midfoot.

4.4 Results

For the three male subjects mean sprinting velocities for a 5 m section around the 20 m mark were 8.12m/s ± 0.41m/s in sprint spikes and 7.92 m/s ± 0.38 m/s barefoot. For the five female subjects mean velocities were 7.26 m/s ± 0.12 m/s in the sprint spikes condition and 6.81 m/s ± 0.19 m/s barefoot. As a group, mean sprinting velocities were significantly lower ($t(7) = -4.4$, $p < 0.05$) in the barefoot condition (7.50 m/s ± 0.65 m/s) compared to the sprint spikes condition (7.80 m/s ± 0.55 m/s). There was no significant difference in mean stance times between conditions, which were 0.125 s ± 0.010 s for barefoot and 0.127 ± 0.009 for sprint spikes.

4.4.1 Kinematics and Kinetics

Figure 4.3 demonstrates MPJ angular motion (4 trials, one participant) throughout the stance phase in the sprint spikes condition. The motion of the MPJ during stance followed the same pattern as described in Chapter 3, with the MPJ rapidly dorsiflexing during midstance then plantarflexing during push-off, although continuing to plantarflex after the point of take-off. Table 4.1 demonstrates that the MPJ range of motion was significantly reduced (by on average 9.2°) in the sprint spikes condition compared to the barefoot with (t (6) = 3.5, p <0.05) respectively. The typical reduction throughout stance can be seen for one participant in Figure 4.4. Table 4.1 also demonstrates that mean MPJ dorsiflexion velocities were also significantly lower wearing sprint spikes (t (6) = 3.1, p <0.05).

Table 4.1. Mean (± s.d) MPJ kinematics and kinetics barefoot versus shod (n =8), * denotes significant difference between barefoot and sprint spikes condition (p<0.05).

Condition	***Barefoot***	***Sprint Spikes***
MPJ Angular range of motion (°)	51.5 (± 3.5)	42.3 (± 5.7)*
Peak MPJ dorsiflexion velocity (°/s)	1172.2 (± 309.8)	873.1 (± 154.9)*
Peak MPJ plantar flexor moment (N.m)	55.6 (± 11.3)	63.9 (± 14.9)*
Peak Positive Power (W) generated after touchdown	300.0 (± 202.5)	140.9 (± 106.3)*
Peak Negative Power (W) generated during MPJ flexion	-712.7 (±207.2)	-780.1 (±228.7)
Total Energy generated (J) after touchdown	2.8 (± 2.1)	1.3 (± 1.0)*
Total Energy absorbed (J) during MPJ flexion	-31.3 (± 7.7)	-29.9 (± 7.7)
Total energy generated (J) during push-off	0.5 (± 0.5)	1.4 (± 1.0)*

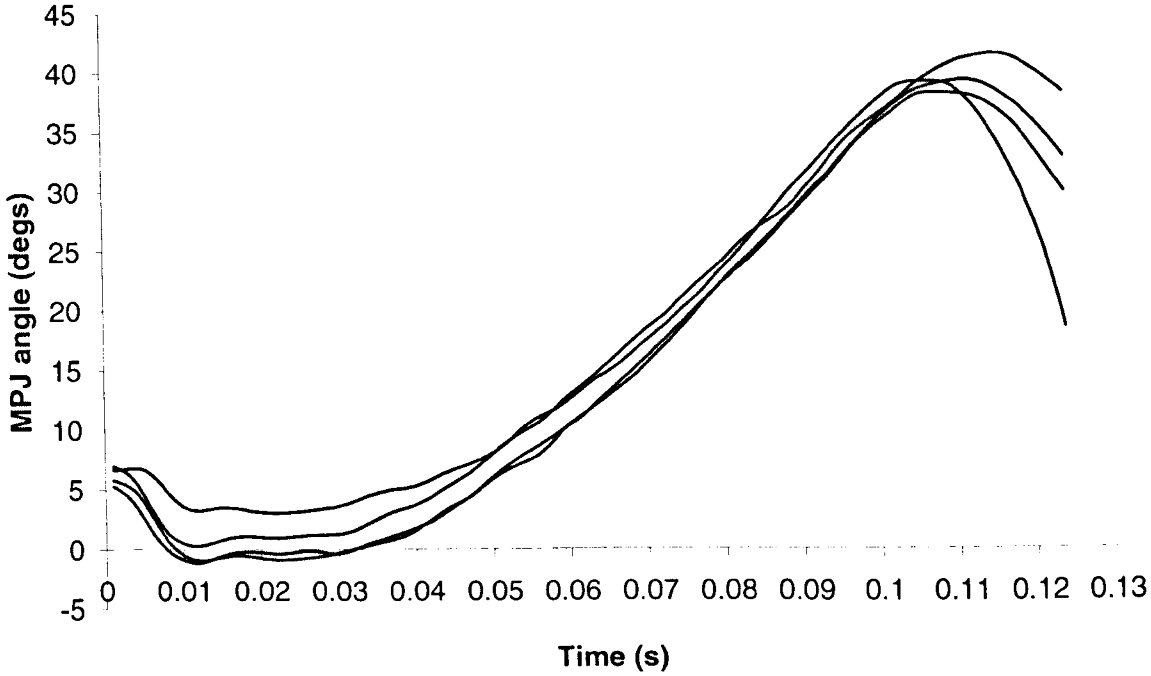


Figure 4.3. MPJ angle, throughout the stance phase of sprinting for one typical female participant, four sprint spikes trials.

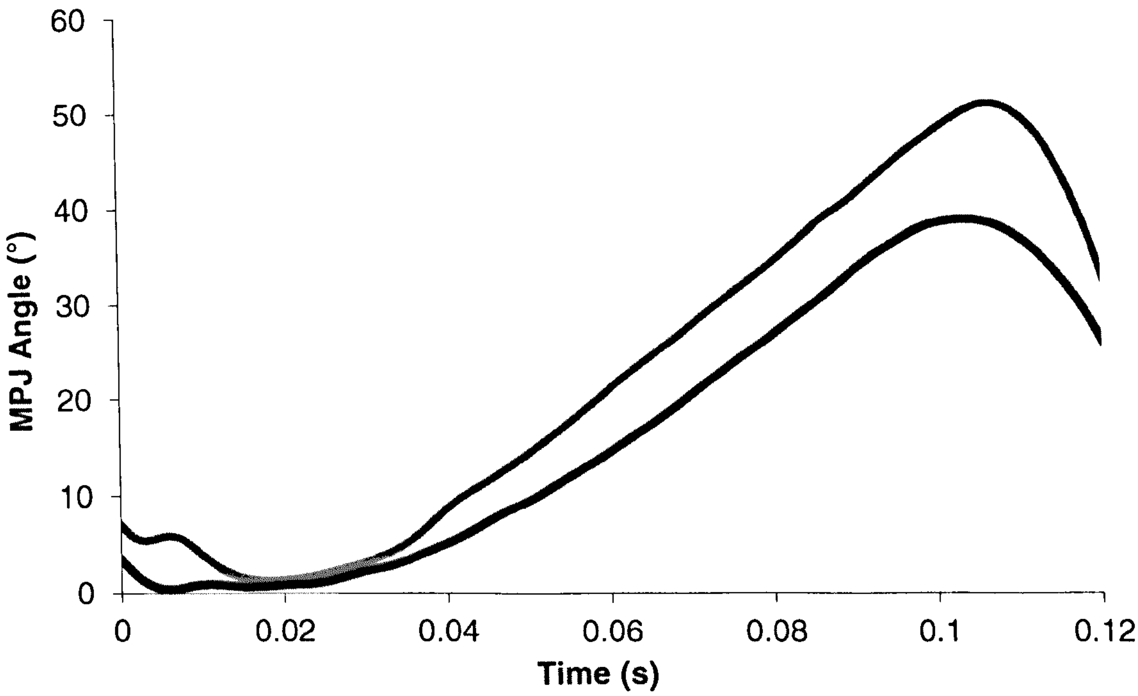


Figure 4.4 Mean (n=4) MPJ angle throughout the stance phase of sprinting for one female participant, sprinting barefoot (grey line) and wearing sprint spikes (black line),

Despite faster sprinting velocities for the sprint spike trials, there were no differences in peak vertical forces with mean Fz values of 2184.9 N ± 263.2 N and 2169.8 N ± 216.0 N for the barefoot and sprint spike condition respectively. Figure 4.5 demonstrated typical Fz and Fy values for one male participant. Mean Fy horizontal propulsive forces were slightly greater for the sprint spike conditions than the barefoot conditions with peak values of 622.0 N ± 158.0 N and 570.8 ± 154.1 N respectively, although the difference was not significant.

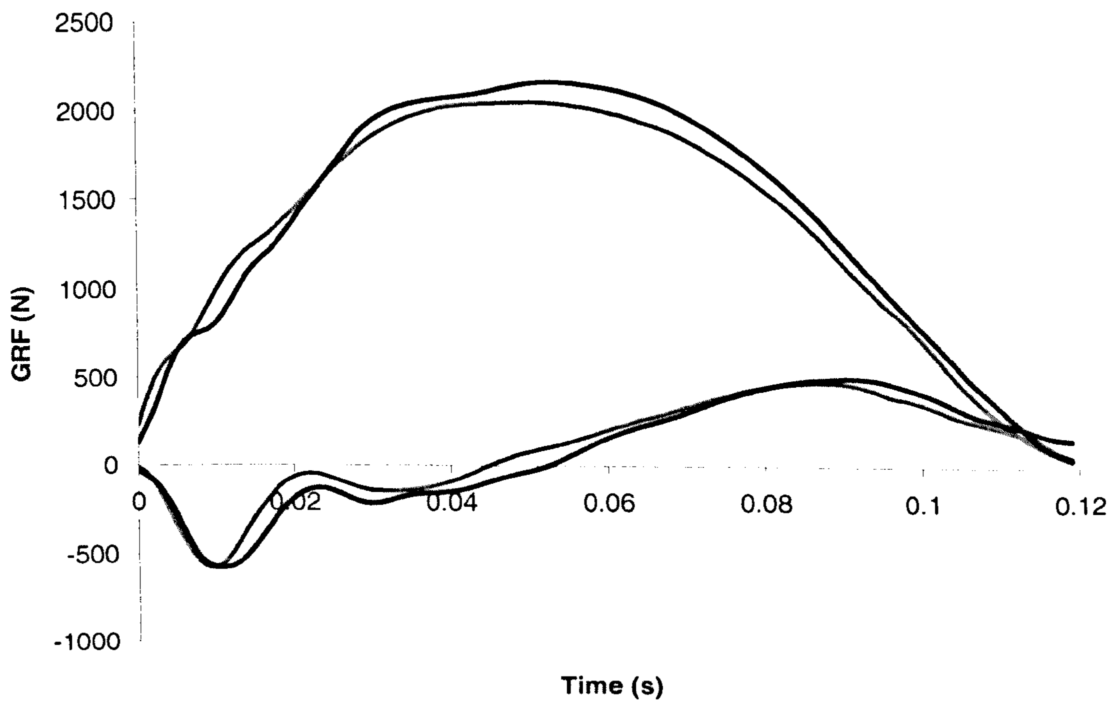


Figure 4.5 Typical Fz and Fy forces for the barefoot (grey line) and sprint spike condition (black line) condition for one male participant.

Figure 4.6 demonstrates the resultant MPJ moment for four typical trials for one participant, which was plantarflexor throughout stance. Resultant peak moments ranged from 51 to 85 N·m for the eight participants wearing sprint spikes. The MPJ moments were significantly higher in the sprint spikes condition (63.9 N·m ± 14.9 N·m) compared to the barefoot condition (55.6 ±

11.3 N·m) (t (7) = -2.7 p <0.05). Figure 4.7 demonstrates the difference in mean resultant joint moment between barefoot and shod conditions for one participant. Peak vertical moment arms were greater in the sprint spikes condition (t (7) = -12.1 $p < 0.05$) with lever distances of 0.041 m ± 0.00 m, compared to 0.027 m ± 0.04 m in the barefoot condition when MPJ peak moments were achieved.

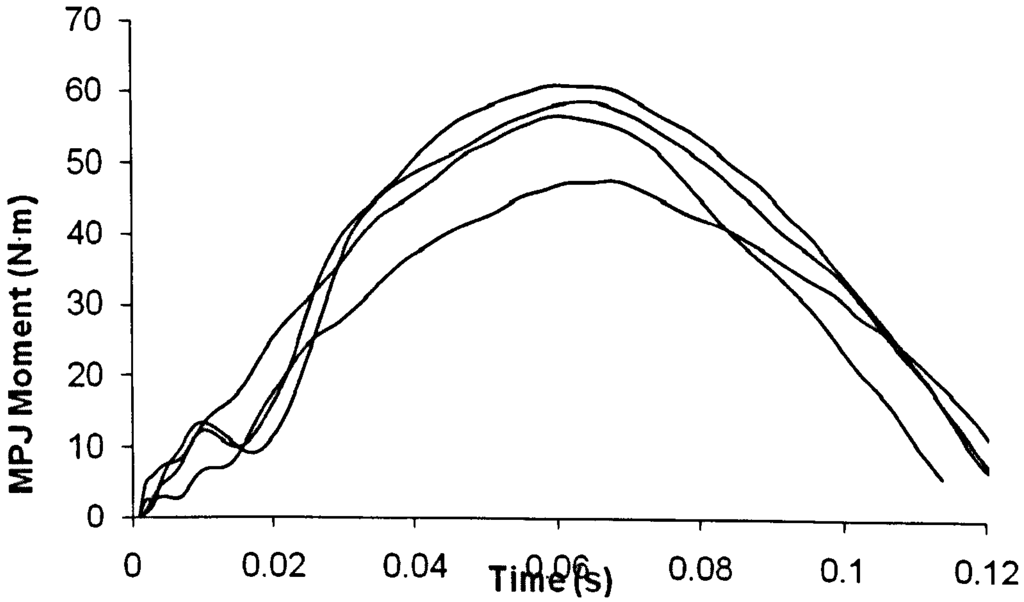


Figure 4.6 MPJ joint moment throughout the stance phase of sprinting for one typical female subject, four sprint spike trials.

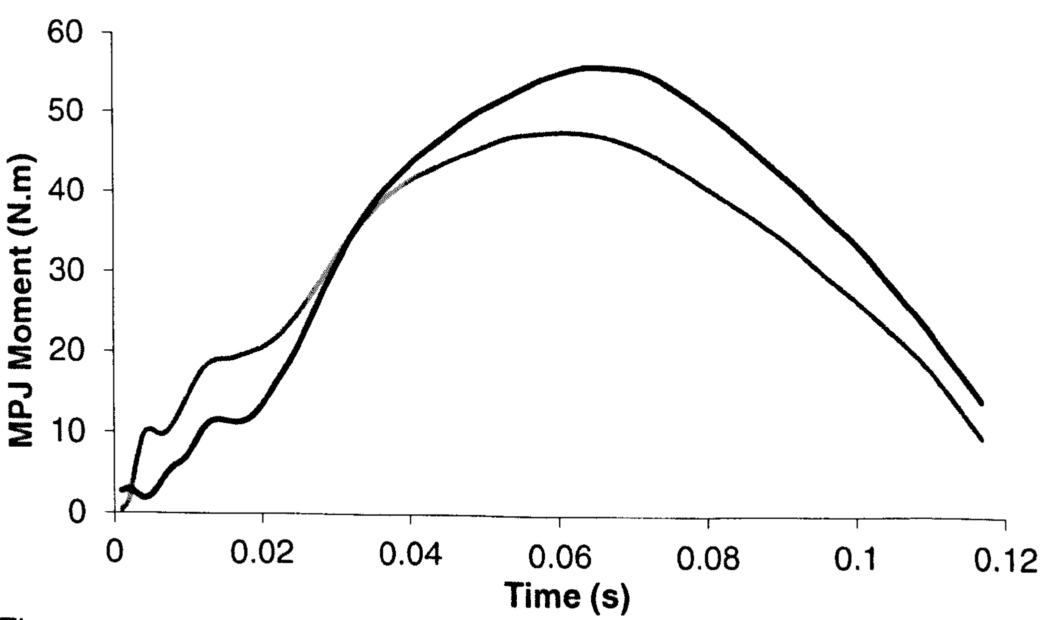


Figure 4.7. Mean MPJ Moment for one participant wearing sprint spikes (black line) and barefoot (grey line), average of four trials per condition.

There were no significant differences in the negative power during stance, however the barefoot condition produced more positive power throughout stance (t (7) = 2.6 $p < 0.05$).

There were no significant differences in the total energy absorbed at the MPJ during stance (-31.3 J ± 7.7 J barefoot, -29.9 J ± 7.7 J sprint spikes), therefore the footwear condition did not affect the total energy loss (Table 4.1). The barefoot condition produced 1.5 J more energy shortly after touchdown (t (7) = 2.8 $p < 0.05$). The sprint spikes condition produced more energy during push-off, even though the amount of energy produced was small 1.4 J ± 1.0 J (t (7) = -3.3 $p < 0.05$). During push–off the peak horizontal moment arms were greater for the sprint spikes condition (0.037 m ± 0.001 m) than the barefoot condition (0.029 m ± 0.001 m) although this difference was not statistically significant.

Typical intra-subject variation in the kinematic and kinetic variables for one participant is shown in Table 4.2, with coefficients of variation ranging from 5.3% to 25.5%. Despite this variation, the magnitude of the significant differences between barefoot and sprint spike conditions in the kinematics and kinetics were high. Calculated effect sizes (0.48 – 0.79) for the kinematic and kinetic variables were moderate to large (Cohen's *d*) suggesting a meaningful localised effect on the function of the MPJ.

4.4.2 Pressure results

Loading predominantly occurred on the forefoot, although three participants demonstrated heel or midfoot contact during early stance. Figure 4.8 demonstrates typical maximum pressure profile plot for one female participant, barefoot and wearing sprint spikes.

Table 4.2. Intra-subject variability: Mean (± s.d) and Coefficient of Variation (CoV) for MPJ kinematic and kinetic variables. One typical participant, barefoot and sprint conditions, four sprint trials per condition.

Condition	***Barefoot Mean ± s.d***	***Barefoot CoV (%)***	***Spikes Mean ± s.d***	***Spikes CoV (%)***
MPJ Angular range of motion (°)	50.1 (± 2.7)	5.3	39.1 (± 2.2)	5.7
Peak MPJ dorsiflexion velocity (°/s)	1417.1 (± 160.7)	11.3	919.7 (± 132.0)	14.3
Peak MPJ plantar flexor moment (N.m)	47.6 (± 4.8)	10.3	56.1 (± 5.8)	10.4
Peak Positive Power (W) generated after touchdown	279.2 (± 60.1)	21.5	105.2 (± 40.5)	38.4
Peak Negative Power (W) generated during MPJ flexion	-604.2 (±151.2)	25.0	-581.3 (± 92.4)	15.9
Total Energy generated (J) after touchdown	2.2 (± 0.5)	22.7	1.9 (± 0.4)	18.8
Total Energy absorbed (J) during MPJ flexion	-29.1 (± 3.4)	11.6	-25.8 (± 3.2)	12.5
Total energy generated (J) during push-off	0.8 (± 0.2)	23.7	1.3 (± 0.3)	25.5

Figure 4.9 demonstrates barefoot versus sprint spike pressure profile for a different female participant at key time intervals throughout stance. In both conditions, touchdown occurred on the lateral portion of the forefoot, after which the centre of pressure progressed medially across the metatarsal heads then anteriorly towards the hallux and second toe for push-off. In the shod condition there were higher localised pressure peaks due to the locations of

the spikes on the sole of the sprint spike shoe. In the barefoot conditions, there was a larger contact area, with more evenly distributed loads over the metatarsal heads and a clear lateral to medial transition of the centre of pressure (Figure 4.11). In the sprint spikes condition the centre of pressure followed more of a straight line anteriorly (Figure 4.10) with loading more confined to the medial side of the forefoot and greater anterior progression to the anterior edge of the spike plate for push-off.

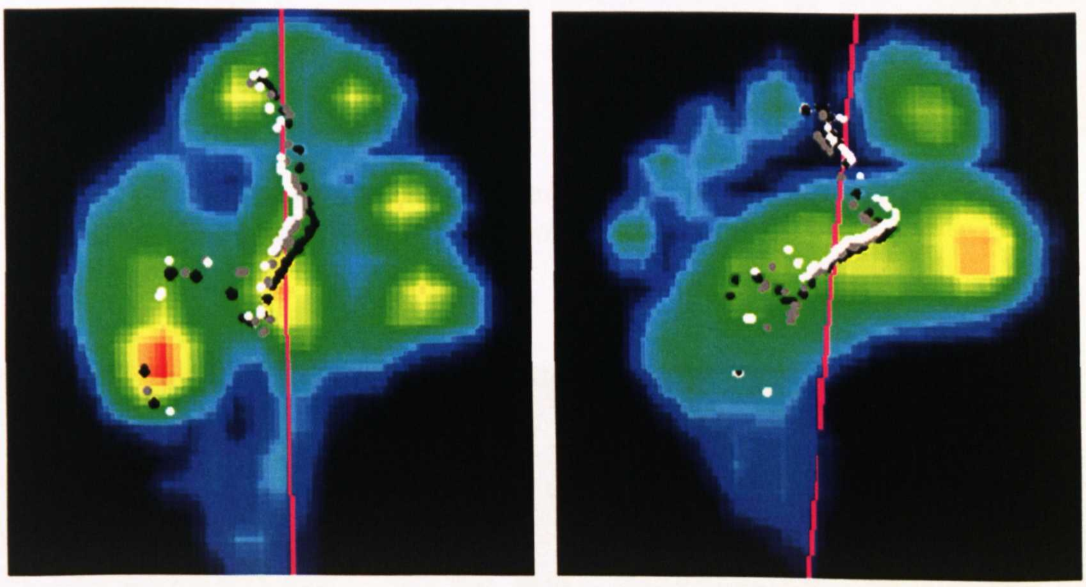


Figure 4.8 Typical pressure distribution for one female participant, wearing sprint spikes (left) and barefoot (right). The path of the CoP is also overlaid for three trials.

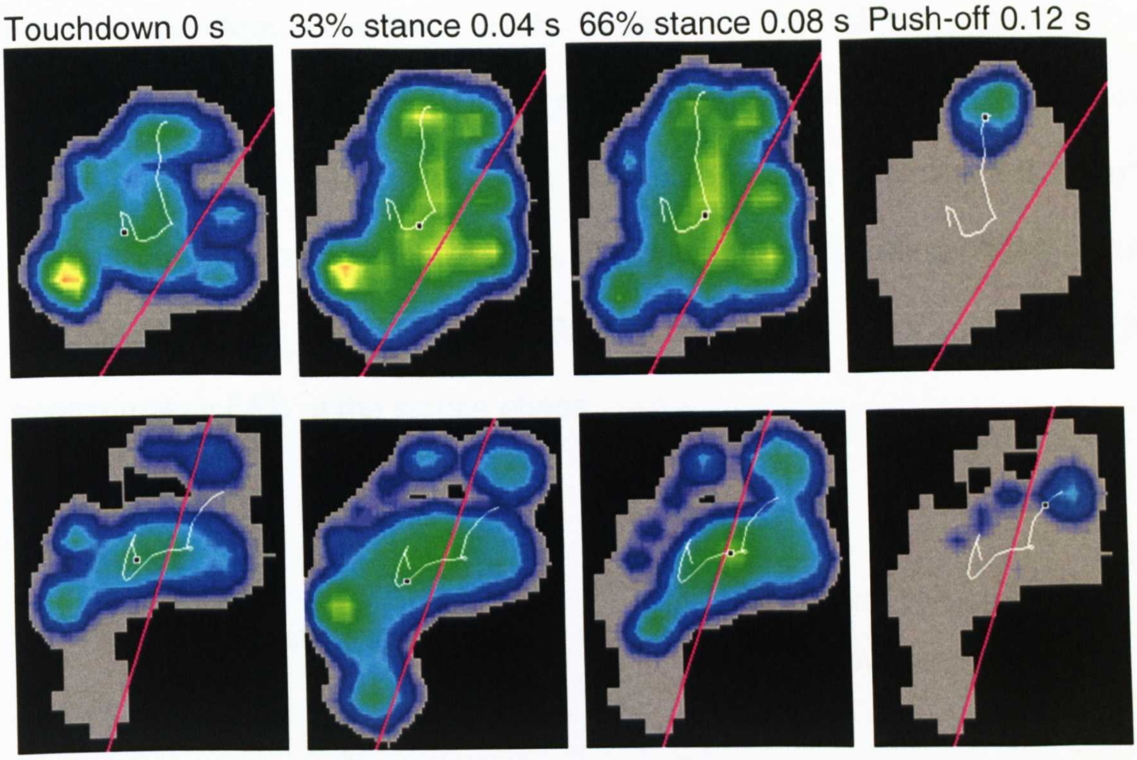


Figure 4.9 Pressure profile during stance for one female participant, wearing sprint spikes (top) and barefoot (bottom).

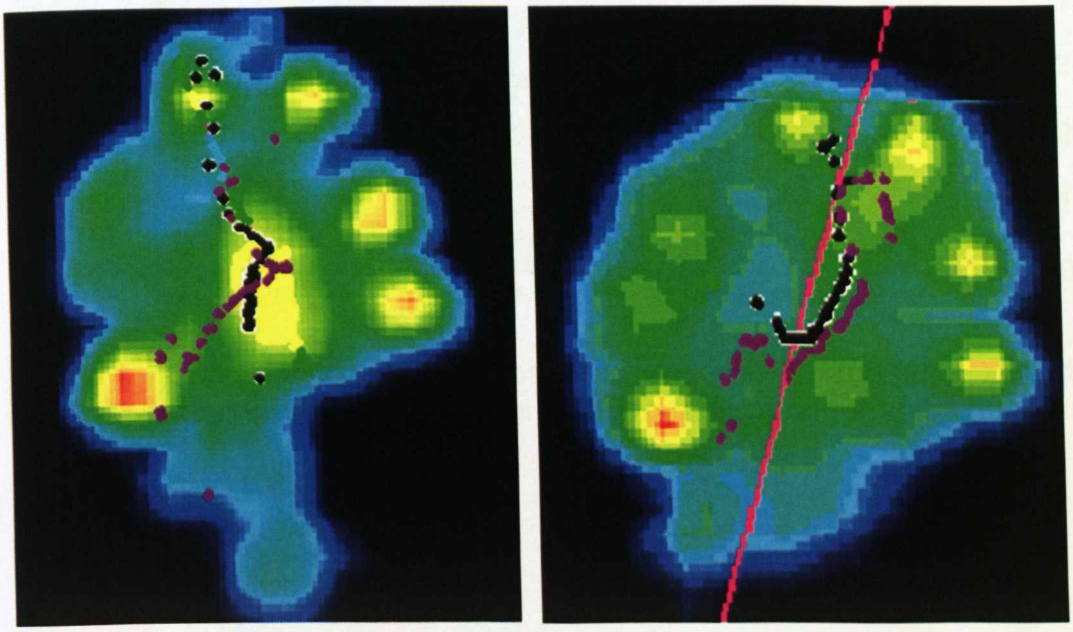


Figure 4.10 Pressure profile for an additional two participants, both male. The path of the CoP is overlaid (black dots) on the sprint spikes footprint, for one typical trial in sprint spikes. The CoP from one barefoot trial has also been overlaid (purple dots).

For the barefoot condition, peak forces on the MTH5 occurred at approximately 25% of stance. The highest loading was at the MTH1 with peak forces of 584.4 N ± 151.5 N, (peak pressures of 29.1 $N \cdot cm^2$ ± 8.8 $N \cdot cm^2$) which occurred at approximately 60% of stance phase. Loading on the hallux was great (465.8 N ± 137.8 N, 22.9 $N \cdot cm^2$ ± 6.1 $N \cdot cm^2$) occurring at approximately 64% of the stance phase.

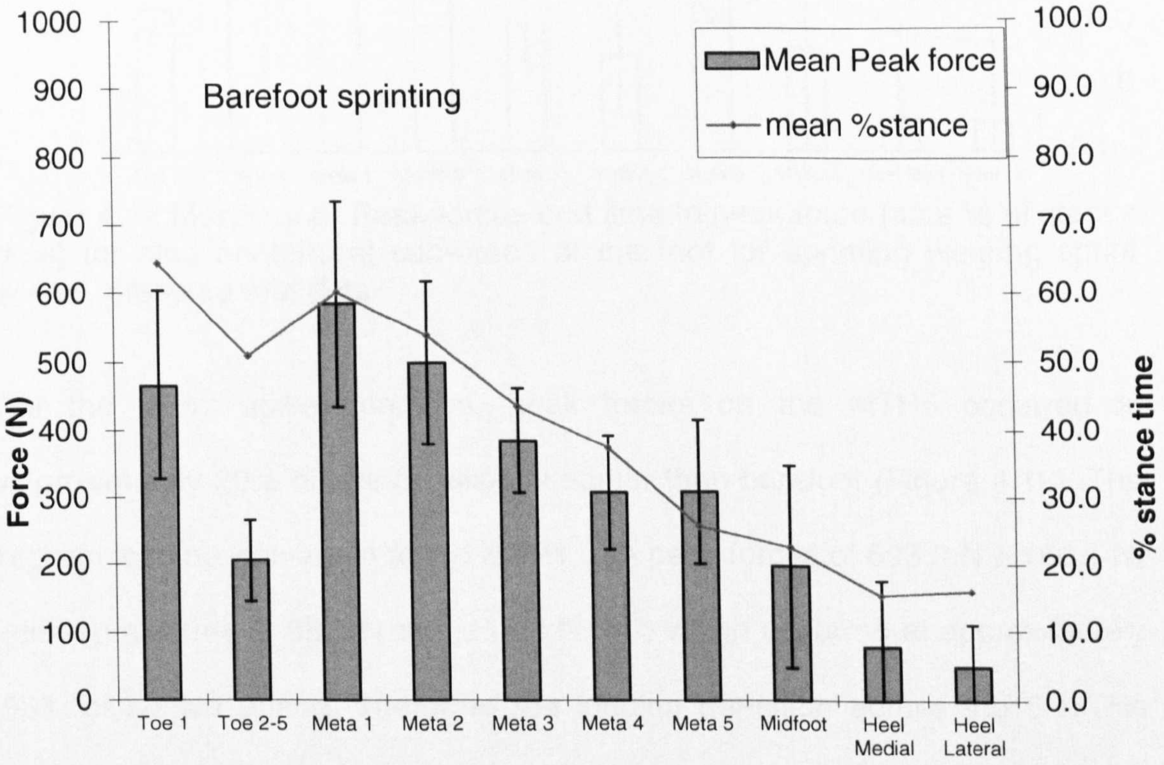


Figure 4.11 Mean (n=8) Peak forces and time to peak force (as a % of stance time) for nine anatomical sub-areas of the foot for barefoot sprinting, pressure mat data.

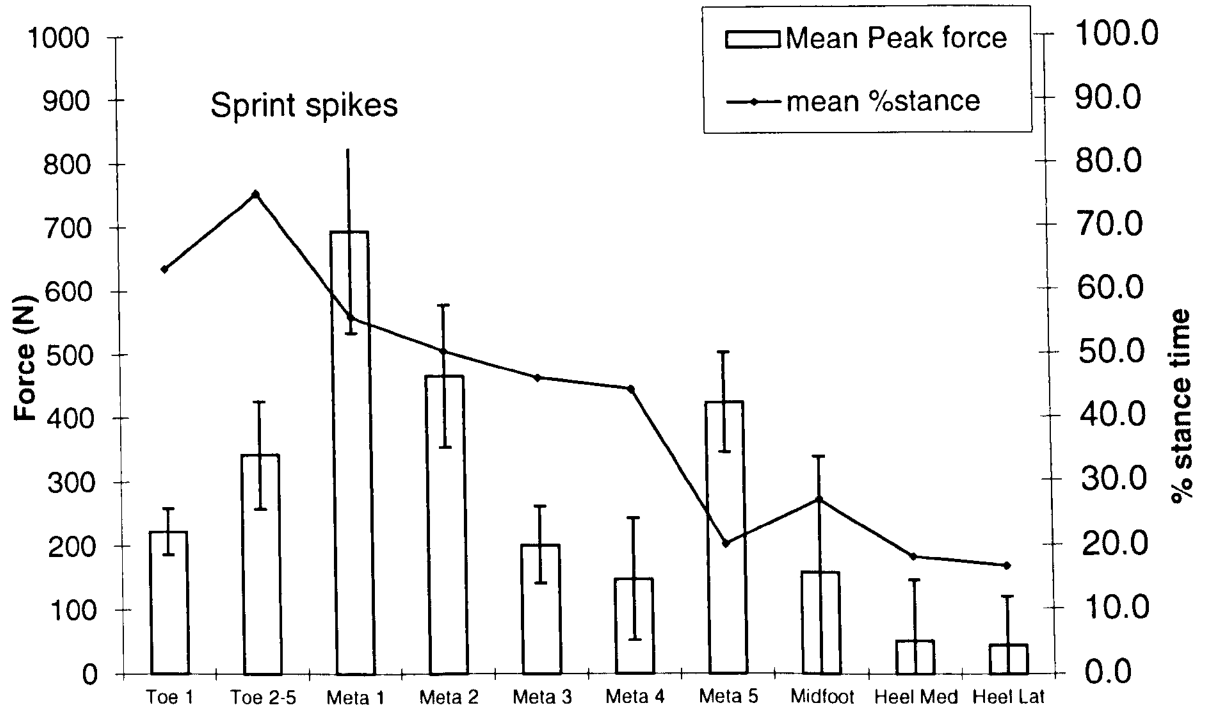


Figure 4.12 Mean (n=8) Peak forces and time to peak force (as a % of stance time) for nine anatomical sub-areas of the foot for sprinting wearing sprint spikes, pressure mat data.

For the sprint spike condition, peak forces on the MTH5 occurred at approximately 20% of stance, slightly earlier than barefoot (Figure 4.12). The highest loading was again at the MTH1 with peak forces of 693.3 N ± 160.1 N, (peak pressures of 38.9 N·cm^2 ± 14.3 N·cm^2) which occurred at approximately 55% of stance phase. Therefore the loading transition across the 5 MTHs occurred approximately 10% quicker in the sprint spike conditions than in the barefoot condition. As the centre of pressure progressed anteriorly, loading was greater on toes 2-5 than the hallux, demonstrating a more central push off, with peak forces / pressures of 465.8 N ± 137.8 N / 22.9 N·cm^2 ± 6.1 N·cm^2 occurring at approximately 75% of the stance phase (slightly later than peak hallux forces in the barefoot condition).

Comparing peak forces and pressures between the barefoot and sprint spike conditions (Table 4.3), the peak forces and relative peak forces under the hallux, and MTH3 and MTH4 were significantly greater ($p<0.05$) in the barefoot trials. Peak forces at the hallux occurred at 64% of stance time. For the sprint spike condition, initial loading was greater on MTH5 (significant increase in peak forces, relative peak forces and pressures). There was also greater loading under MTH1, although not significant. Greater peak forces occurred at the lesser toes, than the hallux during the push-off, with peak forces, relative peak forces and peak pressures significantly greater ($p<0.05$) in the sprint spike conditions than in the barefoot condition. These peak forces on the lesser toes also occurred later – at 75% stance in the sprint spike condition, compared to 51% of stance for the barefoot condition

Table 4.3 Pressure mat data for barefoot and sprint spikes conditions: Mean (±s.d.) peak forces, peak pressures, relative peak forces and time to peak forces (% stance time) for nine anatomical sub-areas of the foot. Relative loads were defined as the % of maximum load under whole foot that was experienced on that area. * denotes a significant difference between the two conditions ($p < 0.05$)

	BARE	SHOD	BARE	SHOD	BARE	SHOD	BARE	SHOD
	Peak force (N)		Peak pressure (N/mm)		% stance to peak force		Relative peak force (%)	
Hallux	465.7 (±137.8)	223.0* (±36.4)	22.9 (±6.1)	19.0 (±4.9)	64.4 (±8.5)	63.4 (±9.3)	14.9 (±5.6)	8.3* (±1.5)
Toes 2-5	206.4 (± 60.4)	342.1* (±84.1)	6.1 (±1.8)	12.2* (±2.5)	50.9 (±23.0)	75.2* (±14.4)	6.3 (±2.4)	12.5* (±3.3)
MTH1	584.4 (±151.5)	693.3 (±160.1)	29.1 (±8.8)	38.9 (±14.3)	60.0 (±8.8)	55.9 (±8.3)	19.2 (±5.5)	25.4 (±6.5)
MTH2	496.7 (±119.2)	465.7 (±111.7)	41.6 (±7.0)	39.5 (±8.5)	53.7 (±12.7)	50.5 (±12.4)	16.2 (±3.0)	17.2 (±4.0)
MTH3	382.5 (±406.6)	201.9* (±60.3)	42.2 (±6.5)	29.4* (±6.5)	43.3 (±12.7)	46.3 (±14.7)	12.7 (±2.1)	7.5* (±1.7)
MTH4	306.0 (±106.6)	147.8* (±95.9)	36.3 (±9.9)	20.5* (±8.9)	37.3 (±13.1)	44.4 (±16.7)	9.8 (±1.9)	5.5* (±2.2)
MTH5	307.3 (±106.6)	423.5* (± 78.7)	23.4 (±9.0)	33.3* (±7.3)	25.6 (±8.8)	20.2 (±6.6)	9.7 (±2.1)	16.2* (±2.2)
Midfoot	197.2 (±149.3)	157.3 (±180.5)	5.3 (±3.0)	4.1 (±3.4)	22.5 (±6.9)	27.1 (±6.2)	5.6 (±4.9)	4.8 (±4.8)
Heel medial	78.4 (±96.4)	50.1 (± 95.9)	6.0 (±8.8)	2.6 (±5.2)	15.3 (±4.9)	18.3 (±4.1)	1.7 (±1.9)	1.4 (±2.5)
Heel lateral	50.2 (±60.3)	43.2 (±77.2)	4.5 (±6.3)	2.4 (±4.1)	15.9 (±4.3)	16.8 (±2.0)	1.1 (±1.3)	1.3 (2.1)

4.5 Discussion

The current study was designed to address the research question: *How do sprint spikes affect the behaviour of the foot in terms of MPJ function and pressure distribution compared to barefoot sprinting?*

The results of this study suggest substantial changes in foot function and performance related parameters between barefoot sprinting and sprinting wearing standardised sprint spikes. Sprint spikes appear to have a controlling effect over the normal behaviour of the foot, by limiting the range of motion about the MPJ and reducing peak MPJ dorsiflexion velocity, however this does not appear detrimental to sprinting performance as participants achieved significantly higher sprinting velocities wearing sprint spikes than sprinting barefoot. Sprint spikes appear to improve MPJ kinetics, by increasing total energy generated during the push-off phase.

4.5.1 Objectives

Objective 1. To characterise normal barefoot behaviour during sprinting.

The athletes in this study performed better wearing sprint spikes, with significantly increased mean sprinting velocities of 7.80 m/s ± 0.55 m/s compared to the barefoot condition 7.50 m/s ± 0.65 m/s, demonstrating that wearing sprint spikes improves sprinting performance. The sprint speeds recorded for the sprint spike conditions were similar to other researchers who collected data at a similar point. Stefanyshyn and Nigg (1997) reported male sprinters velocities ranging from 7.1 m/s to 8.6 m/s at 15 m, although Johnson and Buckley (2001) reported mean sprinting velocities of 8.66 m/s ± 0.37 m/s for six male sprinters at the 15 m mark.

The MPJ underwent a large range of motion during stance, yet the athletes flexed their MPJ minimally at take-off, agreeing with Stefanyshyn et al. (2002), who reported average peak extensions at the MPJ from medial and lateral aspects combined were 36.5 ° and 37.7 ° for male and female Olympic sprinters respectively. Toon *et al.* (2009) reported peak MPJ (medial aspect) dorsiflexion values of 43 ° ± 3 ° for barefoot sprinting and 31 ° ± 3 ° wearing standardised sprint spikes for four sprinters at the 50 m point. The mean MPJ range of motion values in this study (51.5 ° ± 3.5 ° barefoot and 42.3 ° ± 5.7 °) were slightly higher than those reported in the previous research, this may be due to a relatively low stiffness standard sprint spike used, or likely, due to different methodologies employed to measure MPJ angular movement, as both Stefanyshyn *et al.* (2002) and Toon *et al.* (2009) obtained their results from manually digitising the medial aspect of the MPJ from high speed video. Peak MPJ dorsiflexion velocities for this study of 1172.2 °/s ± 309.8 °/s barefoot and 873.1 °/s ± 154.8 °/s are similar to Krell and Stefanyshyn (2006) who reported peak velocities between 900 and 1300 °/s for 100 m Olympic athletes, but are higher than Toon *et al.* (2009) who reported values of 531 °/s to 737 °/s for barefoot and sprint spikes respectively.

The peak Fz forces (2184.9 ± 263.2 N barefoot and 2169.8 ± 216.0 N sprint spikes) and Fy forces (570.8 N ± 154.1 N barefoot and 622.1 N ± 158.1 sprint spikes) were similar to those reported for maximal sprinting (8.8 m/s) by Belli *et al.*, (2002) of 2173 N ± 233 N in the Fz direction and up to 809 N ±136 N in the Fy direction. Bezodis *et al.* (2008) reported higher peak Fz values of approximately 3200N, as sprinters achieved higher velocities exceeding 10 m/s. It has been previously shown that the magnitude of vertical forces is positively related to sprinting velocity (Weyand *et al.,* 2000).

The importance of joint kinetic information to fully understand the role of the MPJ in sprinting performance is paramount. The joint kinetics at the MPJ during sprinting have yet to be fully explained, yet will provide insight into the effect of footwear on the energetics of sprinting and may influence sprint performance. The peak MPJ plantarflexor moment for the sprint spike condition was 63.9 N·m ± 14.9N·m, very similar to the mean value reported in the previous chapter (65.3 N·m ± 12.1 N·m) although due to different methodologies discussed in chapter 3 (such as capture rate, MPJ axis definition) and data processing techniques employed to measure the MPJ moment, these values are slightly lower than 70 to 120 N·m reported by Stefanyshyn & Nigg (1997). As the MPJ moment was plantarflexor throughout, it is assumed that the toe flexor muscles were contracting throughout stance. As reported in Chapter 3, the MPJ was a large energy absorber and produced little energy at push-off. For the barefoot condition, 31.3 J ± 7.7 J was lost during the energy absorption phase and only 0.5 J ± 0.5 J was generated during push off. The respective values for the sprint spike condition were 29.9 J ± 7.7 J and 1.4 J± 1.0 J.

Objective 2. To determine the effect of standardised sprint spikes on MPJ kinematics and kinetics during maximal sprinting.

As sprint spikes resulted in a significant reduction in the range of motion at the MPJ as well as the MPJ dorsiflexion velocity, compared to the barefoot trials, hypothesis 1 is accepted. This provides evidence for the inherent controlling effect of the sprint spikes, which act as a velocity dampener during MPJ dorsiflexion. The results of this study indicated that there was a phase of MPJ plantar flexion during take-off and consequently an opportunity to generate energy at the MPJ during take-off, disagreeing with Stefanyshyn and Nigg (1997) who stated that the toes remain dorsiflexed, thus generating no or very little energy at take-off. As sprint spikes compromised the angular velocity during this phase, this could potentially affect any potential energy generation, however, compared to the barefoot condition, energy generation during push-off was not compromised.

Sprint spikes also resulted in increased resultant joint moments by increasing the length of the moment arm, accepting research hypothesis 2. It is expected that this is due to the increased longitudinal bending stiffness of the sprint spikes, along with the effect of the toe spring design. In order to cope with an increased lever arm and rigid link, the plantarflexors (in particular the triceps surae) need to produce more work, if this additional force can be translated this may result in a more effective transfer of energy and lead to an improvement in sprinting performance.

Hypothesis 3 stated that sprint spikes would reduce the energy absorbed at the joint during MPJ dorsiflexion. Although the sprint spikes did result in slightly reduced energy loss at the MPJ, compared to the barefoot condition, this was not significant, therefore hypothesis 3 is rejected. The increased lever length in the sprint spike condition did not amplify the energy absorption at the MPJ, in fact the increased plantarflexion moment of the MPJ during the barefoot condition lead to increased (although not significant) energy absorption. Hypothesis 4 stated sprint spikes would increase the amount of energy produced during push-off, due to the spring like properties of the sprint spikes. The sprint spikes did result in increased energy production during push-off, due to an increased moment arm, accepting hypothesis 4. Therefore, the stiffer sprint spike condition, compared to the barefoot condition, seemed to increase the effective lever length of the foot about the MPJ, which may facilitate a more effective propulsive system.

Objective 3. To report typical pressure profiles for sprinting, both barefoot and wearing sprint spikes.

Hypothesis 5 stated that loading would occur more on the forefoot and toes in the spike condition, this was accepted. Results from the peak pressure profiles demonstrated a greater loading area in barefoot condition and a more even distribution of loading under the metatarsal heads, than in the sprint spikes. In sprint spikes there was greater localised loading under MTH5 at foot contact, than under the medial forefoot: in particular MTH1 and the lesser toes. As loading was mainly confined to the metatarsals 1 and 2 and the hallux in the sprint spike conditions, this is concurrent with the notion of a two speed construction of the human foot whereby during sprinting (high gear) the push-off is performed about the transverse axis (Bojsen-Moller, 1977).

Overall, the medial forefoot and toes accounted for 63% and 56% of the total loads in shod and barefoot sprinting respectively. These are slightly higher than the values reported for sprinting in the literature. Eils *et al.* (2004) reported the predominant loading areas were found in the medial forefoot, hallux and second toe, accounting for 55% of the total load for sprinting in football boots.

The pressure results suggest that the function of the medial side of the foot is different to the lateral portion and this differs between barefoot and shod conditions with sprint spikes further confining loading to the medial side of the foot. In sprint spikes, the natural lateral to medial roll of the forefoot was somewhat reduced and the loading was quickly more centralized. Furthermore, there was greater rapid progression of the centre of pressure in sprint spikes, with push-off occurring over the hallux and second toe and the CoP progressing straight forward to the edge of the toe box. This is perhaps due to the toe spring promoting forefoot contact, and also coincides with the small amount of energy generated during the last 10 ms of stance, thereby adding evidence to suggest sprint spikes help to create a more effective rigid lever at take-off.

Direct comparisons between the barefoot and sprint spike conditions should be viewed with caution, firstly due to slight differences in the cover mat placed over the pressure mat used to prevent slipping (barefoot trials) and damage to the pressure mat (sprint spike trials) and secondly, the sole of the sprint spike and position of the spikes would have also influenced the peak pressures underneath the foot. Regardless of this limitation, overall transitions in forefoot loading are still evident and full penetration of the sprint spikes into the covering surface ensures that high pressures from the spike locations minimises the distortion in the transitions of the CoP. Pressure insoles were used during pilot testing but were not robust enough to deal with the amount of bending during sprinting, therefore they quickly deformed within one testing session. However, an example of the pressure profile obtained from insole data for submaximal sprinting (see Appendix C) does support the notion of a reduced lateral to medial transition when wearing sprint spikes. This suggests that the differences in forefoot loading profiles between barefoot and sprint spike conditions are not solely due to artefact generated because of the high pressure points under the spikes.

4.6 Study Conclusion

In answer to the research question posed: *How do sprint spikes affect the behaviour of the foot in terms of MPJ function and pressure distribution compared to barefoot sprinting?* This study has demonstrated performance related differences in MPJ kinematics and kinetics between barefoot sprinting and when sprinting in spikes. Hypotheses 1 and 2 were accepted; sprint spikes reduced the range of motion and acted as a velocity dampener but resulted in greater moments at the MPJ, sue to increased moment arms. Hypotheses 3 was rejected but hypotheses 4 was accepted; in the sprint spike condition the MPJ energetics improved, with increased energy production during push-off, therefore the controlling effect of the sprint spikes on the rate of motion of the MPJ did not compromise energy generation. Hypotheses 5 was also accepted, adding evidence that sprint spikes facilitate a more effective push-off phase. The sprint spike condition resulted in significantly improved sprinting velocities, indicating the importance of footwear to performance.

CHAPTER 5. THE INFLUENCE OF SPRINT SPIKE STIFFNESS ON SPRINTING PERFORMANCE AND MPJ FUNCTION.

Overview: It has been suggested in the literature that the longitudinal bending stiffness of sprint spike footwear is a factor influencing sprinting performance. Four known sprint spike stiffness conditions used in this performance study did not elicit an improved sprinting velocity for a group of trained sprinters. There was evidence for changes in sprint spike stiffness to elicit individual improvements to sprinting performance and small differences in MPJ kinematics.

5.1 Introduction

The evidence for sprint spikes of increased stiffness to improve sprinting performance is equivocal. Stefanyshyn and Fusco (2004) reported an increase of 0.02s ($p<0.10$) in 20-40 m sprint time for a group of 34 elite national sprinters when 42 N·mm carbon insoles were inserted into their own running spikes. They argued that based on the minimisation of energy loss concept as shoe stiffness increases, the energy lost at the MP joint decreases and performance should increase. This was only found to be true as stiffness increased to a moderate value, after which average performance decreased and this relationship no longer held. However, given the small magnitude of this reported improvement, along with great individual differences amongst athletes, as well as the unknown overall stiffness of the athletes' shoes and the limited number of trials, the validity of their findings is questionable. The stiffness each athlete required for his or her maximal performance was subject specific.

More recently, Ding *et al.* (2011) found no systematic influence of sprint spike stiffness for 25 m acceleration performance, out of the blocks for a group of young competitive athletes. There is no conclusive kinematic or kinetic evidence to support the notion that increasing the stiffness of sprint spikes may lead to an increased sprinting performance, due to their influence on performance related biomechanical and energetic parameters. Therefore, there currently only exists speculation in the research regarding potential mechanisms why sprint performance may be improved by a stiffer sprint plate.

It is currently unknown how foot behaviour and the function of the MPJ differs between varying mechanical properties of sprint spikes in sprinting, therefore whether footwear stiffness is an important factor affecting performance in sprinting needs to be determined. To date, only the effect of sprint spike midsole bending stiffness on sprint times has been reported, whereas understanding of the associated kinematics of the MPJ is needed to help explain any changes in sprint performance.

The focus of this study was to design and mechanically test four different known stiffness conditions, then to test the performance a group of sprinters wearing the modified footwear, in a training environment.

5.2 Objectives

The research question to be addressed in this current study is:
Can increasing sprint spike stiffness lead to significantly improved sprint performance?

Hypothesis 1 states that increasing the stiffness of sprint spikes (up to a certain limiting point) would improve sprinting performance. Hypothesis 2 states that maximal MPJ dorsiflexion would be reduced by the stiffer sprint spikes, thereby limiting the energy absorbed at the joint. Finally hypothesis 3 states that optimal stiffness would differ between subjects.

Specifically, the objectives of the study were:

1) To assess whether a stiffer shoe elicits a significantly improved group sprinting performance

This objective will address hypothesis 1.

2) To assess the controlling influences of different sprint shoe stiffness on the MPJ range of motion during a maximal sprint performance

This objective will address hypothesis 2.

3) To investigate individual responses in sprinting performance to the sprint spike stiffness conditions (performance)

This objective will address hypothesis 3.

5.3 Methods

The methodology of the study comprised of two components: the fabrication of insoles and mechanical testing of the sprint spikes, followed by performance testing of the four stiffness conditions, including measurements of sprint performance and MPJ kinematics.

5.3.1 Insoles

Commercially available sprint Spikes (Puma Complete Theseus II) were provided by Puma (Puma SE, Herzogenaurach) in sizes 43 and 38.5. Shoe lasts provided by Puma were used as a mould for creating insoles. Four different longitudinal stiffness conditions were created: identified as conditions 1, 2, 3 and 4. Each stiffness condition used a different pair of sprint spikes. Firstly the standard sprint spike was used incorporating the 3 mm sock liner (condition 1). Insoles were made for conditions 2, 3 and 4 by combining layers of glass fibre and carbon fibre (Table 5.1). Layers were glued using Expoxi Laminate Resin and set in a vacuum for approximately 24 hours. These insoles were then cut to mimic the shape of the sock liner, glued to it and placed in the sprint spikes (four pairs of sprint spikes were used for each shoe size). The total thickness of the insoles and sock liners was approximately 6 mm. The sprint spikes were colour coded with a green, orange, red or black (for stiffness conditions 1, 2, 3 and 4 respectively) sticker

marking placed on the heel and side of the sprint spikes, otherwise all conditions were identical.

Table 5.1. Layer components of the insoles, sprint spike conditions 2, 3 and 4.
AG = abreisgewebe: nylon fabric, which was torn off
K = Koperbindung: twill weave (over and under)
DU = unidirectional fibres, longitudinal direction.

Layer	Condition 2			Condition 3			Condition 4		
	Construction	Material	Fibre Direction	Construction	Material	Fibre Direction	Construction	Material	Fibre Direction
Surface	AG			AG			AG		
3							K	carbon	0-90
2				K	carbon	0-90	DU	glass	0
1	K	carbon	0-90	K	carbon	0-90	K	carbon	0-90
0	DU	glass	0						
-1	K	carbon	0-90	K	carbon	0-90	K	carbon	0-90
-2				K	carbon	0-90	DU	glass	0
-3							K	carbon	0-90
Surface	AG			AG			AG		

5.3.2 *Mechanical testing*

The bending stiffness of the four different pairs of sprint spikes including insoles were measured mechanically, using a two point bending test. A Servo hydraulic material testing machine was used (Zwick GmbH & Co. KG, Ulm, Germany, stroke 100 mm, load max. 10 kN) with a LVDT position sensor and a 10 kN load cell (Huppert GmbH Prüf- und Messtechnik, Herrenberg, Germany).

Before testing the sprint spikes were stored in the laboratory for at least 24 h, providing conditioning regarding room temperature (23°C) and humidity. Sprint spikes were placed in a mould (Figure 5.1) and secured with a metal clamp on the forefoot (outer shoe). Inside the forefoot section of the sprint spike, a wooden object, shaped as the front section of the shoe last was placed. This was used to ensure a fixed position (seen from toe-box of the shoe) of the mechanism that clamped the shoe to the mould and it resulted in a straight line, around which the shoe was able to bend. The heel of the sprint spike was placed on the metal bar of the mechanical testing device, with the initial angle dictated by the pre-tension of the shoe itself. The force and the distance moved throughout the 40 mm dorsiflexion of the shoe were measured and from this, the average stiffness was calculated.

The sprint spikes underwent 40 mm of bending at a constant velocity of 10 mm/s. These values were chosen based upon the angular displacement and velocities of the MPJ in the sprint spike condition from Chapter four. The average stiffness (N/m) for a deformation of 0 – 40 mm was measured. Three trials per condition were recorded and averaged, left and right shoes were measured and averaged (the average standard deviations were 15 N/m and 10 N/m between the three trials, variability between left and right shoes is presented in Table 5.4). In addition, three commercially available sprint spikes (Nike Zoom Mazcat II sprint spike, Adidas Techstar Meteor Sprint and Asics

Hyper Sprint Spikes) in size 43 were also mechanically tested using the same testing conditions.

Figure 5.1. Mechanical testing device used for measurement of sprint spike bending stiffness.

Performance testing

5.3.3 Participants

Ethical approval was granted for this study from the Institute of Sports Science, University of Chemnitz and Liverpool John Moores University, as per Chapter 3. Following attainment of informed written consent, twelve participants took part in the performance testing and their characteristics are in Table 5.2. From the twelve participants, nine were trained sprinters who trained three times a week or more. The other three were trained in a sprint related sport (football, rugby). All subjects had shoe size 43 or 38.5.

Table 5.2. Participant characteristics

	Males	*Females*
N	6	6
Age (years)	22.0 (± 3.6)	21.8 (± 4.8)
Height (cm)	182.7 (± 4.5)	167.0 (± 6.2)
Weight (kg)	73.8 (± 4.4)	60.2 (± 4.2)

5.3.4 Protocol

Sprinters performed eight maximal 40 m sprints from a standing start along an indoor 100 m straight, synthetic track surface (indoor athletics training facility). Single beam timing cells (resolution 0.01s) were located at 10, 20, 30 and 40 m from the start, in line with the sprinters' hip height in accordance with Yeadon, Yato and Kerwin (1999).

Prior to data collection, one trained female sprinter performed eight maximal 40 m sprints wearing the same sprint spikes, in order to ascertain the maximal number of sprints appropriate before fatigue, the appropriate rest needed in between trials and determine typical trial to trial variability. The subject accelerated throughout the 40 m, recording highest velocities from 30 to 40 m. Therefore this section of the sprint was chosen to represent highest velocity. Mean 10 m split time and velocity for eight trials were 1.19 s (± 0.02 s) and 8.43 m/s (±0.11 m/s) respectively and the ranges were 0.04 s and 0.29 m/s, demonstrating low levels of variation. It was also determined that eight trials was the appropriate maximum number of trials and eight minutes rest was more than sufficient to avoid fatigue effecting the sprint times.

One high speed video camera (Exilim Pro Ex-F1) was located perpendicular to the direction of the sprint, at ground level, at approximately the 15 m point, during the early acceleration phase of sprinting and captured two dimensional sagittal plane kinematic data of the foot for one right foot contact, The camera sampled at 600 Hz, with manual focus and 1/1000s shutter speed and extra lighting was used to brighten the field of view. The position of the camera was adjusted after the athlete performed practice runs, to ensure the right foot contact was captured, and then the video was calibrated using a 1 m x 1 m, L-shaped calibration frame with vertical and horizontal points at 20 cm, placed in the centre of the field of view.

The sprinters performed their own warm up wearing their own sprint spikes, included two practice strides / runs at approximately 80-90% maximum effort. The order of conditions was mixed with counter balancing and subjects did not know which stiffness condition they were assigned. They performed two maximal sprints with each shoe before moving onto the next shoe condition (thereby running a total of 8 sprint trials). Participants were instructed to sprint flat out through the 40 m mark. They had 8 minutes rest between each trial to allow for full recovery of the anaerobic energy systems (McCartney *et al.*, 1986). After each trial, split times at 10, 20, 30 and 40 m were recorded.

Following all 8 trials, the subjects were instructed to complete a subjective questionnaire, ranking the shoes in order of performance with their reasons for the ranking and also ranking the shoes in order of stiffness and stating whether the shoes were too stiff / too flexible / ok (see Appendix F).

5.3.5 Data processing

The sprint times of the two trials for each shoe condition were averaged. High speed video data was collected for ten participants, out of which 4 participants had one successful right footfall within the camera's field of view, for each of the four stiffness conditions. High speed videos were imported into Quintic Biomechanics (version 9.03 v17, Quintic Consultancy, UK) where they were manually digitised frame by frame (approximate resolution 430 * 130 pixels). Three medial points on the sprint spike were digitised: the heel, the distal end of the hallux and the first MPJ centre (head of metatarsal 1). The medial aspect of the foot was chosen, based on pressure results from previous studies and resulting joint kinetic results where the MPJ was modelled using different joint line definitions. The MPJ angle was therefore defined as the angle between the forefoot (MTH1 – hallux) and rearfoot (MTH1- heel) segments. Digitized pixels were converted to linear measurements and the aspect ratio of the recorded fields was maintained. The MPJ angle was calculated for the entire stance phase and angular velocity was also calculated by differentiation of the joint angle. The raw kinematic data was smoothed using a 4^{th} order Butterworth low pass digital filter at different cut off frequencies. From visual inspection of the resulting MPJ angles and angular velocities the cut off frequencies of 60 Hz for the MPJ angle and 30 Hz for the angular velocity were chosen (see Appendix D). Data consistency was assessed through re-digitizing approximately 10% of high-speed video files captured. Root mean square values of the MPJ angle were calculated for the initial and repeated digitisation. Precision of the manual digitisation process was calculated by determining the smallest possible change in MPJ angle between two consecutive frames with one pixel of heel movement.

5.3.6 Statistical Analysis

All statistical analyses were performed using SPSS for Windows version 16.0 and the level of significance was set at $p \leq 0.05$. Following checks for the assumptions of normality and sphericity of the data, a one way repeated measures ANOVA was conducted to evaluate the intervention of sprint spikes stiffness on sprint time, MPJ range of motion, MPJ maximum flexion and MPJ peak angular velocities, both plantarflexion and dorsiflexion.

5.4 Results

5.4.1 Mechanical stiffness

The stiffness of the four stiffness conditions (both sprint spikes sizes 38.5 and 43) used in this study are demonstrated in Table 5.3. The mechanical stiffness's of the three additional commercially available sprint spikes are shown in Table 5.4. There was some, although low, variability due to small differences between left and right shoes and also slight differences in stiffness conditions between shoe sizes, despite identical construction of the insoles.

Table 5.3. Average stiffness and stiffness normalised to condition 1 (sprint spike, no insole) of four stiffness conditions used in the study, left and right shoes.

Stiffness of Puma sprint spike plus Insole condition	Shoe size 38.5 Average Mechanical stiffness (N/m) of left and right shoe and Normalised stiffness (%)	Shoe size 43 Average Mechanical stiffness (N/m) of left and right shoe and Normalised stiffness (%)
1	254.6 ± 22.7 (100%)	297.4 ± 7.6 (100%)
2	314.6 ± 7.8 (124%	343.1 ± 6.0 (115%)
3	367.0 ± 19.7 (144%)	408.6 10.5 (± 137%)
4	501.7 ± 43.6 (197%)	472.1 ± 31.6 (159%)

Table 5.4. Average stiffness of three additional, commercially available sprint spikes

Sprint Spike	Shoe size 43 Average Mechanical stiffness (N/m) of left and right shoe
Adidas Techstar Meteor Sprint	190.5 ± 5.3
Asics Hyper sprint	197.9 ± 29.6
Nike Zoom Mazcat II	256.1 ± 23.7

5.4.2 Sprint performance

Average 30 m to 40 m sprint time for all trials was 1.18 s (± 0.08 s), corresponding to a mean velocity of 8.50 m/s (± 0.57 m/s). Stefanyshyn and Fusco (2004) reported mean velocities of 9.2 m/s – 9.3 m/s for 20 m to 40 m splits for elite athletes. Figure 5.2 demonstrates mean sprint velocities for the four different stiffness conditions. There were no significant ($p<0.05$) differences in mean sprinting velocity between stiffness conditions. Individual differences existed between sprinters with some sprinters performing better with the sprint spike alone and others performing better with stiff insoles. However, there were no trends for the majority of subjects, therefore different results were highly individualised. Table 5.5 demonstrates that the mean sprint time was exactly the same for stiffness condition 1, 2 and 3 with stiffness condition 4 marginally slower. However, no one participant demonstrated this same result, although half of the subjects did produce their fastest condition in two different conditions. Typical within-participant variation between the two sprint trials in each condition was very low – 0 to 0.03 s and between participant variation was also relatively low – 0.08 s and consistent between stiffness conditions (Table 5.5).

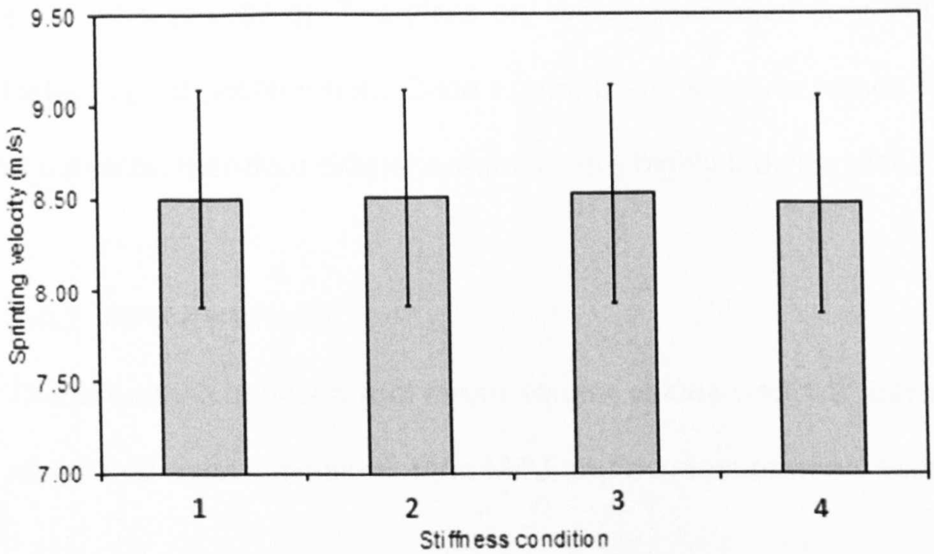


Figure 5.2. Mean (± s.d.) sprinting velocity for twelve sprinting subjects performing two sprints in each stiffness condition.

Table 5.5. Mean 30 to 40 m sprint time for each participant (numbered 1 -12) in the four different stiffness conditions. Fastest condition or 2 conditions is highlighted in bold for each subject.

Participant	1	2	3	4	5	6	7	8	9	10	11	12	Mean (±s.d.)
Stiffness condition 1	1.19	1.29	**1.21**	1.13	1.06	**1.06**	**1.17**	1.25	1.11	1.17	**1.30**	**1.24**	1.18 (±0.08)
Stiffness condition 2	1.20	**1.25**	**1.21**	1.12	**1.05**	1.07	**1.17**	1.26	**1.10**	1.18	1.32	**1.24**	1.18 (±0.08)
Stiffness condition 3	**1.11**	1.26	1.23	**1.09**	1.09	1.08	1.19	**1.24**	1.12	**1.16**	1.31	1.25	1.19 (±0.08)
Stiffness condition 4	1.18	**1.25**	1.22	**1.09**	1.07	1.09	1.20	1.27	**1.10**	1.18	1.33	1.25	1.18 (±0.08)

Average 10 m to 20 m sprint time for all trials was 1.30 s (± 0.06 s), corresponding to a mean velocity of 7.70 m/s (± 0.39 m/s). Athletes were accelerating during this phase. Johnson and Buckley (2001) reported mean velocities of 8.66 m/s (± 0.37 m/s) at the 14 m for their elite male 100 m athletes. There were no significant differences ($p<0.05$) in mean 10 to 20 m sprint time between the four different conditions. Mean sprint time for stiffness condition 1, 2, 3 and 4 were 1.30 s (± 0.06 s), 1.30 s (± 0.07 s), 1.29 s (± 0.07

s), 1.30 s (± 0.07 s). Therefore on average stiffness condition 3 resulted in fastest sprint acceleration. Once again, there were no trends for the majority of subjects, therefore different results were highly individualised.

5.4.3 MPJ Kinematics

The difference between root mean square values was 1.2° between initial and redigitised data. Precision of the MPJ angle measurements was 0.5°.

The mean MPJ angular range of motion during stance (Figure 5.3) was reduced in the stiffer sprint spikes with mean ranges of motion 40.5 ° (± 3.0 °), 38.1 ° (± 4.5 °), 38.3 ° (± 2.8 °) and 37.5 ° (± 4.7 °) for stiffness conditions 1, 2, 3 and 4 respectively. This difference was also evident in the peak dorsiflexion of the MPJ which was 44.6 ° (± 5.9 °) in the stiffest shoe, compared to 48.4 ° (± 5.6 °) in the most flexible condition. However, none of these differences were not significant ($p < 0.05$).

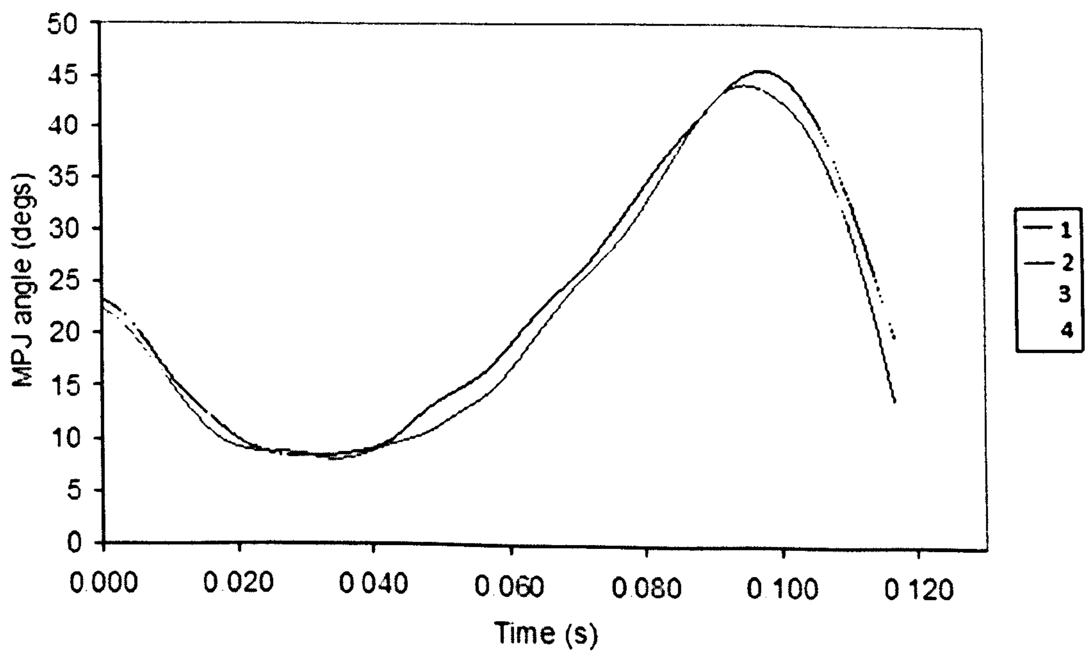


Figure 5.3. Mean MPJ angle during stance for four stiffness conditions (1,2,3,4) four participants.

Overall, there were no differences in the maximum dorsiflexion velocity (Figure 5.4) between the four different stiffness conditions, for the four participants tested. Mean peak dorsiflexion velocity ranged from 987.9 °/s (± 59.8 °/s) in stiffness condition 3 to 1062.1 °/s (± 47.6 °/s) in stiffness condition 2. There were no statistical differences in maximum MPJ dorsiflexion angular velocities across the four stiffness conditions ($p < 0.05$). The MPJ continued to plantarflex after take-off. Peak MPJ plantarflexion velocity occurred approximately 5 ms after take-off. Figure 5.5 demonstrates there were small differences in peak MPJ plantarflexion velocity after take-off, with increasing stiffness conditions resulting in reduced peak MPJ plantarflexion velocities. However, there were no statistical differences in maximum plantarflexion angular velocities across the four stiffness conditions ($p < 0.05$).

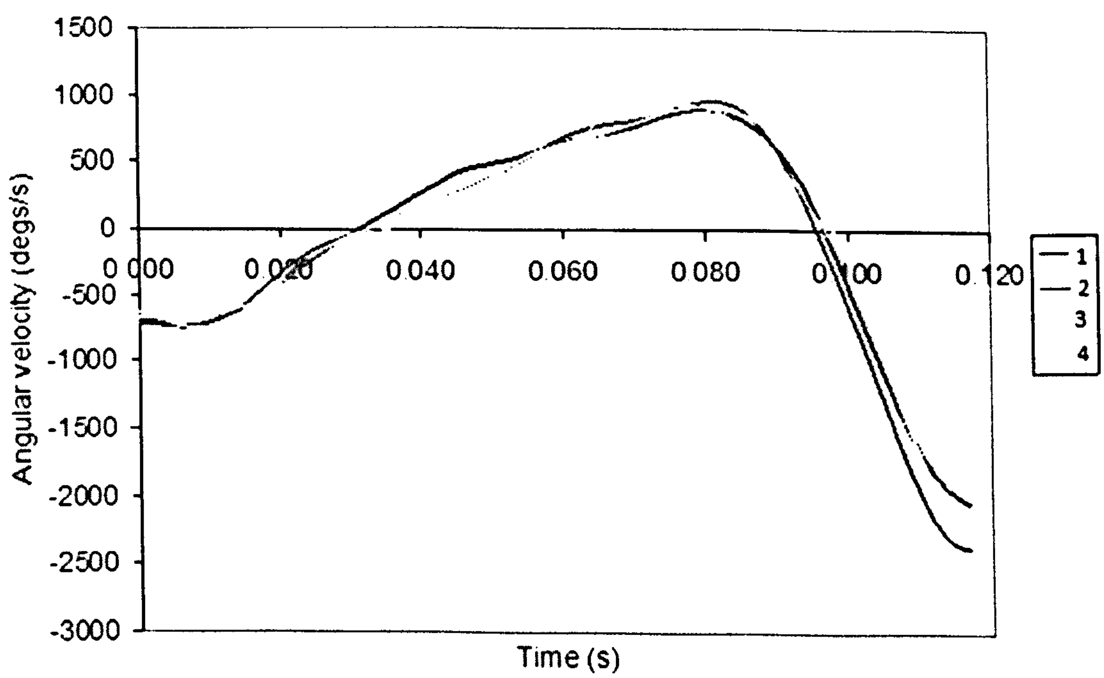


Figure 5.4. Mean MPJ angular velocity during stance for four stiffness conditions (1,2,3,4) for four participants. Positive values reflect MPJ dorsiflexion.

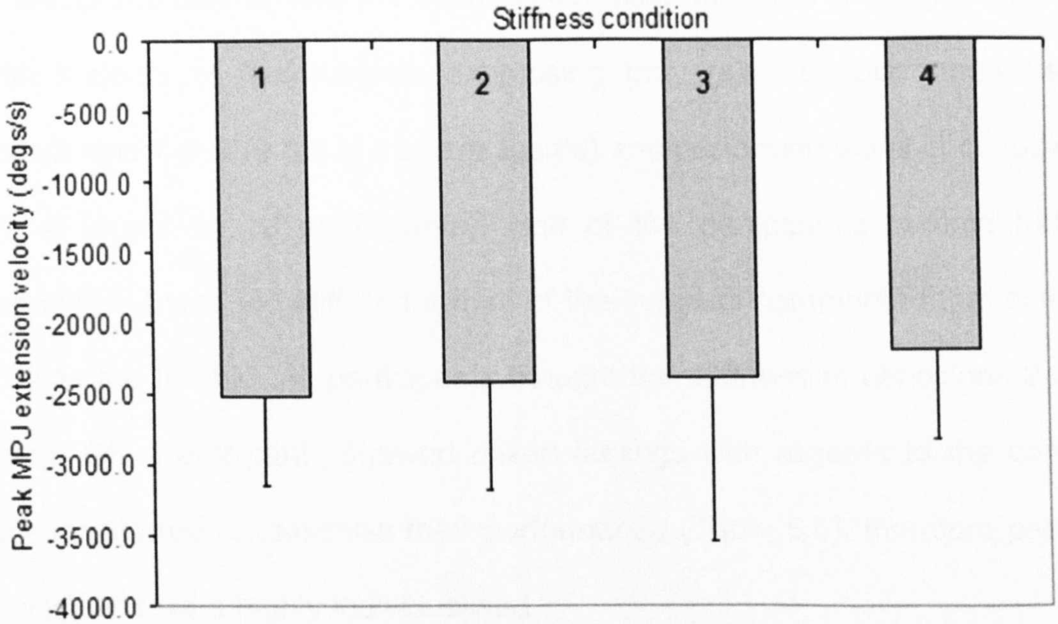


Figure 5.5. Peak MPJ plantarflexion velocities for four stiffness conditions, four participants.

5.4.4 *Perceived effects*

Out of the twelve participants, ten correctly identified that condition 4 had the highest stiffness (Figure 5.6). Fewer participants were able to correctly identify differences between conditions 1, 2 and 3.

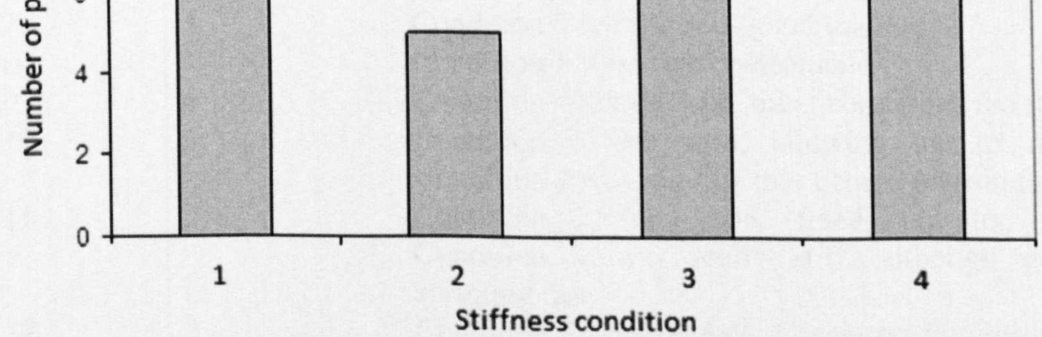


Figure 5.6 Number of participants (out of 12) that correctly identified the relative stiffness of the shoe conditions.

Participants also ranked the stiffness condition in terms of performance, with the majority of the subjects expressing they felt they performed best in conditions 1 or 2 (9 out of 12 participants) and performed worst in conditions 3 or 4 (8 out of 12 participants). Half of the participants commented that condition 4 was too stiff and a third of the subjects commented that condition 1 was too flexible. All participants thought the stiffness of conditions 2 and 3 were OK. Participants showed mixed feelings with regards to the condition they preferred to maximise their performance (Table 5.6), therefore preferred conditions were highly individualised.

Table 5.6 Individual perception comments of the stiffness conditions 0, 1, 2, and 3

Participant	*Preferred condition for performance*	*Comment*
1	1	Flexible conditions felt more comfortable, condition 4 too hard
2	2	Condition 2 was best, allowing me to run on my toes, conditions 3 and 4 were too stiff and heavy
3	2	Condition 2 felt best, allowing for pressure on the balls of my feet. Condition 4 created an off-balance feeling during acceleration but was felt good for maximal sprinting
4	1	Condition 1 was the most comfortable, had good stability, condition 4 was too stiff
5	None	No preference / differences between conditions
6	2	Condition 2 most comfortable but little differences
7	3	Condition 3 felt the best, good traction
8	1	Condition 1 was most comfortable
9	4	Condition 4 felt the best, other conditions too flexible
10	3	Condition 3 felt hard, allowing me to run fast. Condition 4 was too stiff that I could not run fast
11	2	Condition 2 felt good, fixed, not too flexible. Condition 4 was really stiff, although good for sprinting fast.
12	2	Condition 2 was perfect, 1 was too flexible, 3 and 4 were too stiff

5.4.5 *Individual results*

When considering individual results, for 30 – 40 m sprint times, 7 participants demonstrated improved performance in a stiffer insole condition compared to the standard sprint shoe condition, the other 5 participants had their best performance in the standard condition. 6 participants recorded their equal best 20 – 40 m sprint time in two shoe conditions. Therefore from table 5.5, the number of participants that recorded their best 30 to 40 m sprint time for conditions 1, 2, 3 and 4 respectively were: 5, 6, 4 and 3. Examples of four individual's results are shown below. Participant 5 performed best in stiffness condition 2 (Figure 5.7) for both phases of sprinting,10-20 m and 30-40 m. Peak MPJ dorsiflexion was reduced for each stiffness condition (Figure 5.8) with a decrease of 5.8° in the stiffest condition compared to the standard sprint spike and peak MPJ dorsiflexion velocity was also reduced in the stiffer conditions (Figure 5.9). Participant 5 did not detect differences in the stiffness of the sprint shoes, although did correctly identify they had their best performance in stiffness condition 2.

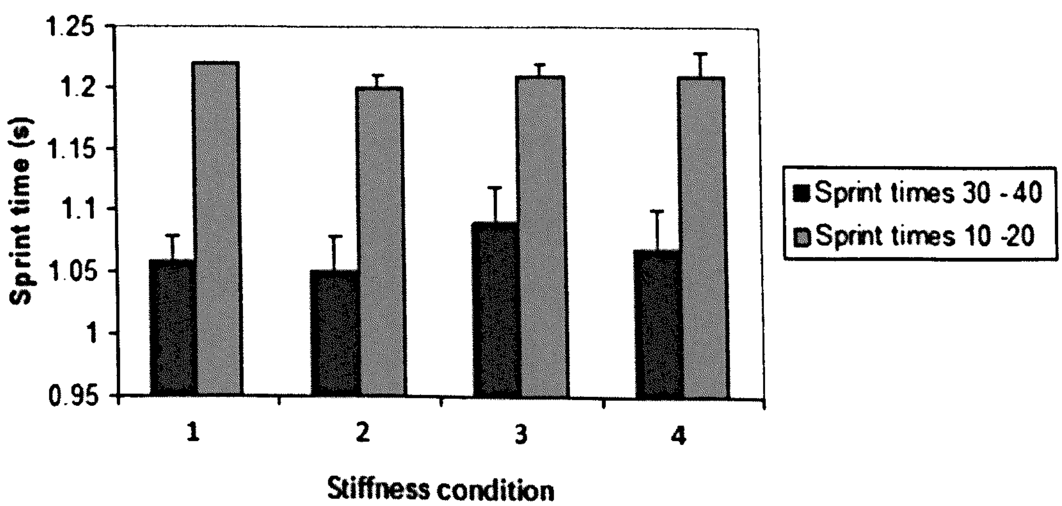


Figure 5.7. 10 to 20 m and 30 to 40 m sprint times for participant 5 in four stiffness conditions

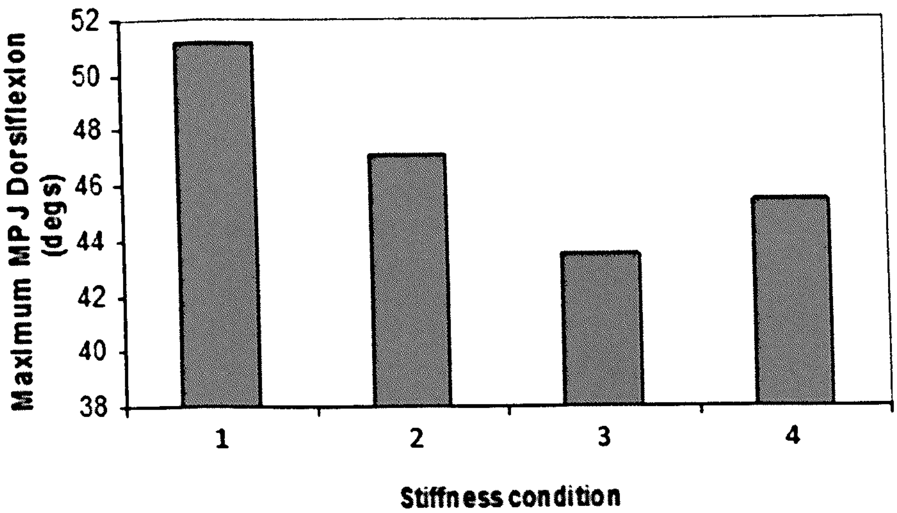


Figure 5.8. Maximum dorsiflexion angle of the MPJ for participant 5 for four different stiffness conditions

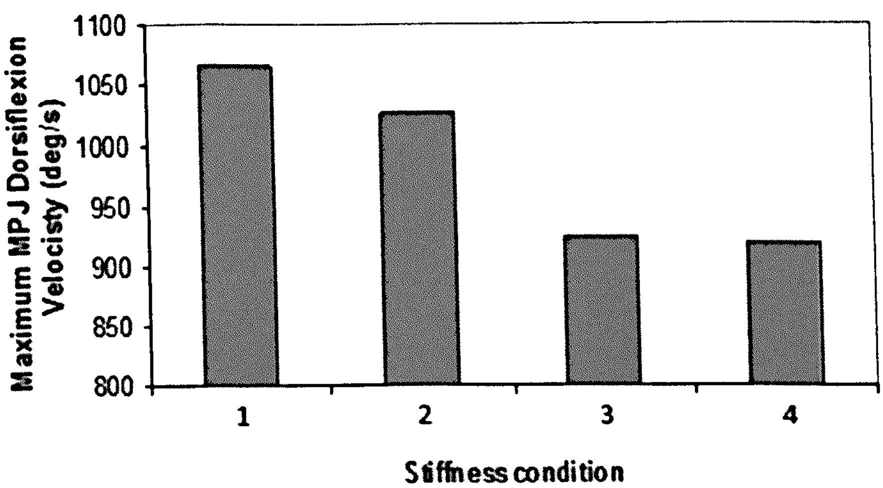


Figure 5.9. Maximum dorsiflexion velocity of the MPJ for participant 5 for four different stiffness conditions

Participant 3 demonstrated no differences in sprinting performance between the four stiffness conditions at 10 and 20 m, although at 30 to 40 m, conditions 3 and 4 resulted in slower performance (Figure 5.10). Maximum MPJ dorsiflexion was reduced in condition 4 by 4.2° compared to the standard sprint spike (Figure 5.11). Maximum MPJ dorsiflexion velocity did not follow the same pattern as maximum dorsiflexion angle and was greater in the stiffer conditions: 2, 3 and 4 compared to the standard sprint spike (Figure 5.12). However this was clearly not a systematic increase with increased stiffness.

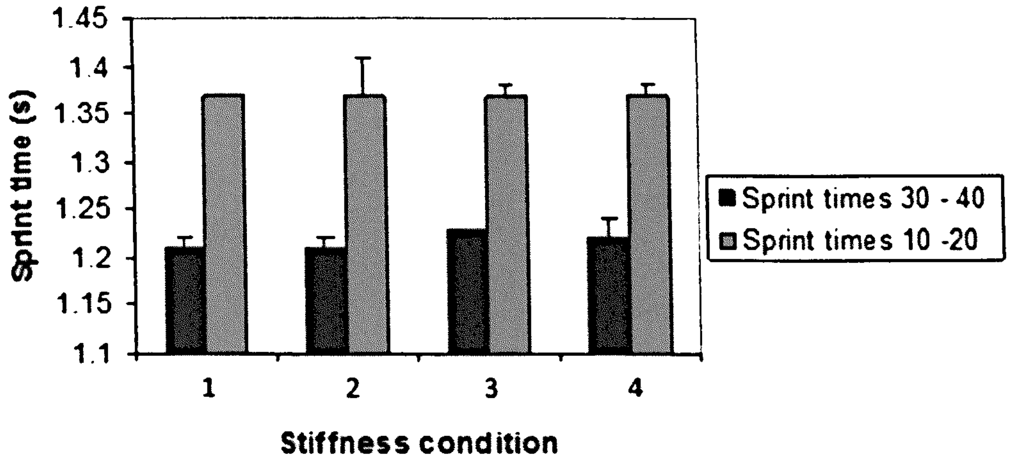


Figure 5.10. 10 to 20 m and 30 to 40 m sprint times for participant 3 in four stiffness condition.

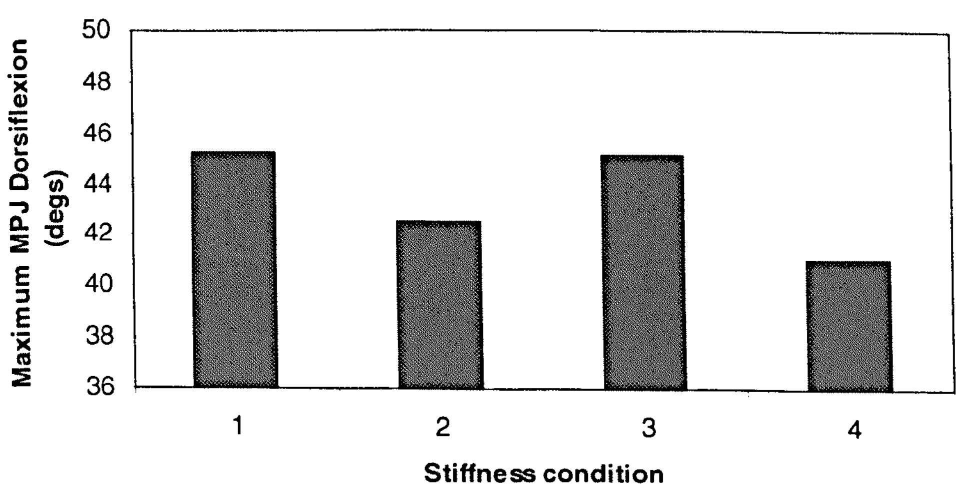


Figure 5.11 Maximum dorsiflexion angle of the MPJ for participant 3 in four different stiffness conditions

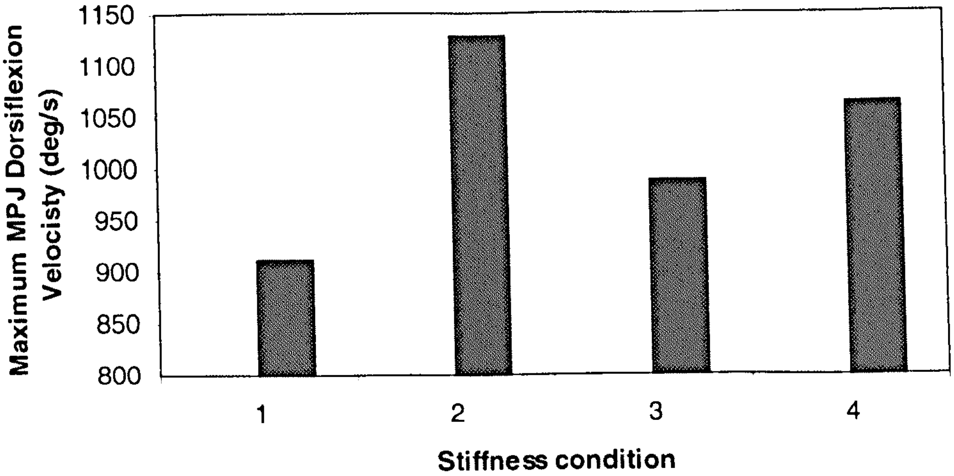


Figure 5.12 Maximum dorsiflexion velocity of the MPJ for participant 3 in four different stiffness conditions

Participant 9 demonstrated little difference in sprint times for 10 to 20 m but exhibited faster sprint times in condition 2 for 30 – 40 m (Figure 5.13). Maximum MPJ dorsiflexion was reduced in the stiffer shoes, although this reduction was not systematic (Figure 5.14) and the difference between stiffness condition 1 and 4 was only 2°. Maximum dorsiflexion velocity was again reduced by the stiffer shoes (Figure 5.15).

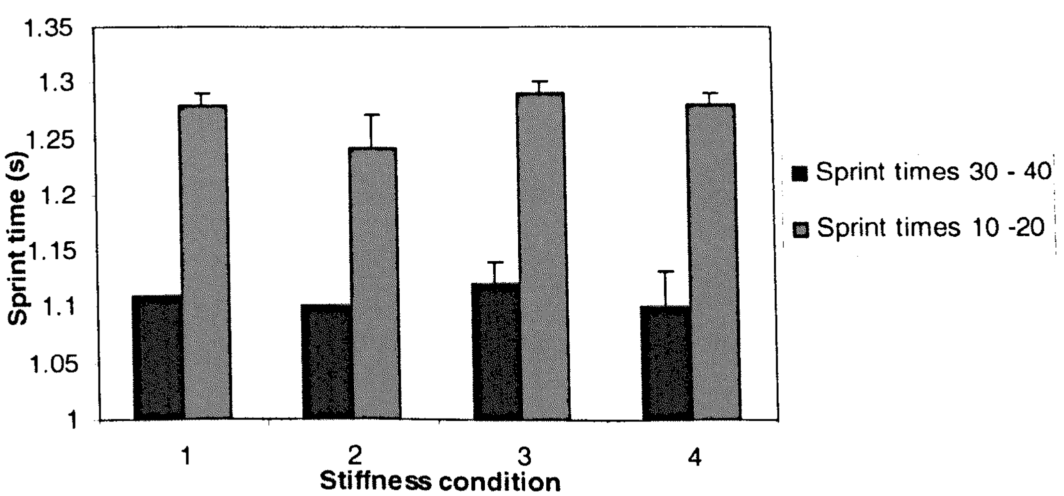


Figure 5.13 10 to 20 m and 30 to 40 m sprint times for participant 9 in four stiffness conditions

Figure 5.14 Maximum dorsiflexion angle of the MPJ for participant 9 in four different stiffness conditions

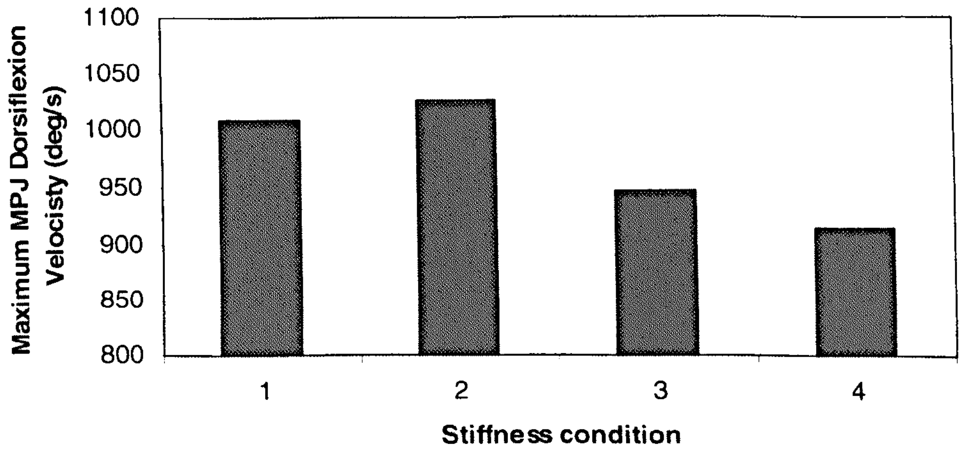


Figure 5.15 Maximum dorsiflexion velocity of the MPJ for participant 9 in four different stiffness conditions

Participant 12 exhibited slightly faster sprint times at 30 to 40 m for more flexible conditions 1 and 2, although conversely demonstrated fastest sprint times at 10 to 20 m for stiffness condition 4 (Figure 5.16). There were minimal differences in MPJ maximum dorsiflexion between conditions 2, 3 and 4 (Figure 5.17), although maximum dorsiflexion velocity was greatest for the stiffest shoe (Figure 5.18).

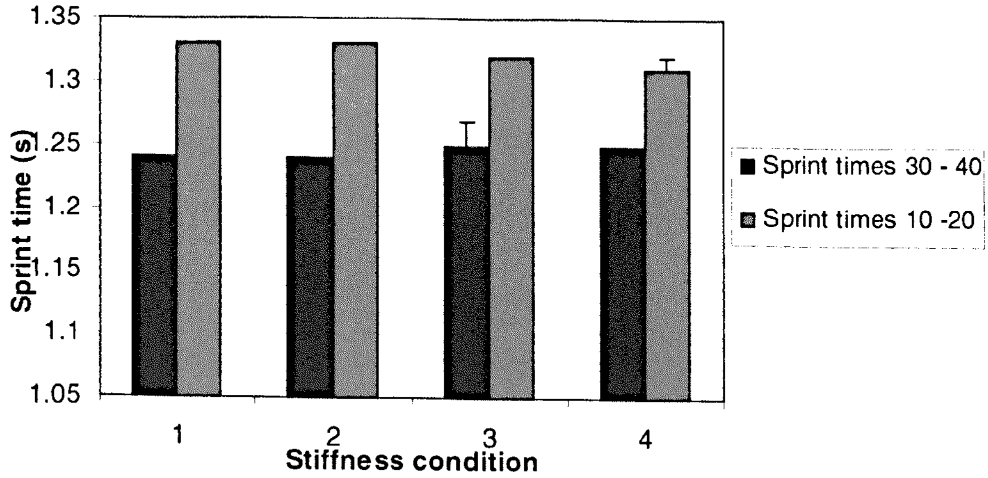


Figure 5.16 10 to 20 m and 30 to 40 m sprint times for participant 12 in four stiffness conditions

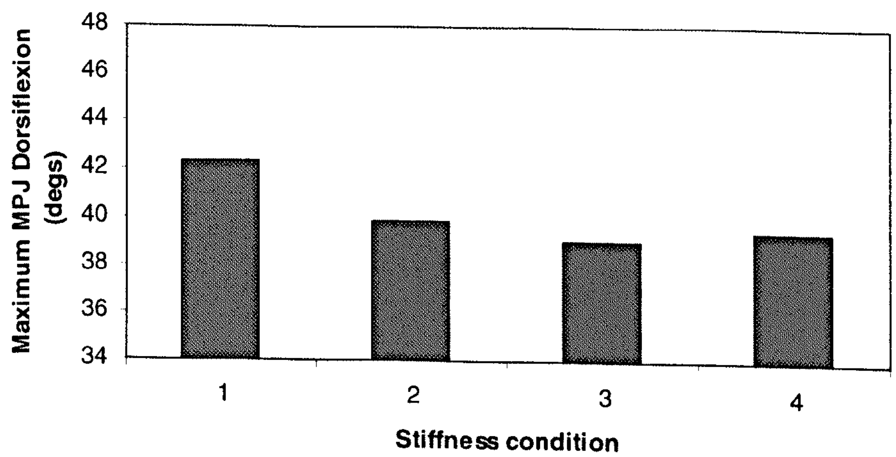


Figure 5.17 Maximum dorsiflexion angle of the MPJ for participant 12 in four different stiffness conditions

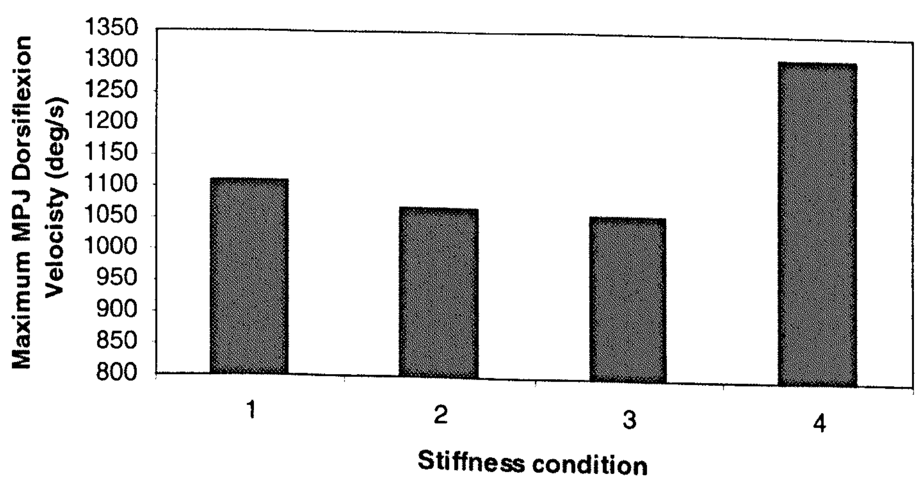


Figure 5.18 Maximum dorsiflexion velocity of the MPJ for participant 12 in four different stiffness conditions

5.5 Discussion

This study addressed the research question *Can increasing sprint spike stiffness lead to significantly improved sprint performance?* through quantifying the effect of sprint spike bending stiffness for four known stiffness conditions on both sprinting performance and the kinematics of the MPJ for a group of trained sprinters.

5.5.1 Objectives

Objective 1) To assess whether a stiffer shoe elicits a significantly improved group sprinting performance

The outcome of this study does not agree with the notion that increased bending stiffness improves sprinting performance, therefore hypothesis 1 is rejected. The findings of this study disagree with Stefanyshyn and Fusco (2004) who found that, on average, 20 m to 40 m group sprint times were significantly reduced when wearing a sprint shoe with a plate stiffness of 42 $N.mm^{-1}$, compared to the athletes' own standard sprint spikes.

Comparing the results of this study to the study by Stefanyshyn and Fusco (2004) several key differences were noted, possibly accounting for the different outcomes of the study. Firstly, it is not possible to compare the mechanical stiffness of the sprint spike conditions used, particularly as the participants in their study used their own sprint spikes with unknown stiffness. In this study, the stiffness of the standard Puma sprint spike (condition 1) was greater than the three other commercially available sprint spikes that were mechanically tested. It is possible that the stiffness conditions used were too great and that increasing the stiffness of the shoe beyond the stiffness of the original Puma sprint spike has no additional benefits on sprint performance. Secondly, the sprinting velocities of the athletes used by Stefanyshyn and Fusco (2004) were on average 0.7 – 0.8 m/s higher than the subjects used in this study, reflecting the difference in performer level and age of the athletes used. The age and level of the athletes used in this study may be one reason why some of the participants got slower throughout the series of sprints, demonstrating evidence of fatigue, even with plentiful rest in between trials.

Any potential differences between stiffness conditions may have been masked by the variability between trials, However, most participants showed only small differences between the two trials in the same conditions, with 30 m to 40 m sprint times ranging by just 0.01 – 0.02s, this suggests that two trials per condition was sufficient. Hopker, Coleman, Wiles and Galbraith (2009) indicated that reliable data can be derived for single maximal sprint measures, using fixed distance protocols. Hunter, Marshall and McNair (2004) reported the reliability of various biomechanical variables in sprint running. The variables related to the horizontal velocity of the centre of mass had the greatest reliability, sprint velocity had a CV of 1.0% (one trial) but with five trials, the CV improved to 0.4%.

During pilot testing, where one subject performed eight maximal sprints in the same footwear condition, variability between trials was low (standard deviation

0.02s, range 0.04s), however, the magnitude of expected improvements in sprint times were also low. Stefanyshyn and Fusco (2004) stated a mean difference in 20 m sprint time of just 0.02 s (equivalent of 0.01s) between the standard sprint shoe and the stiffness condition which significantly improved performance ($p < 0.10$). This difference was significant probably due to an increased alpha level, a higher number of subjects used (34 sprinters) and a more homogeneous subject group. In their pilot testing, 20 m to 40 m sprint times (nine trials, standard shoe) ranged by 0.02 s with a standard deviation of 0.008s. With the resolution of the timing system of 0.01 s, it is questionable whether a difference of 0.02 s (0.69%) reflects a significant improvement in performance between a standard shoe and a stiff shoe. However, Stefanyshyn and Fusco (2004) argued that performance enhancements of 0.7% are quite large in elite sprinting, as Hopkins, Hawley and Burke (1999) suggested that the smallest performance enhancement (equivalent of 0.36 and 0.63%) is worthwhile for an athlete at an elite level and may make a difference in finishing position in a race.

No shoe differences were detected in this study, which may be the result of insufficient statistical power due to one or more of: low subject numbers; low numbers of trials; low effect size; and high variability. The number of subjects and trials used in this study was not dissimilar to previous biomechanical studies. Stefanyshyn and Fusco (2004) used the same number of trials but had greater statistical power due to a higher number of subjects (n=34). Guidelines on necessary sample sizes (Cohen, 1992) demonstrate that using an alpha level of ≤ 0.05 and statistical power ≥ 0.8 the minimum number of subjects required for a medium effect size of r = 0.3 and a large effect size of r = 0.5 is 85 and 28 respectively.

Whilst a double beam photocell system would have been ideal, the single beam timing gates were aligned with hip height with a large break criterion, following guidelines set by Cronin and Templeton (2008) and Yeadon, Kato and Kerwin (1999), who reported errors of 0.1 m/s using similar system set up. Therefore with such small improvements in sprint times expected, it is likely that the variability between trials of the subjects masked any differences in sprinting performances.

Objective 2) To assess the controlling influences of different sprint shoe stiffness on the MPJ range of motion during a maximal sprint performance

Increasing the bending stiffness of the sprint spikes resulted in reduced angular range of motion at the MPJ. However, there were no statistically significant differences between the stiffness conditions, therefore hypothesis two is rejected. Mean values were between 37.5° and 40.5°. These values compare relatively well to the mean value of 42.6° (± 8.2°) for MPJ angular range of motion recorded using high speed three dimensional analysis for subjects wearing standardised Nike zoom sprint spikes (lower stiffness) reported in Chapter 3. Kleindienst *et al.* (2005) also reported that increasing midsole bending stiffness of running shoes resulted in restricted dorsiflexion at the MPJ, in subjects running at speeds of 3.0 m/s and 4.5 m/s. Toon *et al.* (2009) reported mean peak values at the 50m point of a sprint of 35° (± 8°), however used aggregated medial and lateral data, although it has been previously shown that the medial and lateral aspects of the MPJ do not undergo similar ranges of motion. It is suggested that the sprinters were unable to generate the additional forces required to overcome the bending stiffness of the sprint spikes, although this will need confirming with kinetic data.

Sprint spike stiffness did not appear to effect maximum dorsiflexion velocity at the MPJ. Values recorded ranged from 987.9 °/s to 1062.1 °/s, agreeing with Krell and Stefanyshyn who reported values between 600 °/s and 1400 °/s for elite Olympic athletes, whereas Toon *et al.* (2009) reported no significant differences in mean values of 625 °/s for barefoot and 634 °/s for sprint spike conditions. Whilst the MPJ dorsiflexes and the heel is raised, during this phase, energy is absorbed at the MPJ. Stefanyshyn and Fusco (2004) suggested that as shoe stiffness increases, the energy lost at the MPJ during this phase may decrease and consequently performance should increase. However, sprint spike stiffness did not affect the rate of MPJ dorsiflexion during this energy absorbing phase, suggesting that the moment arms and forces generated at the MPJ are more influential, if this is to be the case.

The MPJ extended during the push off phase and reached maximum plantarflexion velocity a few frames after the instant of take-off. Stiffer sprint spikes did appear to affect the rate of the MPJ plantarflexion, with the stiffest sprint spike reducing the peak extension by approximately 300 °/s compared to the standard spike condition. This suggests that during the take-off phase, there is some evidence of MPJ plantarflexion (although complete plantar flexion does not occur until after ground contact) and consequently a small opportunity to generate energy at the MPJ. However, stiffer sprint spikes compromise angular range and velocity during this phase. Therefore, sprint spikes appear to have controlling influences over the kinematics of foot segments.

One limitation of the kinematic analyses was the motion of the foot was analysed for just four subjects, due to ground contact often occurring outside of the field of view of the camera. Secondly, the foot-shoe complex was digitised to present movement of the foot. Landmarks on the surface of the sprint spikes were digitised, but subject to human error during the process. The sprint spikes were tightly secured to the feet, therefore relative movement between the foot and the shoe were tightly coupled. As video data was captured at approximately the 15 m mark, kinematic results cannot be generalised to maximal sprinting. It is possible that at maximal sprinting, the influence of bending stiffness of the sprint spikes is reduced. Whilst this study reported the kinematics of the MPJ during sprinting, the importance of including kinetic measurements in the analysis of sprint running performance is great. Increasing the bending stiffness of sprint spikes may increase effective lever length and the resultant MPJ moment to facilitate a more effective propulsive system, as long as the athlete can generate enough force to overcome the shoe stiffness. In contrast, this may compromise the start and acceleration phases, when a shorter effective lever length might be more suitable. The application of appropriate bending stiffness could have a positive effect on MPJ joint moments and subsequently on injury prevention.

Objective 3) To investigate individual responses in sprinting performance to the sprint spike stiffness conditions (performance)

The results from this study do agree with the suggestion that the stiffness each athlete requires to produce maximal performance is subject specific. Individuals responded differently to the stiffness conditions, thus hypothesis 3 is accepted.

It has been suggested that the only way to detect differences that exist between shoes is to employ a single subject analysis (Bates *et al.,* 1987). With footwear, due to individual anatomical difference and neurological responses, no one shoe can be 'best' for all people, but some shoes appear to be better for more individuals than others (Dufek & Bates, 1991). This was seen to be the case in this study and in fact the average performance profile did not represent individual subjects, supporting the notion that group models can sometimes describe a mythical average performer (Dufek *et al.,* 1995).

Individuals' perceptions of the sprint spike differed greatly in this study, whilst some subjects could identify the stiffest shoes they were unable to identify correctly in which shoes they performed best. Subject variability can be the result of unique strategies / response patterns employed which in turn will depend on the subject's recognition of the perturbation (Bates *et al.,* 2004).

However, in this task it is perhaps unlikely that subjects employed response patterns as many subjects were unable to identify and classify correctly the perturbation or level of perturbation. However, it is possibly more likely that anatomical, biomechanical and physiological differences may account for highly individualised results.

5.6 Study conclusion

In answer to the research question posed: *Can increasing sprint spike stiffness lead to significantly improved sprint performance?* the stiffer sprint spike conditions used in this study did not elicit an improved group sprinting performance, as has been previously reported. For this particular group of trained, albeit relatively young sprinters, increasing sprint spike stiffness did not lead to significantly improved sprint performance, therefore hypothesis 1 was rejected. Stiff sprint spikes did appear to have some controlling influences over the range of motion and the plantarflexion velocity at the MPJ, although differences between conditions were not significant, therefore hypothesis 2 was rejected. The influence of varying the mechanical properties of footwear on the energetics of sprinting performance are still unknown and the resultant kinetics at the foot require further research. The individuals in this study differed greatly in their performance responses to the different shoe conditions. Hypothesis 3 was accepted and furthermore they demonstrated a wide range of different perceptions of the stiffness and performance of the four sprint shoes. Therefore, it is suggested that an athlete's optimal sprint shoe stiffness is highly subject-specific and dependent upon their perceptions and previous experiences.

CHAPTER 6 ATHLETE-SPECIFIC ANALYSES OF THE EFFECT OF SHOE BENDING STIFFNESS ON FOOT FUNCTION AND SPRINTING PERFORMANCE.

Overview: This study investigated the effect of four sprint spike bending stiffness on sprinting velocity, MPJ kinematics and kinetics, and pressure distribution for two participants. It employed a single subject research design with repeated use of a baseline measure. There were clear individualised performance and biomechanical responses to the different stiffness conditions. Participant one demonstrated improved sprinting velocity in the two stiffest sprint shoes, participant two performed best in the second highest stiff shoe. Although the improvements were small, this study suggests that individual sprinting velocity may be improved by implementation of relevant shoe mechanical characteristics.

6.1 Introduction

In the previous study, there was some evidence for individual responses to sprint spikes of varying longitudinal bending stiffness. The stiffness each athlete required to produce maximal sprinting performance was subject-specific. Therefore, the notion of improving an individuals' sprint velocity by varying the stiffness of sprint spikes remains a possibility. In order to investigate the potential biomechanical mechanisms through which improved performance may result, joint kinetic information is vital. Sources of power production within the lower limb, along with the energetic contribution of individual joints may provide insight into sprinting performance. Furthermore, pressure data can provide additional insight into the function of the foot during sprinting.

Stefanyshyn and Fusco (2004) speculated that changing the shoe bending stiffness would result in a change in the point of application of the ground reaction force, moving the centre of pressure anteriorly and thereby increasing the resultant joint moment and energy production at push-off, resulting in a more effective push-off. Whether such changes results in any measurable improvement in running velocity is unknown. However, stiffer running shoes have been shown to significantly reduce MPJ energy absorption and consequently improve performance in a vertical jump (Stefanyshyn & Nigg, 1998b).

Footwear modifications may elicit different biomechanical and performance responses in different athletes. Toon *et al.* (2009) found that individual differences existed in foot kinematics when introduced to the same footwear condition, suggesting that shoe selection is specific to the function requirements of individual requirements. Stefanyshyn and Fusco (2004) agreed that the optimal equipment for each athlete is dependent upon their ankle plantarflexor strength, force-length and force-velocity characteristics. Kleindenst *et al.* (2005) demonstrated differences in responses to running shoes of varying stiffness coincided with runner's preferences and subjective perception. It has therefore been suggested that the only way to detect differences that exist between shoes is to employ a single-subject analysis (Bates, 1996). Single subject analyses have been advocated in the research when a task could potentially elicit different responses / strategies in individuals and when the group mean does not describe well any individual subject response (Revill *et al.*, 2008).

6.2 Objectives

The research question to be addressed in this study is:

How does the stiffness of sprint spikes affect the biomechanical characteristics and function of the MPJ for individual athletes?

It was hypothesised that increases in sprint spike stiffness would result in improved individual sprinting velocities for the two participants in the study (Hypothesis 1). This was based upon the previous results in Chapter 3, where the majority of participants demonstrated improved sprint times with an increase in sprint spike sole stiffness. However, it was also thought that stiffer sprint spike conditions would result in an increased MPJ moment and energy production during push-off (Hypothesis 2).

Specifically the objectives of the study were:

1) *To investigate how an individual responds to different sprint spike stiffness conditions using a single subject approach.* This objective will address hypotheses 1 and 2.
2) *To provide some insight into inter-individual differences in the two athlete's responses to the footwear conditions.*

6.3 Methods

6.3.1 Participants

One female and one male athlete participated in the study. The female participant (participant one) was aged 28 years with a height of 179 cm and body mass of 68 kg. She was a trained sprinter who specialised in sprint hurdles (100 m hurdles personal best 14.12 s). The male participant (participant two) was aged 28 with a height of 182 cm, body mass of 73 kg and a 400 / 800 m runner with a 400 m personal best of 50.8 s. Both participants were regularly training at the time of testing. Informed written consent was obtained from all subjects and ethical approval was obtained, as per Chapter 3.

6.3.2 Protocol

Each participant visited the laboratory on three occasions. Three sprint spike conditions with increasing stiffness (C1, C3, and C4 – conditions corresponding with Chapter 5) were compared to a baseline sprint spike on each testing session (B1, B2, B3: for testing session 1, 2 and 3). The order of the three sprint spike conditions, over the three testing days, was randomized (Figure 6.1).

Session 1												
Sprint Trial number	1	2	3	4	5	6	7	8	9	10	11	12
Condition	B1	C4	B1	C4	B1	C4	B1	C4	B1	C4	B1	C4
Session 2												
Sprint Trial number	1	2	3	4	5	6	7	8	9	10	11	12
Condition	B2	C1	B2	C1	B2	C1	B2	C1	B2	C1	B2	C1
Session 3												
Sprint Trial number	1	2	3	4	5	6	7	8	9	10	11	12
Condition	B3	C3	B3	C3	B3	C3	B3	C3	B3	C3	B3	C3

Figure 6.1 Example Testing protocol for three data collection sessions, Conditions: B = baseline, C = Stiffness condition 1/2/4 (increasing stiffness).

The three sprint spike conditions appeared identical; these were Puma sprint spikes with the same outers, but with glass / carbon fibre insoles of differing stiffness's. The mechanical stiffnesses of the sprint spike conditions had been previously determined in Study 3 using a 2 point bending test, and are shown in table 6.1 (conditions 1, 3 and 4 from study 3). The Asics hypersprint spike was chosen from a selection of commercially available shoes, which were also mechanically tested (stiffness ranged from 190.5 ± 5.3 to 256.1 ± 23.7 N.m for shoe size 43).

Table 6.1 Measured stiffness of four conditions used in the study, average of left and right shoes. Asics sprint spike was selected as the baseline measure. % values represent normalised stiffness to the baseline measure.

	Shoe size 38.5 Average Mechanical stiffness (N/m) of left and right shoe and Normalised stiffness to baseline (%)	Shoe size 43 Average Mechanical stiffness (N/m) of left and right shoe and Normalised stiffness to baseline (%)
Asics sprint spike (baseline - B)	183.66 ± 6.25 (100%)	197.9 ± 29.6 (100%)
Puma sprint spike condition C1 (no insole)	254.6 ± 22.7 (139%)	297.4 ± 7.6 (150%)
Puma Insole condition C3	367.0 ± 19.7 (200%)	408.6 10.5 (207%)
Puma Insole condition C4	501.7 ± 43.6 (273%)	472.1 ± 31.6 (239%)

In each testing session, twelve maximal sprinting trials were collected, six in the baseline condition and six in one of the Puma stiffness condition. This number of trials was selected based upon pilot testing (see Appendix E). After every trial the participant swapped shoes. The sprints were performed on a 70 m indoor runway with an indoor synthetic track surface and the athletes accelerated for approximately 35 m before data collection and were likely still accelerating during data collection.

As detailed in Chapter 4, kinematic and kinetic data were collected at 1000 Hz and pressure data were also captured at 250 Hz. The joint marker set and model was identical to that documented in Chapter 4. The position of the markers was marked with a marker pen onto the foot, so that when the sprint spikes were swapped, the markers remained in the same position on the sprint spikes. Furthermore, when the sprint spikes were swapped, a new standing calibration and pointer trial was performed. Data were also processed as previously explained in Chapter 3 with MPJ kinematics and kinetics accordingly calculated.

After each sprint spike trial, participants completed a perception assessment. At the end of the testing session, they also completed further questions to determine their perceptions of the stiffness and performance of the different footwear conditions (see Appendix F).

6.3.3 Pressure analysis

To analyse the pressure data the foot was divided into six anatomical sub-regions using the RSScan gait software, version 7.97. The six regions were the hallux, the lesser toes (toes 2-5), and the forefoot was split into four areas due to the location of the spikes under the shoes. These four areas were easier to repeatedly identify, and therefore more accurate than if the five MTHs were to be estimated. This was due to the location of the spikes on the sole of the shoes which made it impossible to identify exactly where the underlying metatarsal heads were located. The plantar pressure variables that were obtained for each subject were peak pressure, loading rate, impulse and time to peak pressure (% of stance phase) for each region.

6.3.4 Statistical analysis

Each participant achieved six successful trials in each condition, all of which were used for statistical analysis. Kinematic, Kinetic and pressure data was analysed for the assessment of shoe differences using a within-subject statistical technique. The Model Statistic (Bates *et al.*, 2004; Dufek & Bates, 1991) compares observed condition differences to a critical value, to determine statistical significance. Mean absolute differences between conditions are compared with the critical difference, which is the product of the estimated population standard deviation and a critical value based upon sample (trial) size. The critical value used in this study was 1.2408 for six trial comparisons. For both participants, only four successful pressure trials were analysed, due to some problems with the RSScan gait software automatically recognising forefoot contacts, therefore for the pressure data a critical value of 1.5056 was used for four trial comparisons. The Model Statistic was conducted at $\alpha = 0.05$ level of significance. The interpretation of a mean absolute difference greater than the critical difference is that the difference is due to sampling error at a probability of α, therefore the null hypothesis is rejected.

6.4 Results

6.4.1 Participant one:

The female participant achieved highest mean sprinting velocities in the two stiffest conditions, C3 and C4 (8.20 ± 0.08 m/s and 8.20 ± 0.09 m/s), however the differences between footwear conditions were small and the only significant difference, to the corresponding baseline conditions, was found for the stiffest condition C4, which resulted in just a 2% increase in measured velocity. There were no significant differences in contact time between conditions. Table 6.2 demonstrates key MPJ kinematic and kinetic results for all conditions. Range of motion about the MPJ was reduced by conditions C3 and C4, in comparison to baseline, although again only condition C4 was significantly different to baseline, with this, the stiffest condition, reducing the range of motion about the joint by approximately 5.5 degrees during stance. Conversely, range of motion was significantly greater in condition C1 than the corresponding baseline. There were no significant reductions in peak MPJ plantarflexion angular velocities during stance, but all three stiffness conditions significantly reduced peak MPJ dorsiflexion velocities which occurred shortly after take-off.

Table 6.2. Participant one: Mean (n=6) sprinting velocity and selected MPJ kinematic and kinetic variables for all three stiffness conditions (C1, C3, C4) and corresponding baseline conditions (B1, B2, B3). * denotes a significant difference to the corresponding baseline measure ($p<0.05$).

	B1	C1	B2	C3	B3	C4
Speed (m/s)	8.15 (± 0.23)	8.14 (± 0.14)	8.07 (± 0.18)	8.20 (± 0.08)	8.04 (± 0.14)	8.20* (± 0.09)
MPJ Range of Motion (°)	37.9 (± 3.8)	42.3* (± 3.1)	41.1 (± 2.5)	39.2 (± 1.2)	46.8 (± 2.0)	41.3* (3.0)
Peak MPJ Dorsiflexion angular velocity (°/s)	1182.7 (± 314.4)	1228.3 (± 156.7)	1128.2 (± 158.1)	1068.5 (±184.8)	1334.2 (± 261.3)	1197.8 (± 66.0)
Peak MPJ Plantarflexion angular velocity (°/s)	-3071.0 (± 143.1)	-2729.3* (± 209.7)	-3472.0 (± 272.9)	-3008.5* (± 280.1)	-3080.8 (± 48.7)	-2476.5* (± 170.2)
Peak Fz (N)	2057.8 (± 36.3)	2144.0* (±36.7)	2083.2 (±39.1)	2136.1* (±36.6)	2037.2 (±34.8)	2078.1* (± 29.3)
Peak MPJ plantar flexor moment (N.m)	53.7 (± 5.6)	64.0* (± 5.9)	56.0 (± 4.4)	72.7* (± 7.8)	58.7 (± 9.1)	67.6* (± 11.5)
Peak Positive Power (W) generated during MPJ plantarflexion	201.2 (± 100.0)	292.3 (± 73.5)	183.7 (± 81.0)	253.3 (±33.9)	333.9 (±148.9)	266.5 (±57.7)
Peak Negative Power (W) generated during MPJ dorsiflexion	-691.7 (± 61.2)	-867.0 (± 55.0)	-777..4 (± 90.9)	-855.1 (±90.6)	-1000.2 (± 169.2)	-927.9 (±50.0)
Total Energy generated (J) during MPJ plantarflexion	1.4 (± 1.1)	0.5* (± 0.2)	1.3 (± 1.0)	0.9 (± 0.5)	2.0 (± 1.5)	1.3 (± 0.9)
Total Energy absorbed (J) during MPJ dorsiflexion	-30.6 (± 2.3)	-33.3 (± 2.0)	-31.2 (± 2.8)	-36.4* (± 3.9)	-34.1 (± 4.3)	-36.9 (± 6.6)
Total energy generated (J) during push-off	0.3 (± 0.1)	2.3* (± 0.6)	1.6 (± 0.7)	2.1 (± 0.4)	0.9 (± 0.4)	2.1* (± 0.5)

Peak vertical ground reaction forces were higher in all the Puma stiffness conditions (C1, C3, C4) than the corresponding baseline conditions. Similarly, the resultant joint moment about the MPJ was also significantly higher in all three Puma stiffness conditions, in comparison to the corresponding baseline condition. Figure 6.2 demonstrates the mean resultant MPJ moment for baseline measurements B3 and stiffness condition C4, whereby the peak resultant joint moment was increased by approximately 8.9 N m (a 15 % increase).

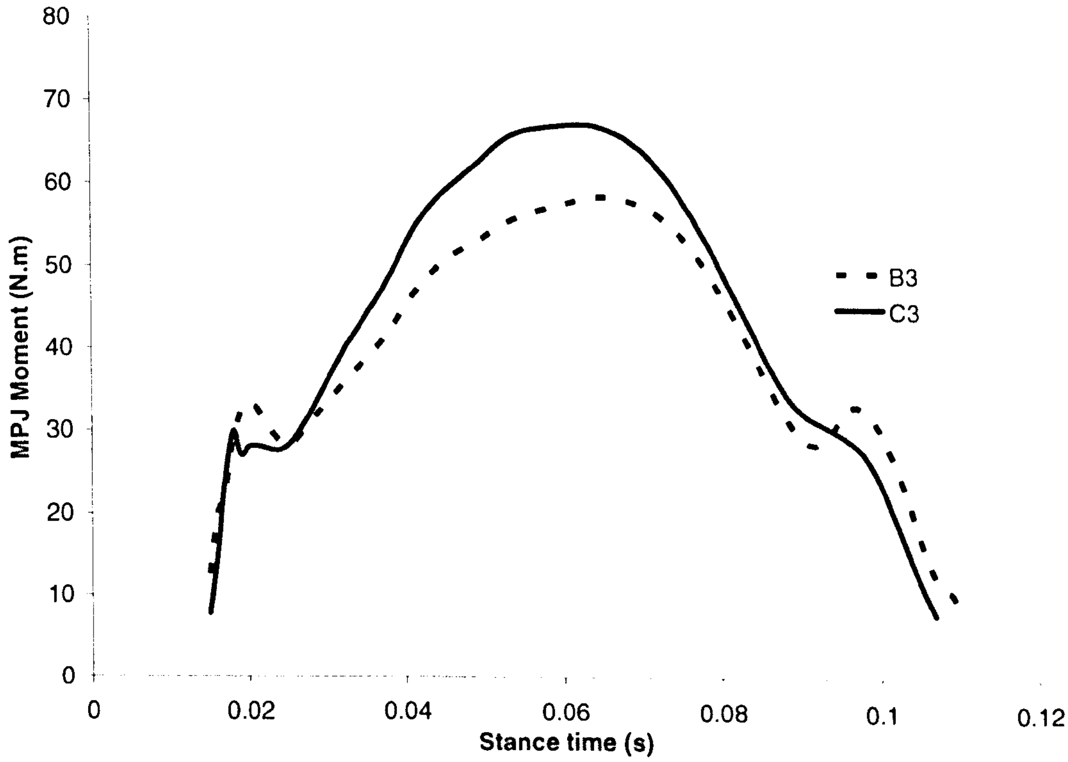


Figure 6.2 Mean resultant MPJ moment (n = 6) for participant one, conditions baseline 3 and Puma stiff insole condition 4. MPJ moment was plantarflexor throughout stance, peak moment was significantly ($p < 0.05$) different between the two conditions.

There were no significant differences in peak negative or positive powers exhibited at the MPJ during stance. A net loss of energy was evident at the joint in all baseline and stiffness conditions, with total net energy between -28.3 J and -33.5 J, minimal energy production and little difference between shoe conditions. Between 30.6 and 36.9 J of energy was lost during the energy absorption phase (Figure 6.3). The effect of shoe stiffness on this energy loss was small, and although the three stiff conditions all increased the amount of this energy loss at the MPJ during dorsiflexion of the joint during stance, the difference to baseline was only significant in the C3 stiffness condition. All three stiffness conditions resulted in increased positive energy created during the push-off phase at the end of stance, this was significantly greater to baseline in condition C1 (producing 2.3 J energy) and condition C4 (producing 2.1 J energy).

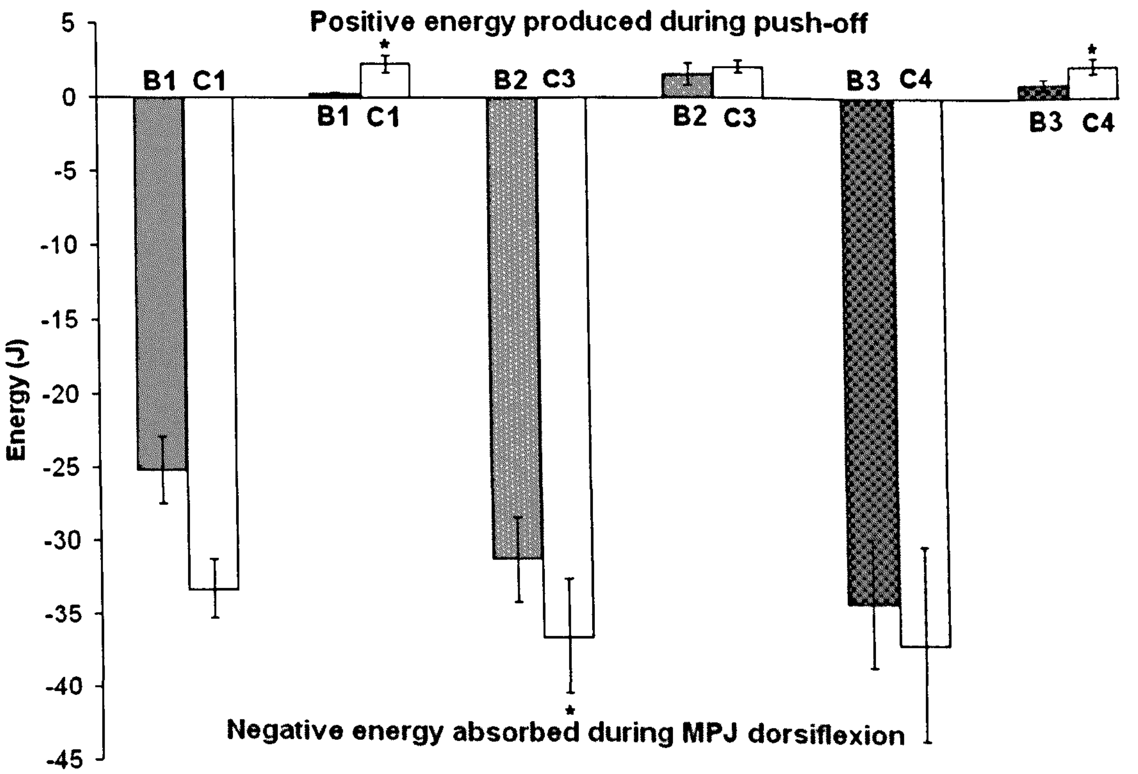


Figure 6.3 Mean (n=6) energy absorbed during MPJ dorsiflexion and produced during push-off for participant one, for three shoe stiffness conditions (C1, C3, C4) and corresponding baseline conditions (B1, B2, B3). * denotes a significant difference in energy to the corresponding baseline measure ($p<0.05$).

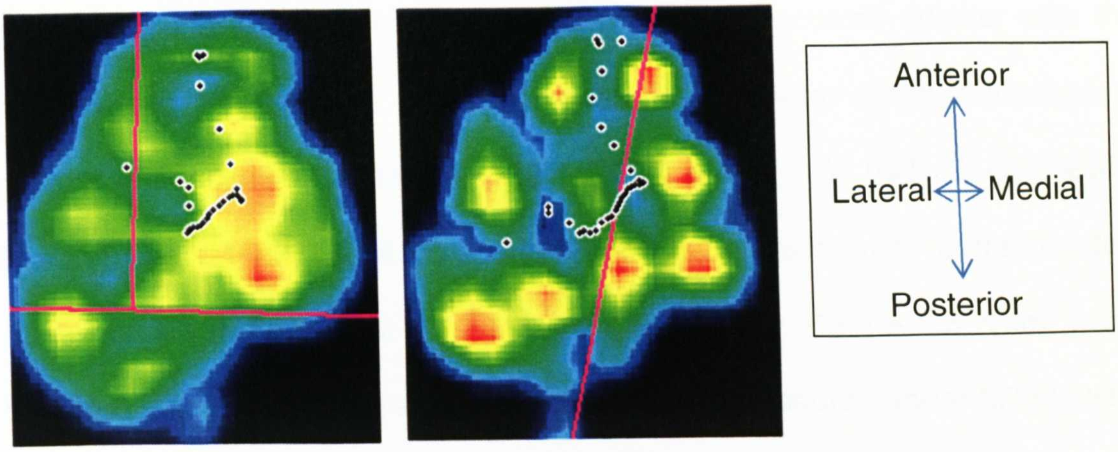


Figure 6.4 Typical pressure profiles for baseline condition (left) and stiffest sprint spike condition (right), participant one.

In all the sprint trials, the pressure pattern under the shoe was similar, with the centre of pressure quickly shifting from the lateral to medial forefoot followed by anterior progression to the end of the toe box (Figure 6.4). For this athlete, contact with the ground was only made with the forefoot and the centroid of pressure was anterior to the metatarsal heads throughout the stance phase. Pressure patterns were consistent between trials but there were few statistically significant differences between the stiffness and baseline conditions. In the three stiffness conditions, C1, C3, and C4, the maximum pressures on the hallux and lesser toes were greater than baseline, although not significant. For example, in condition C4 the maximum pressure on the hallux was 48.8 ± 27.1 $N{\cdot}cm^2$ and lesser toes 43.58 ± 16.3 $N{\cdot}cm^2$, compared to baseline values of 33.7 ± 28.7 $N{\cdot}cm^2$ and 31.8 ± 18.6 $N{\cdot}cm^2$ respectively. The time difference between peak pressures on MTH1 and MTH5 was lowest in condition C4 (44% stance time), possibly providing evidence to suggest that the shift from lateral to medial loading occurred quicker in the stiffer condition. Peak pressure on the lesser toes (toes 2-5) occurred significantly earlier in the Puma spikes condition C4, in comparison to baseline conditions B3, suggesting the forward progression onto the toes occurred quicker with the stiffer shoes. Furthermore, peak pressures on the centre of the forefoot were greater in conditions C1 (MTH2: 132.1 ± 22.4 $N{\cdot}cm^2$), C3 (MTH2: 86.5 ±34.6 $N{\cdot}cm^2$) and C4 (MTH2: 89.6 ± 7.1 $N{\cdot}cm^2$) in comparison to corresponding baselines B1 (MTH2: 47.2 ± 19.2 $N{\cdot}cm^2$), B2 (MTH2: 62.3 ± 3.3 $N{\cdot}cm^2$), and B3 (46.6 ± 4.6 $N{\cdot}cm^{2)}$. These differences in peak pressures under MTH2 were significant and significant increases were evident for MTH3 peak pressures too.

From the subjective questionnaires, the participant preferred every stiffness condition over the baseline condition and perceived that their performance was improved in the Puma sprint spikes, by an increase of between 3% and 8%. The athlete felt that they performed best in the C1 condition. The participant also identified that the puma sprint spikes conditions felt stiffer, which forced the athlete 'to run more up on my toes', in comparison to the Asics baseline shoe which 'felt too flexible, flimsy and not supportive enough'. However, the athlete was unable to correctly identify stiffness increases between the different Puma stiff conditions C1, C3 and C4 while sprinting.

6.4.2 Participant two

The male participant achieved highest mean sprinting velocities in stiffness condition C3 (8.12 ± 0.27 m/s) however there were no significant differences in velocities between any of the stiffness conditions and the baseline condition. The difference in mean sprint velocity measured for condition C3, in comparison to the corresponding baseline was less than 1%. In the other two testing sessions, the participant exhibited higher velocities in the baseline conditions B1 and B3 than the stiffness conditions C1 and C4.

This participant demonstrated larger variability in sprint performance, with higher standard deviations within sessions and larger differences in baseline velocities measured, between sessions (Table 6.3). For example, in the testing session for conditions B1 and C1, the athlete was also not able to replicate the sprinting velocities of the other two sessions, and mean resultant velocities were 0.3 – 0.5 m/s slower.

There were no significant differences in contact time between conditions. Table 6.3 demonstrates key MPJ kinematic and kinetic results for all conditions. Range of motion about the MPJ was significantly reduced by conditions C1, C3 and C4, in comparison to baseline, with condition C3 demonstrating the greatest reduction in the range of motion about the joint, by approximately 9.3 degrees during stance. The angular displacement undergone at the MPJ during stance, for one typical trial in both the baseline condition and stiffness condition C3 is shown in Figure 6.5, which clearly demonstrates both the peak MPJ flexion and rate of flexion were reduced. Peak MPJ dorsiflexion angular velocities during stance were significantly reduced in all three stiffness conditions, again condition C3 resulting in the largest reductions of approximately 31%. Peak MPJ plantarflexion velocity occurred after toe-off and was also significantly diminished by conditions C1 and C3 (with the largest reduction of 14%), although the decrease for condition C4 was not significant.

Table 6.3. Participant two: Mean (n=6) sprinting velocity and selected MPJ kinematic and kinetic variables for all three stiffness conditions (C1, C3, C4) and corresponding baseline conditions (B1, B2, B3). * denotes a significant difference to the corresponding baseline measure ($p<0.05$).

	B1	C1	B2	C3	B3	C4
Speed (m/s)	7.66 (± 0.22)	7.60 (± 0.15)	8.05 (± 0.27)	8.12 (± 0.27)	7.94 (± 0.21)	7.90 (± 0.24)
MPJ Range of Motion (°)	37.1 (± 1.4)	31.6* (± 1.5)	37.8 (± 1.2)	28.5* (± 2.0)	36.5 (± 1.6)	30.3* (2.5)
Peak MPJ Dorsiflexion angular velocity (°/s)	844.2 (± 49.1)	703.3* (± 84.2)	773.6 (± 74.3)	534.4* (± 46.4)	783.8 (± 44.8)	697.2* (±68.5)
Peak MPJ Plantarflexion angular velocity (°/s)	-2359.0 (± 95.0)	-2146.8* (± 107.0)	-2507.4 (± 175.4)	-2150.2* (± 187.9)	-2288.0 (± 392.7)	-2120.2* (±331.5)
Peak Fz (N)	2261.9 (±60.0)	2194.5* (±79.5)	2271.7 (± 36.9)	2345.6* (±72.1)	2253.2 (±98.6)	2324.9 (± 89.1)
Peak MPJ plantar flexor moment (N m)	83.1 (± 2.4)	87.6* (± 2.9)	72.1 (± 5.0)	79.7* (± 3.3)	75.8 (± 4.6)	87.6* (± 6.6)
Peak Positive Power (W) generated during MPJ plantarflexion	220.3 (± 23.9)	292.3 (± 73.5)	241.9 (± 24.2)	241.8 (±22.0)	265.5 (±52.3)	386.5* (±128.5)
Peak Negative Power (W) generated during MPJ dorsiflexion	-1053.7 (± 78.1)	-976.2* (± 97.6)	-963.8 (± 102.4)	-678.2* (±88.2)	-927.2 (± 78.2)	-916.2 (±121.0)
Total Energy generated (J) during MPJ plantarflexion	0.6 (± 0.4)	0.4 (± 0.2)	0.2 (± 0.2)	0.2 (± 0.2)	0.5 (±0.2)	0.8 (± 0.6)
Total Energy absorbed (J) during MPJ dorsiflexion	-42.0 (± 2.3)	-35.8* (± 3.0)	-35.0 (± 4.1)	-26.3* (± 1.3)	-37.8 (± 2.0)	-34.2* (± 3.3)
Total energy generated (J) during push-off	2.2 (± 0.3)	3.3* (± 0.6)	2.4 (± 0.4)	2.7 (± 0.4)	2.8 (± 0.7)	3.2 (± 0.6)

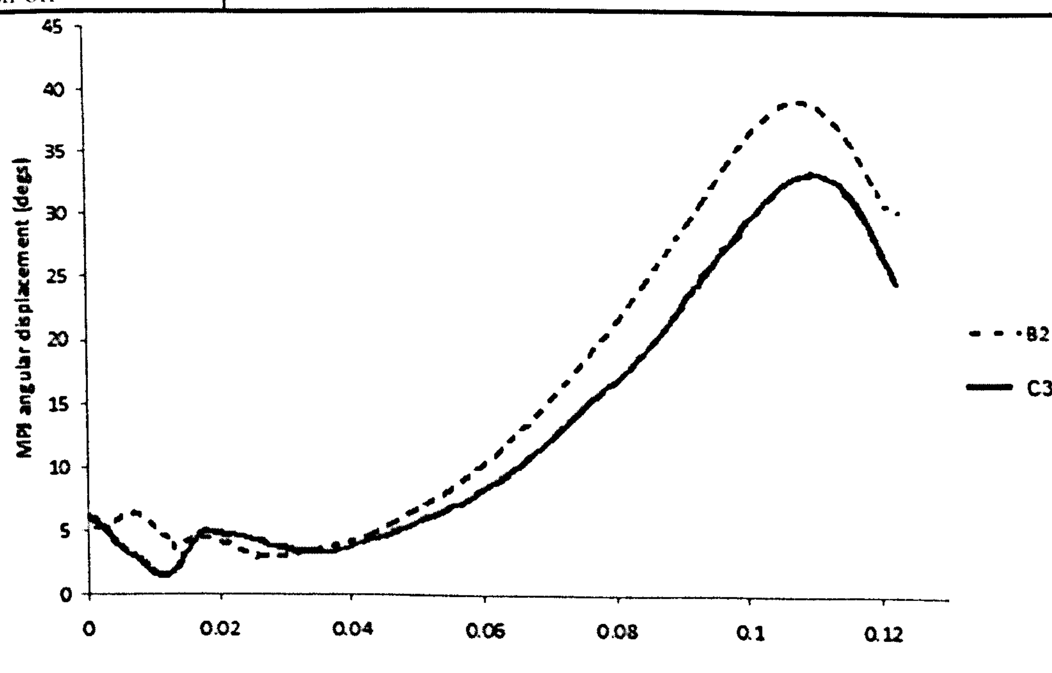


Figure 6.5 MPJ angular displacement during stance for one typical trial, conditions baseline 2 (B2) and Puma stiff insole condition 2 (C3), for participant two. MPJ range of motion was significantly reduced in C3.

Peak vertical ground reaction forces were significantly higher in condition C3 than baseline B2 (table 6.3). The resultant joint moment about the MPJ was significantly higher in all three Puma stiffness conditions in comparison to the corresponding baseline conditions. Figure 6.6 demonstrates the mean resultant MPJ moment for baseline measurements B3 and stiffness condition C4, whereby the peak resultant joint moment was increased the greatest, by approximately 11.8 N·m.

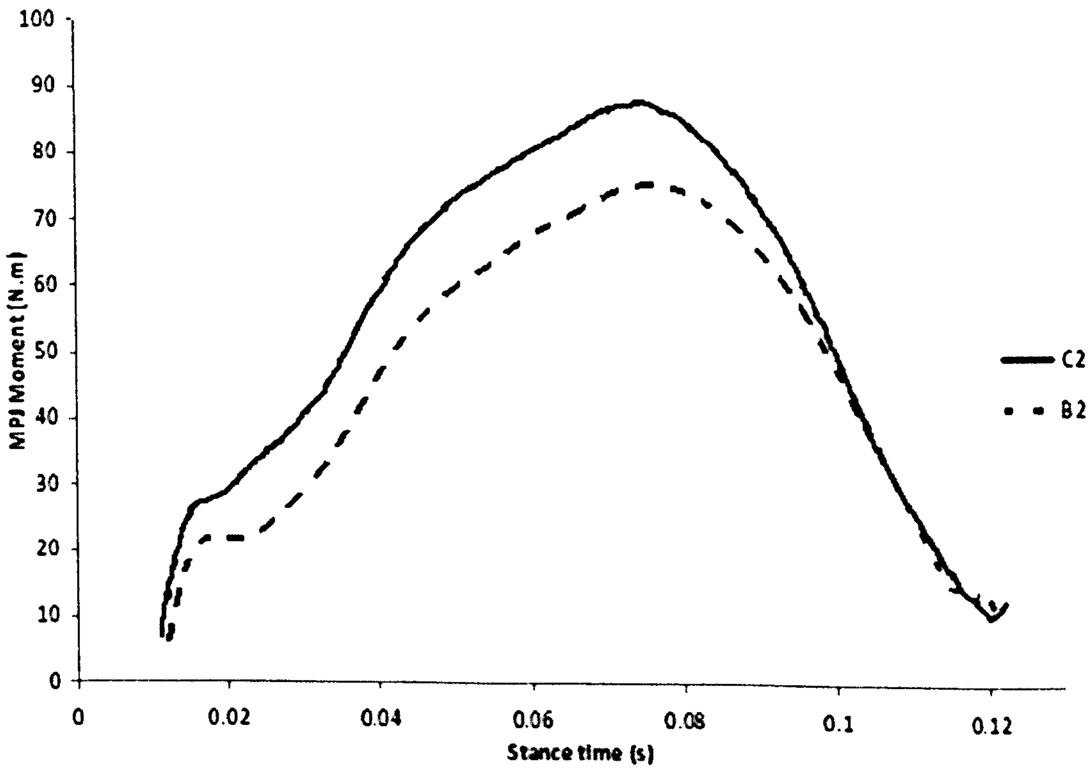


Figure 6.6 Mean resultant MPJ moment (n = 6) for participant two, conditions baseline 3 and Puma stiff insole condition 4. MPJ moment was plantarflexor throughout stance, peak moment was significantly ($p < 0.05$) different between the two conditions.

The MPJ was a large absorber of energy during stance with up to 42.0 J lost during the energy absorption phase in the baseline condition (Figure 6.7). All three shoe stiffness conditions C1, C2 and C3 significantly reduced the amount of energy lost, compared to the baseline measures. The largest reduction was in condition C3 with 8.2 J (23%) difference in energy loss compared to the corresponding baseline. All three stiffness conditions, also resulted in increased positive energy created during the push-off phase at the end of stance, this was only significantly greater to baseline in condition C1 (producing 3.3 J energy, compared to 2.2 J energy in B1).

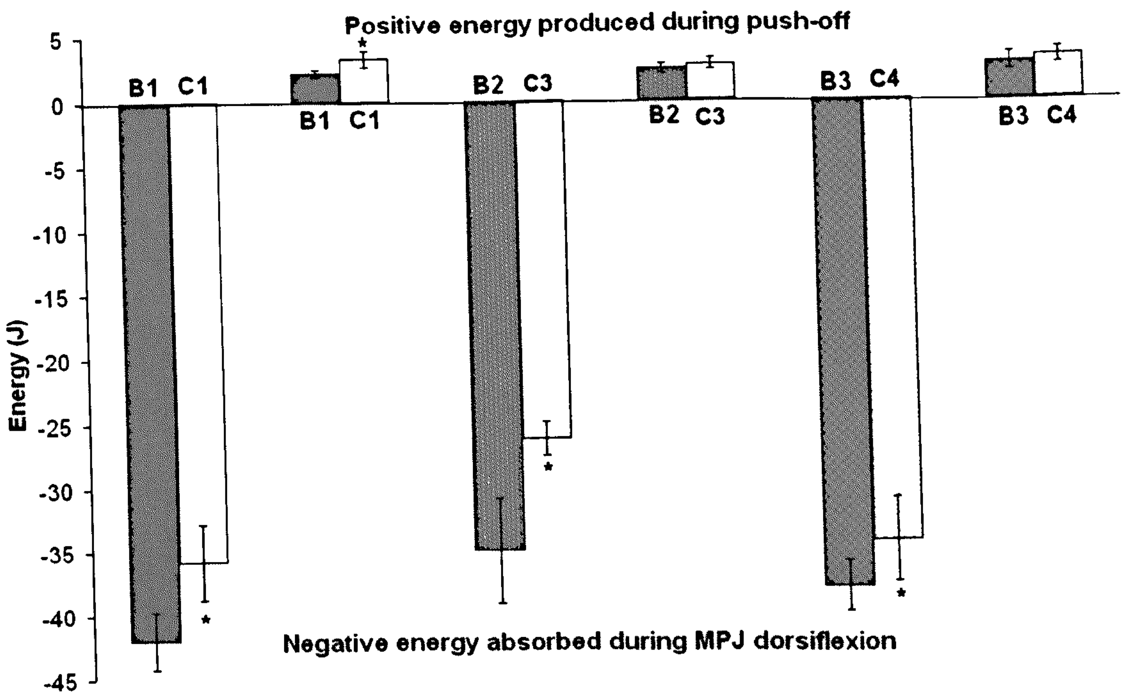


Figure 6.7 Mean (n=6) energy absorbed during MPJ dorsiflexion and produced during push-off for participant two, for three shoe stiffness conditions (C2, C3, C4) and corresponding baseline conditions (B1, B2, B3). * denotes a significant difference in energy to the corresponding baseline measure ($p<0.05$).

The pressure distribution pattern under the shoe was very similar to participant 1, with the centre of pressure quickly shifting from the lateral forefoot to medial followed by anterior progression to the end of the toe box. For this athlete, pressure patterns were less consistent between trials and in a couple of trials some midfoot contact was made. Figure 6.8 demonstrates two typical pressure patterns, when contact was made with the forefoot, with the centre of pressure demonstrating a clear lateral to medial transition. In the three stiffness conditions, C1, C3, and C4, the maximum pressures on the hallux were higher, although not significant. Peak pressures on the centre of the forefoot were greater in conditions C1, C3 and C4 in comparison to baselines, with significantly higher peak pressures under MTH2 and MTH3. The time to peak pressures on the hallux and medial and central metatarsal heads MTH1,2 and 3 were significantly reduced in the conditions C1, C3 and C4, in comparison to the baseline shoe. For example, in condition 2 peak pressures on MTH1 occurred at 37% stance and on the hallux at 83% stance, compared to 51% and 90% stance for condition B3. This could indicate a quicker shift to the centre of the spike plate and a faster forward progression to the anterior edge of the toe box.

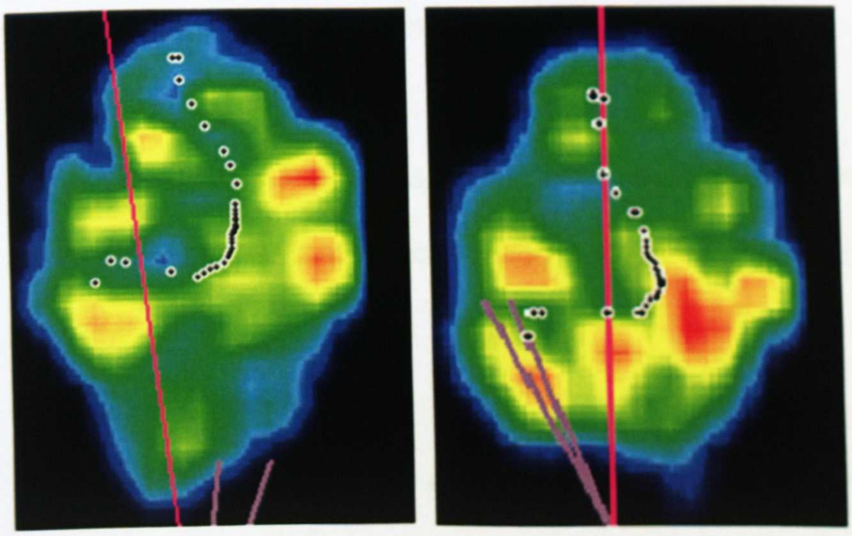


Figure 6.8 Typical pressure profiles for baseline condition B3 (left) and stiffest sprint spike condition C4 (right), participant two

Participant two rated the Puma sprint spike conditions higher for performance than the baseline shoe, with mean performance ratings of 106%, 125% and 150%, relative to the baseline (100%) for condition C1, C3 and C4 respectively. The participant commented that these conditions 'felt faster', 'were easier to sprint in' and 'required less effort to run on your toes and develop speed'. The participant also identified that the Puma sprint spike conditions were stiffer than the baseline shoe, although ranked the stiffnesses in the following order: C1 (lowest stiffness), C4, C3.

6.5 Discussion

This study addressed the question of *How does the stiffness of sprint spikes affect the biomechanical characteristics and function of the MPJ for individual athletes?* A single subject design was utilized to explore the effect of sprint spikes with four different stiffness's on the kinematics and energetics of the MPJ during sprinting and on the pressure distribution under the soleplate of the forefoot for individual subjects.

6.5.1 Objective 1) To investigate how an individual responds to different sprint spike stiffness conditions using a single subject approach.

6.5.2 Sprinting performance

Hypothesis 1 was accepted as participant one demonstrated highest sprinting velocities in stiffness conditions C3 and C4 (C4 significantly higher than baseline), participant two demonstrated highest sprinting velocity in stiffness condition C3. The velocity increases associated with the participant's best shoe, were however minimal, with just a 2% improvement for participant one and 1% improvement for participant two. Stefanyshyn and Fusco (2004) reported mean sprint time improvements of 1.2% across thirty-four subjects, when comparing their best plate condition to the standard shoes, and 0.7% between the standard shoes and stiff plates (42 $N.mm^{-1}$). The authors deemed this improvement worthwhile for an athlete at an elite level. For the two sprinters who took part in this study, it is, however, questionable whether this degree of enhancement is useful, considering the higher degree of variability between sprint trials and lower levels of performance. Participant one demonstrated low variability and achieved the highest velocity in the stiffest condition, which was approximately 2.5 times stiffer than baseline. It is not possible to suggest an optimal shoe for sprinting velocity, based on just four measured stiffness conditions, nor is it known whether increasing the sole stiffness even further may have improved sprint velocity. Stefanyshyn and Fusco (2004) did not measure the stiffness of the athlete's shoes, as athletes wore their own sprint spikes, nor did they approximate any measure of stiffness for the sprint spikes plus their insole conditions, therefore it is not possible to directly compare the stiffnesses of the sprint spike conditions.

6.5.3 MPJ kinematics

For both participants, the stiff sprint spikes did appear to have a controlling effect over the MPJ kinematics, significantly reducing the range of motion and also reducing the extent and rate of MPJ dorsiflexion during stance and plantarflexion after push-off, agreeing with Smith & Lake (2010) and Toon *et al.* (2009). However, both athletes demonstrated MPJ plantarflexion during push off, suggesting that the athletes were able to overcome the stiffness of the plate to enable some push off with the toes, as has been shown in previous studies (Smith & Lake, 2009; 2010). These results clearly demonstrate that varying the sole stiffness effects sagittal plane kinematics of the lower leg during the stance phase of sprinting. The reduction of MPJ motion during midstance, by the stiffer shoes, may influence the effectiveness of the Windlass mechanism (Hicks, 1954). Dorsiflexion of the MPJ during stance causes progressive tightening of the plantar aponeurosis, the plantar fascia shortens and pulls the calcaneus and metatarsal heads together, elevating the longitudinal arch such that the foot becomes a stable lever system for propulsion (Bolgla & Malone, 2004; Dugan & Bhat, 2005; Fuller, 2000). In stiff sprint spikes, MPJ dorsiflexion is reduced and therefore the functionality of the Windlass mechanism and hence the efficiency of the natural lever system in the foot may be compromised (Boggle & Malone, 2004). However, the high bending stiffness of the sprint spikes may allow the athletes to push off but still achieve substantial rigidity from the foot and shoe as a system.

Participant one exhibited higher ranges of motion than participant two, indicating differences between the participants in their natural range of motion and flexibility within the foot and differing inherent foot stiffness or foot structures (e.g. high / low arch, plantar fascia stiffness). Participant two experienced larger reductions in angular motion and peak angular velocities with the stiffer sprint spikes, suggesting greater responses to shoe stiffness, which resulted in higher compromised angular motion at the MPJ.

6.5.4 MPJ kinetics

Moments around the MPJ were higher for the male participant (72.1 – 87.6 N·m) than the female participant (53.7 – 72.7 N·m), although similar to the range of 40 – 80 N·m reported by Stefanyshyn & Nigg (1997) for sprinting (own sprint spikes). For both participants there was a significant increase in resultant moment across all stiffness conditions. Furthermore the stiffer the sprint spike condition, the greater the difference in joint moment to the baseline condition. It is speculated that the resultant moment may also increase at the ankle, resulting in enhanced performance. Although, this may only be realised if the athlete has sufficient ankle plantar flexor strength combined with an appropriate rate of force generation, and therefore it is likely to be highly athlete specific.

Stefanyshyn and Nigg (2000) reported that running shoes with stiff midsoles reduced the amount of energy loss at the MPJ during both running (average reduction 8-10 J) and jumping (average reduction 5.4 J), primarily due to a reduction in dorsiflexion around the joint. However, the participants in this study demonstrated different responses to the shoes. For participant one, the stiffer shoes actually increased the amount of energy loss around the joint (significantly only for condition C2 with a 5.2 J increase). On close inspection of the contributing factors, this increase of energy loss was a result of a larger area under the joint power curve, primarily due to increases in the MPJ moment. Although the angular velocity of the MPJ was reduced with the stiff condition, the effect of the increased joint moment was greater, therefore resulting in higher joint powers and energies. For participant two, all stiffer shoes resulted in a significant reduction of energy loss (up to 8.2 J reduction). This was primarily due to a substantial reduction in the MPJ angular velocity, which resulted in lower negative joint powers and resultant negative energy loss.

This study again demonstrated that a small amount of energy production during push off does occur, as the MPJ plantarflexes prior to take-off, in contrast with the findings of Stefanyshyn and Nigg (1997, 1998b, 2000). The stiffer sprint spikes did not compromise angular range during push off for the two participants in this study, nor did they compromise the athletes' ability to generate any energy. In fact the stiff sprint spikes increased the lever length of the foot during push off, thereby facilitating an effective propulsive system. Both participants in this study demonstrated that this energy production during the last 10 ms of stance was increased by a stiffer sprint shoe. However, these increases were small, and only significant in condition C1 for participant one and conditions C1 and C4 for participant two, therefore showing no systematic improvement with further increases of stiffness, rejecting Hypothesis 2.

6.5.5 Pressure

Fourchet *et al.* (2007) demonstrated that wearing sprint spikes significantly increased plantar loading under the midfoot and forefoot during sprint running. The heel pitch and toe spring angles, features of sprint spikes are designed to promote forefoot contacts and a rapid progression of the centre of pressure (Toon *et al.*, 2009). The results of this study suggest that stiffer sprint spikes further increase the loading applied on the central forefoot and may also facilitate a quicker lateral to medial followed by anterior progression of the centre of pressure to enable effective propulsion during push off, although due to variation in foot touchdown (particularly in participant two) few significant differences were found.

In this study loading was not measured at the foot – sole interface. Furthermore, the location of the spikes underneath the plate would have influenced the peak pressures recorded at the foot- ground interface. Therefore the pressure results must be treated with caution.

6.5.6 Objective 2) To provide some insight into inter-individual differences in the two athlete's responses to the footwear conditions.

This study clearly demonstrated that the two individuals responded differently to the four different footwear conditions, both in terms of sprint performance and foot biomechanics. Participant one demonstrated a significant improvement sprinting velocity in the stiffest condition, in comparison to respective baselines, even though the amount of energy absorbed at the MPJ was actually increased, therefore somewhat confounding the relationship between performance and mechanical energy. Whereas, for participant two, the stiff sprint spikes appeared to have a positive effect on the energetics of the MPJ, but there were no significant differences in sprint velocity, in comparison to respective baselines. Stefanyshyn and Nigg (2000) noted that the highest jump performance in their study was not necessarily the jump in which the participant had the maximal reduction in energy loss at the MPJ. They explained this by the fact that maximal performance is not simply the result of mechanical energy, but the result of a large variety of factors, that could be physical, physiological or psychological. There are other important biomechanical factors to consider in this study, such as the athlete's plantarflexor strength and their personal force-length and force-velocity characteristics, as well as individual MPJ stiffness. Whilst Ding *et al.* (2011) found no systematic effect of MPJ stiffness on sprinting performance, Oleson *et al.* (2005) found that for most of the bending range, forefoot stiffness is much larger than the bending stiffness of the shoes. Although they tested running shoes and not sprint spikes, they argued that the total forefoot stiffness is more dominated by the stiffness of the human foot than the bending stiffness of footwear. This adds support to dismissing the notion of minimising energy loss by increasing shoe stiffness. However, the smaller forefoot stiffness reported by Oleson *et al.* (2005) near toe-off, suggests that shoe bending stiffness may be of some importance in toe-off.

Different responses to the footwear conditions may be due to anatomical differences and neurological responses (performance strategies). A participant's response will depend on their recognition / perception of the perturbation, which in turn will be a function of the individual's past experiences (Bates, 1996). Response patterns vary along a continuum (from a Newtonian response whereby the perturbation is completely ignored to a fully accommodating response). Bates (1996) stated individual performance strategies are likely to be the rule rather than the exception, due to the complexity of the human machine and its numerous associated functional degrees, along with the different experiences, perceptions and expectations of the participants. Examples of these strategy responses have been observed for impact forces during running and landing (Dufek & Bates, 1990; Nigg & Segesser, 1992; Stergiou, Bates, & James, 1999). For running, Lees and Bourcier (1994) provided evidence of a "movement pattern fixation" whereby various running patterns were exhibited, with athletes selecting one solution from a pool, which suit the particular requirements on that particular day of testing. The results from this study provide support for the notion of individual performance strategies, which if analysed in a group design, would result in increased intersubject variability and may also lead to false support of the null hypothesis (Bates *et al.,* 2004). By implementing a single subject design, this study allowed individual differences to be explored, which would often not be possible in group designs, whereby group statistics may demonstrate limited insight into the research question. Furthermore, the single subject analyses allowed for a more complete and meaningful assessment.

This study demonstrated that for the two participants investigated, there were shoe stiffness related changes in sprint velocity and MPJ function (ROM, moment and energy production). It is impossible to state an optimal stiffness condition for performance, based solely on the three different stiffness conditions plus baseline conditions used. Participant one demonstrated significantly improved sprinting velocity in the stiffest sprint shoe, participant two performed best in the second highest stiff shoe (although not significant). The current study added to the evidence that stiffer sprint spikes reduce the range and rate of motion at the MPJ; they also resulted in an increased MPJ moment and energy production at take-off for both participants. Whilst performance-related parameters, such as the energy lost and produced at the MPJ during stance were influenced by sprint spikes, the relationship between these parameters and sprinting performance are not clear, as the two individuals exhibited different and somewhat contradictory responses. The use of a repeated baseline measure was implemented for all testing sessions. In previous research, the repeated usage of a baseline condition between conditions has been deemed critical to the evaluation of treatment effects for biomechanical studies utilizing a single subject design (Stergiou & Scott, 2005).

6.6 Study Conclusion

In answer to the research question posed: *How does the stiffness of sprint spikes affect the biomechanical characteristics and function of the MPJ for individual athletes?* this study demonstrated that the stiffness properties of sprint spikes have a significant influence on the kinematics and energetics of the MPJ during ground contact of maximal sprinting for the two individuals investigated. Hypothesis 1 was accepted as both participants demonstrated improved sprinting velocity with a stiffer sprint spike, although the magnitude of this improvement was small. Hypothesis 2 was rejected, although joint moments were greater in the stiffer sprint spike condition for both participants, there was no systematic effect of sprint spike stiffness on energy production during take-off.

The individual responses exhibited in this study suggest that sprinting velocity may be improved by implementation of relevant shoe mechanical characteristics, as suggested by previous researchers (Stefanyshyn & Fusco, 2004). It is clear that shoe selection is specific to the functional requirements of individual athletes and athletes will respond differently to footwear modifications, likely based on their past experiences and expectations. Athletes and coaches are therefore recommended to experiment with shoes of different stiffness to obtain optimal sprint time performance.

CHAPTER 7: DISCUSSION

7.1 Introduction

The aim of this research was to evaluate how foot function may contribute to sprinting performance and the interaction between the mechanical properties of sprinting footwear and performance, with a focus on the role of the MPJ.

The research questions in Chapter 1 focussed the investigations undertaken throughout the thesis towards achieving this aim. The four research questions are therefore revisited in this chapter to outline how they were addressed by the series of investigations described in Chapters 3 to 6, and to summarise the key findings of this thesis. Following this, the appropriateness of specific aspects of the methodology used throughout the thesis will be discussed and potential future investigations will be proposed.

7.2 Addressing the Research Questions

The research question addressed in Chapter 3 was:

What influence do methodological issues have on the analysis of MPJ function during sprinting?

To answer this research question, kinematic and kinetic data for the foot were collected for a group of sprinters within the laboratory. A three segment foot model was developed, with forefoot, midfoot and rearfoot segments, the forefoot segment was based upon the modelling approach used by Oleson *et* *al* (2005). As the focus of the thesis was the investigation of the role of the MPJ during sprinting, the axis representation of the MPJ was analysed and different axes definitions were compared. The influence of two methodological issues (joint axes representations and the choice of cut off frequency for data processing) on the analysis of MPJ function during sprinting was ascertained.

In response to the research question, this study demonstrated that MPJ kinetic calculations are highly sensitive to errors in the modelling of the MPJ line and it was recommended to represent the MPJ axis as an oblique axis from MTH1 to MTH5. A perpendicular axis overestimated resultant joint moments by approximately 81% in comparison to an oblique axis. Secondly, it was concluded that both a high sample rate and low-pass filtering cut-off frequency for both kinematic and kinetic data, are vital for accurate assessments of high speed impacts and movements, as evident in sprinting. Using a cut of frequency of just 8 Hz severely underestimated the extent and rate of peak MPJ flexion during stance (by approximately 15° and 711°s).

The methods proposed in Chapter 3 provide a more detailed and realistic estimate of MPJ function during sprinting, in comparison to methodological approaches utilised by previous researchers (Stefanyshyn *et al.*, 1997; 1998b; 2000; Toon *et al.,* 2009).

The research question addressed in Chapter 4 was:

How do sprint spikes affect the behaviour of the foot in terms of MPJ function and pressure distribution?

This study reported typical foot function during sprinting when barefoot and wearing standardised sprint spikes. The large degree and rate of angular MPJ motion (typical MPJ range of motion was 42.3 ° and peak dorsiflexion velocity was 873.1 °/s in sprint spikes) and the resultant MPJ kinetics, such as the high joint moments (63.9 N.m) reported in this study, demonstrate the importance of including this joint in lower limb kinetic analyses of sprint running, agreeing with Bezodis *et al.* (In press), who demonstrated large errors in ankle, knee and hip moments (upto 40%) when ignoring this joint.

MPJ motion during stance was identified as follows. Immediately after touchdown the heel is lowered towards the floor and the MPJ plantarflexes. The MPJ then rapidly dorsiflexes during midstance, absorbing energy then finally plantarflexes during push-off, although continuing to plantarflex after the point of take-off. During push-off some, albeit, minimal power production occurs, however throughout stance the MPJ is primarily an energy absorber.

In response to the research question, there were significant and meaningful differences in MPJ kinematics and kinetics between the barefoot and shod conditions. MPJ range of motion and dorsiflexion velocities were significantly reduced wearing sprint spikes, indicative of a controlling effect of the sprint spikes. Whilst energy loss was not significantly reduced wearing sprint spikes, sprint spikes did result in a significant increase in the energy produced during push-off. Pressure results, such as greater anterior progression of the centre of pressure demonstrated in sprint spikes, also added evidence for more effective propulsion during push-off when wearing sprint spikes. Sprinting velocities were greater in the sprint spike conditions, possibly indicating that the positive effect of sprint spikes on the energetics of the MPJ during stance could have a positive influence upon sprinting performance.

The research question addressed in Chapter 5 was:

Can increasing sprint spike stiffness lead to significantly improved sprint performance?

There were two components to the study documented in Chapter 5: the fabrication and mechanical testing of known stiffness conditions, followed by performance testing on a group of sprinters. In response to the research question, there was little evidence for an increase in sprinting performance for this group of athletes, with no significant differences in sprinting velocities between four known sprint spike stiffness conditions. There was evidence for changes in sprint spike stiffness to elicit individual improvements to sprinting performance and small differences in MPJ kinematics, with stiffer sprint spike conditions eliciting controlling effects over normal MPJ function. Results suggest that an athlete's optimal sprint shoe stiffness is highly subject-specific and may be dependent upon their perceptions and previous experiences. The results do not provide specific information about the shoe selection strategy an individual should take to improve performance, MPJ kinematics alone are not sufficient to predict performance measures in sprinting and optimal shoe stiffness is specific to the functional requirements of an individual athlete.

The research question addressed in Chapter 6 was:

How does the stiffness of sprint spikes affect the biomechanical characteristics and function of the MPJ for individual athletes?

In order to address this research question, a single subject research design was employed, which assessed foot functional and behavioural responses to different stiffness conditions for two individual athletes. In response to the research question, there were clear individualised biomechanical responses (MPJ kinematics and kinetics and pressure distribution) to the different stiffness conditions. Participant one demonstrated improved sprinting performance in the two stiffest sprint shoes, participant two performed best in the second highest stiff shoe. This study suggests that individual sprinting performance may be improved by implementation of relevant shoe mechanical characteristics. As there was little evidence to suggest that the resultant energy loss at the MPJ is a key factor influencing sprinting performance, the role of the MPJ may not as dominant as that of the ankle, knee and hip joints. However, this study demonstrated that the function of the MPJ is affected by manipulations in shoe stiffness for two individual athletes.

Results from Chapters 4 and 6 demonstrated no systematic or clear differences in the amount of energy absorbed between footwear conditions. This suggests that the concept of reducing energy loss at the MPJ, by increasing the bending stiffness of sprint spikes, may not be viable. The notion that such a reduction in energy loss would elicit improvements in sprinting performance is not supported. Nigg (2010) argued that footwear manufacturer's attempts at enhanced energy return have been unsuccessful and Shorten (1993) argued that any potential contribution of energy return from the shoe would be minimal compared to energy storage and return in muscles and tendons (with estimations of energy storage and return of approximately 66 J for the knee extensor muscles, 32 J for the Achilles tendon and 17 J for the arch of the foot). However, results from this thesis have demonstrated some positive energy production by the MPJ during push-off. Whilst it is agreed this this energy production is minimal, and unlikely to have a meaningful effect on the overall energetics of the lower limb, Oleson *et al.* (2005) demonstrated that the stiffness of the human forefoot is dramatically reduced during push-off, therefore the stiffness of the sprint spike becomes more influential and important during this propulsive phase.

7.3 Methodological Approach

The detailed analyses and assessment of MPJ function throughout the thesis has provided a large amount of biomechanical data to further the scientific understanding of the specific role of the foot during sprint running. Whilst focussing only on the MPJ provided enough scope for a series of biomechanical studies, it is acknowledged that the role of the other joints (ankle, knee and hip) along with any compensatory effects of work performed at those joints were not determined. The collection of a combination of 3D video, force data, alongside pressure mat data within this research has necessitated the development and validation of data collection methods and experimental design to allow quantification of specific MPJ kinematic and kinetic data.

The 3D kinematic data were collected with the use of shoe and skin mounted joint markers assumed to reflect the motion of the underlying segments. Error associated with marker placement on the skin was minimized by marker placement precautions, in particular the use of the DEXA scanner. Errors associated with marker placement on the shoe and the motion of the shoe relative to the skin were minimized by placing joint markers through the holes of the sprint spike upper, as well as the use of the virtual marker. Despite these precautions, marker placement was subject to some degree of human error, in particular as the participants swapped footwear between trials and error due to soft tissue artefact from the markers moving relative to the bones. During pilot work, it was ascertained that a 5 mm shift in the location of the MTH1 and MTH5 markers could result in a difference of up to 15% in the resultant joint moment. This was minimised as much as possible by palpating the landmarks and marking the marker locations on the skin. Furthermore, for particular anatomical landmarks (such as MTH2) the use of a virtual marker was used to minimise movement artefact.

Extensive attention was given to the processing of the data required as inputs to the IDA, in particular the accuracy of the CoP. For each sprint trial the CoP profile was visually inspected and thresholds were implemented for the beginning and end of foot contact to reduce error. Some trials had to be discarded due to foot contacts occurring outside or towards the edges of the force plates, which unfortunately reduced the number of trials for some participants. Whilst the contribution of individual muscles were not measured, implementing an IDA allowed the underlying joint kinetics to be described, through the calculation of resultant joint moments, powers and energies, providing insight with which research questions i, ii, and vi were answered.

The research in this thesis has provided a detailed assessment of foot function for a group of trained athletes during maximal velocity sprinting. Methods were developed to maximise the number of trials that could be recorded from trained sprint runners covering a range of ability levels. However, the low number of athletes used throughout Chapters three to five in this thesis reduced the validity of the statistical analysis. It is possible small differences may be teased out with more participants / trials. Nonetheless, in Chapter four, significant effects were found for a barefoot versus shod comparisons for a group of eight participants and the magnitude of the effects were deemed large. For ascertaining smaller differences between multiple footwear conditions (i.e. Chapter five and six), reporting data that documents the size or practical significance of the effects would help to identify meaningful differences (Knudson, 2009). Small differences, along with relatively large variability in kinematic / kinetic responses, may often make it challenging to identify a meaningful effect of an intervention. Alternative statistical analyses may be more sensitive than traditional analyses to detect such small systematic differences in joint motion. Therefore, future work may consider alternative methods such as principal component analysis, which can be performed with smaller sample sizes (Boyer, Federolf, Lin, Nigg & Andriacchi, 2012).

Typical coefficient of variations were presented in Chapter four and Appendix E, with typical values ranging from 5-15% for kinematic variables and 15-20% for kinetic variables (with the exception of energy produced during push-off). These values are not dissimilar to those presented by Salo & Grimshaw (1998) and Korhonen *et al.* (2010), showing that athletes were able to reproduce most parts of their performance within reasonably low limits. However, some of the differences between MPJ kinetic variables were not large enough to be significant; therefore a number of the key findings were not substantiated. Furthermore, the wide variation in individual responses to footwear conditions, demonstrated in Chapter five, meant that no clear trends were observed between footwear conditions and the resulting levels of performance, possibly leading to a 'mythical average performer'. It was therefore decided to continue the investigation using a single-subject approach.

The single-subject approach adopted in Chapter six, allowed specific individual responses, both performance and biomechanical, to be assessed without masking important effects by averaging group data. However, it included only two participants and whilst they were experienced track athletes, they were not elite level; therefore the degree of variability, particularly evident in the pressure patterns, may possibly have been reduced if elite athletes were recruited. Due to a relatively low number of trials (total of 12 sprints per testing session), the critical values used in the Model Statistic approach were high, which may have hindered the likelihood at determining significant differences, compared to other single subject studies that have used a higher number of trials, such as 25 landings used by Dufek and Bates (1991). Such a high number of maximal sprinting trials per footwear condition was not feasible for this study.

It is a recognised assumption that the results of the single subject analyses cannot be generalized to a wider population. The results of the final study therefore can only indicate that the intervention does have an effect on someone and can provide only preliminary data on the variability of results between subjects and indication on the consistency of the results (Reboussin & Morgan, 1996). Overall whilst a small number of subjects may not be sufficient to make definitive conclusions, performing multiple single subject designs on a few subjects may additionally provide some suggestion of the consistency of the effect of a condition (Reboussin & Morgan, 1996) and therefore may be of additional benefit. Data is needed on more participants to suggest the consistency of the effect of the intervention and to examine the variability of response patterns between participants. A combined approach (group and single subject analyses) would be beneficial for future work to provide further insight into the effect of shoe stiffness on sprinting performance and foot biomechanics.

7.4 Practical implications

The importance of the MPJ as a joint, which along with the ankle, knee and hip joints, has a role in the energetics of the lower limb, should be recognised by coaches. This research has shown that although the joint is a large absorber of energy, some active (energy-producing) dorsiflexion of the toes occurs during the push-off phase of sprinting. In strength and conditioning training, therefore the foot should not be ignored. It is suggested that strengthening exercises should not only target the extrinsic foot/ankle muscles (e.g. gastrocnemius and soleus, posterior tibialis, flexor hallucis longus, peroneus longus and brevis), but also include the intrinsic foot muscles (e.g. abductor hallucis and flexor digitorum brevis). To enable effective propulsion, the foot/ankle muscles must: be strong enough to stabilize the foot during the stance phase and therefore adjust to the underlying surface or shoe stiffness and to allow and facilitate recoil-reuse of the elastic energy by the elastic materials (Achilles tendon, Plantar fascia). Therefore it is recommended that exercises such as barefoot toe walks / heel walks / calf raises should be included alongside more traditional sprint training strength exercises. Furthermore, recent evidence by Potthast *et al.* (2005) suggests that barefoot-training footwear (Nike free shoe) initiates increases in toe flexor strength, MPJ active path of motion and increases in muscle volume of toe flexors. Coaches should, however, exercise caution if implementing barefoot training programmes or advising the use of barefoot footwear. It is not recommended that athletes perform sprint running barefoot.

It is evident that shoe selection is specific to the functional requirements of individual athletes and athletes will respond differently to footwear modifications, likely based on their past experiences and expectations. It was suggested in Chapter 5 that athletes and coaches should experiment with shoes of different stiffness to obtain optimal sprint time performance. It is speculated that highly trained / elite athletes are most likely to benefit from any small additional benefits of a stiff sprint spike. Therefore, when purchasing sprint spikes, it is not recommended that athletes always stick to the same brand / model of sprint shoes but, if possible, try a selection of shoes during training sessions, including stiffer shoes. For world class sprinters, it is recommended, with the support of footwear manufacturers / sponsors, that they have individually customized sprint spikes, based upon the results of laboratory biomechanical testing with differing sprint spike conditions, similar to the testing presented in Chapter 6. Footwear manufacturers may wish to consider matching the location of the bending axis of the sprint spikes with the location of the athlete's MPJ bending line and also increasing the stiffness of the sprint spike under the medial side of the forefoot, to match the results of the athlete's loading profiles from pressure measurements.

The same recommendations are not relevant for younger, developing athletes, with reduced ankle / foot muscular strength. As stiffer sprint spikes compromise MPJ angular motion, it is speculated that repeated use of extremely stiff sprint spikes for all training sessions may have a detrimental effect on the natural stiffness or flexibility of the MPJ, may reduce the efficiency of the windlass mechanism and increase the likelihood of injuries, especially in the Achilles and Plantar Fascia (due to increased work performed at the MPJ to overcome the high stiffness), especially when the muscles are still developing during adolescence.

7.5 Future Directions

Chapter six provided some evidence for different response strategies amongst two individuals who responded differently to sprint footwear conditions, both in terms of their foot function and sprinting performance. There are other individual factors: biomechanical; morphological; environmental, these are sources of variation which influence patterns of movement, which warrant exploration. Athletes will exhibit different responses to footwear conditions, due to the complexity of the human machine and its associated functional degrees of freedom, along with the different experiences, perceptions and expectations the participant brings to the experimental set up (Bates, 1996). Whilst these factors may be more difficult to explore, future work should identify key anatomical or biomechanical factors that influence foot function during sprinting. For example, the influence of the natural stiffness of the human forefoot, and the strength and velocity characteristics of the ankle plantarflexors and toe flexors, are morphological factors previously highlighted throughout the thesis which may influence foot function during sprinting and may dictate appropriate shoe selection. Potthast *et al.* (2005) demonstrated that a training footwear intervention could initiate biopositive adaptations within the foot (including significantly increased toe flexor strength and reduced MPJ dorsiflexion in walking gait). These adaptations could potentially be advantageous to sprinting performance, through stiffening of the MPJ. The efficiency of an individual's windlass mechanism function could potentially have a role on the MPJ function, this has been demonstrated with windlass enhancing running shoes (Payne *et al.*, 2006) yet it is unknown whether this could have any effect of running or sprinting performance. Therefore, one question for future research is: can training shoes elicit meaningful changes in the toe muscle strength and natural MPJ stiffness? And, in turn, what is the resultant effect on lower limb kinetics / energetics during sprinting and sprinting performance?

The inclusion of the magnitude of work performed, and the resultant energetics, at the ankle joint, possibly along with the knee and hip joint, would provide additional useful information for determining the effect of footwear characteristics on lower limb mechanics and the contribution of the lower extremity joints to mechanical energy, during sprinting. The determination of the relevance of shoe energetics to whole body dynamics requires an understanding of other passive energy exchange mechanisms in the body. These mechanisms include the transfer of potential and kinetic energy within and between body segments (*Hunter et al.*, 2004c) and the storage and recovery of strain energy in stretched muscle, tendon and connective tissue. Therefore, it is suggested that future research investigates the effect of both natural MPJ stiffness and sprint spike stiffness on the energetics of the whole of the lower body. A stiffer shoe system, by creating a more rigid link between the ankle joint, to the MPJ, to the point of application of the ground reaction force vector, may facilitate a more effective transfer of force

Besides changes to the overall longitudinal bending stiffness, other mechanical characteristics of sprint spike design warrant future investigation. For example, little is known regarding the influence of toe spring angle and the natural location of the bending line in sprint spikes, and the effects of these factors on the resultant motion at the foot. As this thesis has demonstrated substantial differences between the function of the medial and lateral components of the forefoot, perhaps future developments in sprint spike design could explore the effect of localised bending stiffness, for example greatly increasing the stiffness, specifically under MTH1-2, the hallux and second toe. The potential for customizing footwear (manipulations to localised stiffness or toe spring angle) relative to an individual's musculo-skeletal geometry, such as the lengths of their toes and the orientation of their MPJ axes, which could be obtained from DEXA scans, would be an interesting area for future research.

7.6 Conclusion

The aim of this research was to evaluate how foot function may contribute to sprinting performance and the interaction between the structure of sprinting footwear and performance, with a focus on the role of the MPJ.

Four specific research questions were addressed through a series of empirical investigations, thereby providing an original contribution to knowledge in the field. The assessment of two key methodological issues highlighted the importance of using an appropriate anatomical joint axis definition for the MPJ, along with a high cut-off frequency for processing high speed joint motion during sprinting. An analysis of kinematic, kinetic and pressure data from trained sprinters, identified normal patterns of foot behaviour and the functions of the MPJ during sprinting. Whilst the MPJ is clearly a large absorber of energy as the joint dorsiflexes during stance, the foot does appear to aid in propulsion of the sprinter, by creating a rigid lever for push-off and producing some (albeit small) power as the toes begin to plantarflex prior to the instant of take-off. It is clear from the considerable range of motion undergone at the MPJ during sprinting, along with the additional requirement of energy loss, that researchers should not ignore this joint in future analyses of sprinting biomechanics. The effect of different footwear conditions (barefoot versus sprint spikes and different stiffness's of sprint spikes) on foot function and sprinting performance was assessed. Sprint spikes appeared to have a clear localised effect on the function of the MPJ, increasing the work performed at the joint, facilitating a quicker loading transition and anterior progression of the centre of pressure, and enabling a more effective push-off. Due to high variability between athletes and highly individualised responses to stiffness pertubations in footwear, it was recommended that multiple single-subject analyses be undertaken. From the single-subject analyses, based upon the sprinting velocities of two athletes, there was evidence that for individual athletes, sprinting performance may be improved with the appropriate selection of footwear. Whilst varying the mechanical characteristics of sprint spikes clearly showed controlling influences over the natural motion of the MPJ, the relatively minimal effect on the resultant MPJ energetics, potentially suggests that the minimisation of energy loss concept does not hold true for sprint spikes.

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APPENDIX A. ACCURACY OF THE CENTRE OF PRESSURE

A.1 Caltester reports for the force platforms

Laboratory Calibration Test Report
Liverpool John Moores University

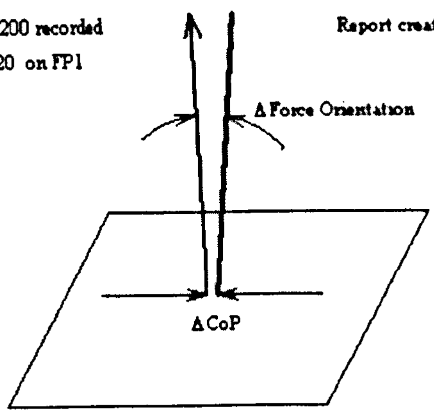
large middle.c3d

Number of Frames: 6816 of 7200 recorded

Minimum Force Setting (N): 20 on FP1

Collection Date: No Date in C3D File

Report created 8 - May - 2008



Report Format

	Min (at xxx N)	Mean ± 1 Std Dev	Max (at xxx N)

The mean, standard deviation, and range (minimum and maximum) of the report variables are provided on the graph and with numerical values. The applied force at the time of the minimum and maximum value is indicated alongside the value

Summary of Differences

	Min	Max
Δ Force Orientation Error (°) = 0.7 ± 0.3	0.0 (at 80 N)	2.6 (at 24 N)

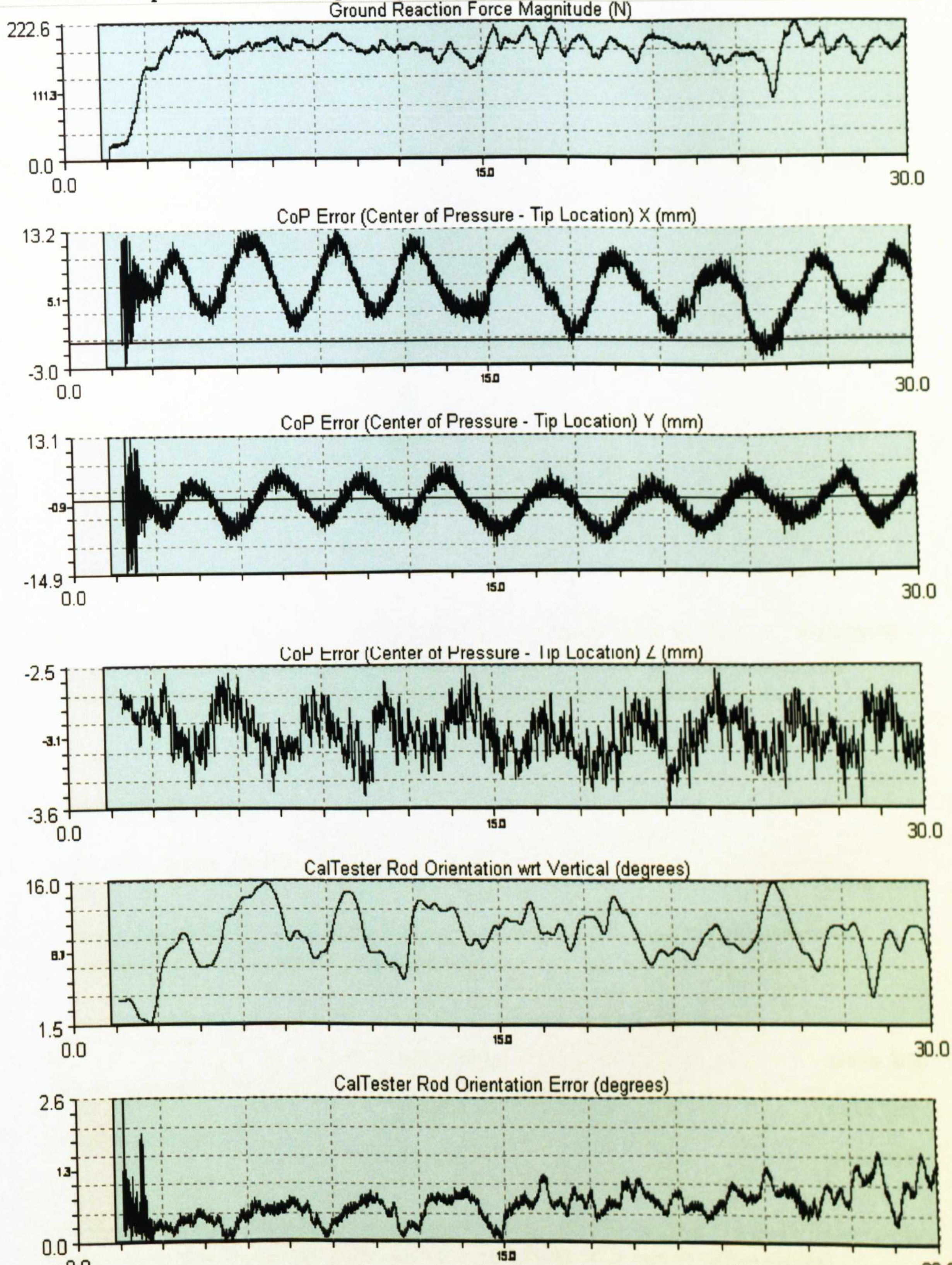
Δ Force Orientation is the angle between the orientation of the axial component of the applied force and the orientation of the long axis of the device, as determined from the target data.

	Min	Max
Δ CoP x (mm) = 6.3 ± 3.1	-3.0 (at 28 N)	13.2 (at 200 N)
Δ CoP y (mm) = -0.8 ± 3.3	-14.9 (at 29 N)	13.1 (at 27 N)
Δ CoP z (mm) = -3.0 ± 0.2	-3.6 (at 200 N)	-2.5 (at 173 N)

Δ CoP x,y,z are the components of the displacement vector between the Center of Pressure location measured by the force platform and the endpoint of the calibration testing device (adjusted for a vertical height of 2.4 mm above the force platform.)

Report generated by CalTester (© C-Motion Inc.)

CalTester Report : Selected Output Signals



Report generated by CalTester (© C-Motion Inc.)

Caltester report for top left corner of plate

Laboratory Calibration Test Report
Liverpool John Moores University

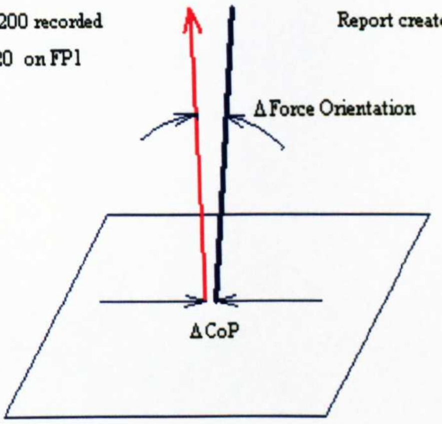
large top left corner.c3d

Number of Frames: 6471 of 7200 recorded

Minimum Force Setting (N): 20 on FP1

Collection Date: No Date in C3D File

Report created: 8 - May - 2008



Report Format

Min (at xxx N) Mean ± 1 Std. Dev. Max (at xxx N)

The mean, standard deviation, and range (minimum and maximum) of the report variables are provided on the graph and with numerical values. The applied force at the time of the minimum and maximum value is indicated alongside the value.

Summary of Differences

Δ Force Orientation Error (°) = 0.8 ± 0.4

0.0 (at 206 N) 11.8 (at 36 N)

Δ Force Orientation is the angle between the orientation of the axial component of the applied force and the orientation of the long axis of the device, as determined from the target data.

Δ CoP x (mm) = 2.6 ± 2.3

-32.7 (at 36 N) 12.6 (at 29 N)

Δ CoP y (mm) = -9.0 ± 3.8

-28.0 (at 27 N) 6.7 (at 29 N)

Δ CoP z (mm) = -1.8 ± 0.2

-2.4 (at 173 N) 2.9 (at 36 N)

Δ CoP x,y,z are the components of the displacement vector between the Center of Pressure location measured by the force platform and the endpoint of the calibration testing device (adjusted for a vertical height of 0.0 mm above the force platform.)

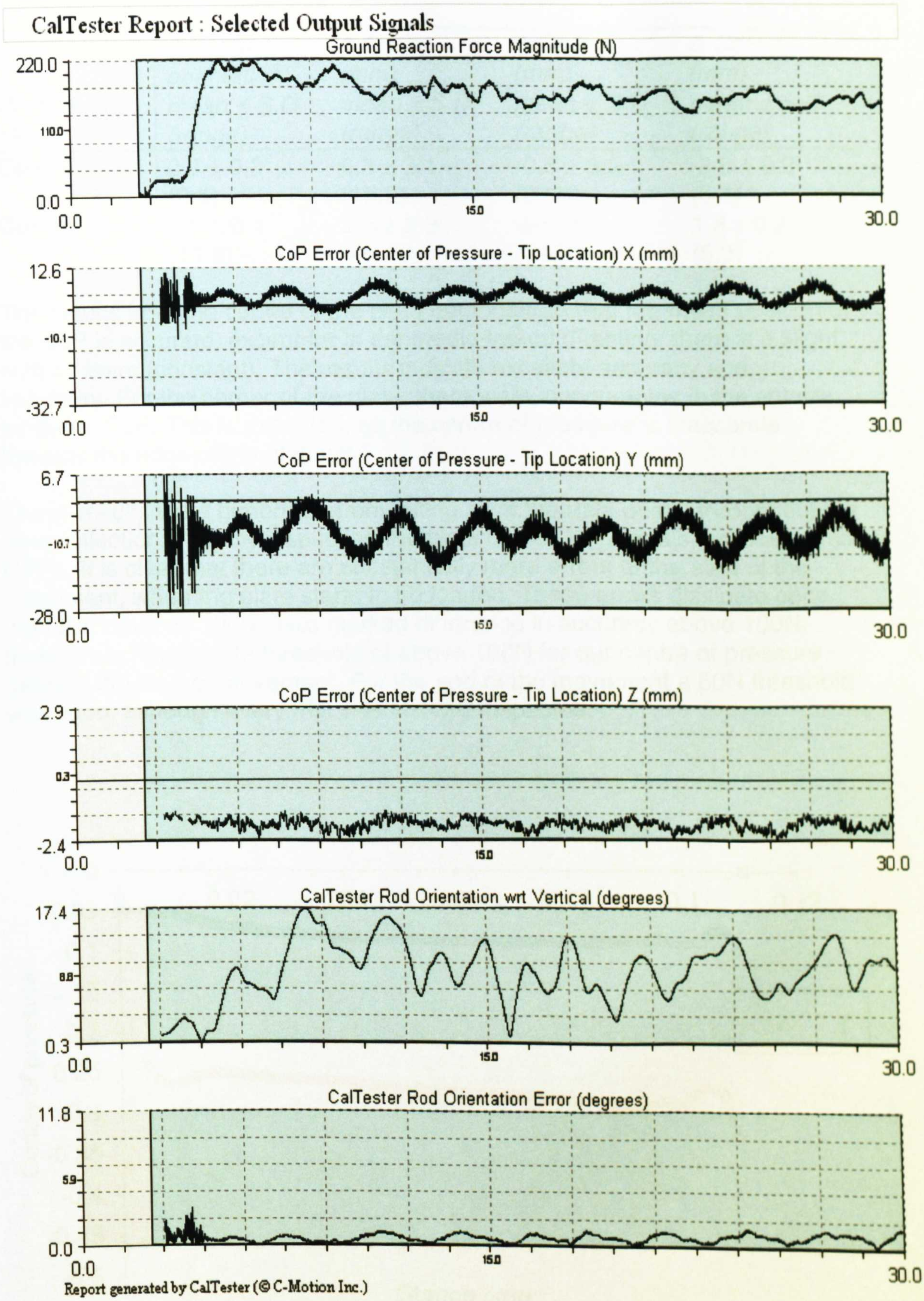

CalTester Report : Selected Output Signals
Ground Reaction Force Magnitude (N)
220.0
110.0
0.0
0.0
15.0
30.0
CoP Error (Center of Pressure - Tip Location) X (mm)
12.6
-10.1
-32.7
0.0
15.0
30.0
CoP Error (Center of Pressure - Tip Location) Y (mm)
6.7
-10.7
-28.0
0.0
15.0
30.0
CoP Error (Center of Pressure - Tip Location) Z (mm)
2.9
0.3
-2.4
0.0
15.0
30.0
CalTester Rod Orientation wrt Vertical (degrees)
17.4
8.8
0.3
0.0
15.0
30.0
CalTester Rod Orientation Error (degrees)
11.8
5.9
0.0
0.0
15.0
30.0
Report generated by CalTester (© C-Motion Inc.)

Table A.1 Caltester results

Position on plate	*Δ Force orientation°* *mean ± S.D* *(range)*	*Δ CoP x (mm)* *mean ± S.D* *(range)*	*Δ CoP y (mm)* *mean ± S.D* *(range)*	*Δ CoP z (mm)* *mean ± S.D* *(range)*
Centre	0.7 ± 0.3 (2.6)	6.3 ± 3.1 (15.2)	-0.8 ± 3.3 (28.0)	-3.0 ± 0.2 (6.1)
Corner	0.8 ± 0.4 (11.8)	2.6 ± 2.3 (45.3)	-9 ± 3.8 (34.7)	1.8 ± 0.2 (5.3)

The results from the centre of the plate demonstrate that the mean position of the CoP is accurate, except for in the medio-lateral direction; there is a slight shift by 6mm (constant). The results indicate excellent accuracy and variability. For the corner of the plate, there were inaccuracies in the antero-posterior CoP. This is expected, as the centre of pressure is inaccurate towards the edge of the platform.

These results were generated from taking all of the data points throughout the data collection. Visually inspecting the plots of the CoP signals, as well as the GRFs, it is clear that there are substantially more errors at the start of the movement, when the plate starts to be loaded. These errors dissipate once the GRF reaches 100N. This marked difference in accuracy above 100N, therefore a Force plate threshold of above 100N for our centre of pressure data for the start of movement. For the end of the movement a 50N threshold was used, although every trial was visually inspected.

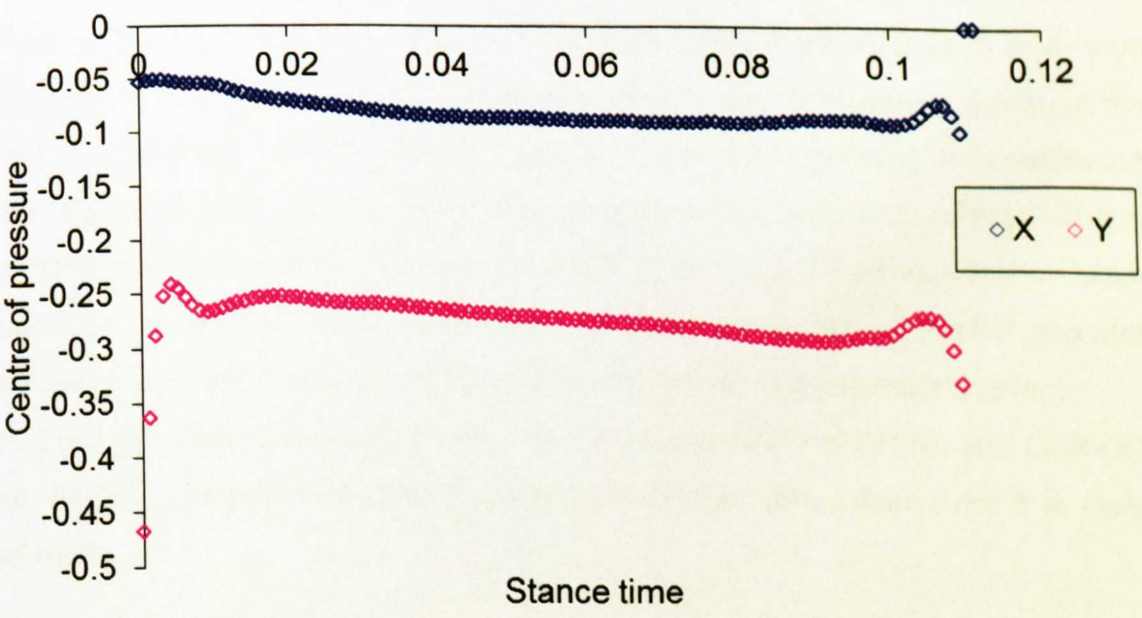


Figure A.1 Typical centre of pressure trace for one sprinter.

A.2 Inspection of marker locations relative to the CoP

Figures A.2 to A.9 demonstrate the location of the CoP (X coordinate = horizontal axis, Y coordinate = vertical axis) and marker locations of MTH1, MTH2 and MTH5 throughout stance for shod sprinting – one typical trial from eight participants.

Figures A.2 to A.9 demonstrate the CoP very quickly moves over from the MTH2-5 axis to the MTH1-2 axis, normally within the first 15% stance, where it then remains for take-off. Some subjects showed midfoot contact, but in most cases the CoP was always forward, therefore the joint moment arm positive and joint moment plantarflexor. The scatterplots were presented in the FP coordinate system with (0,0) being the centre of the FP. It is evident that most subjects demonstrated some degree of out-toeing on the force platform. However this was very minor, as can be seen from the scatterplots. The slight rotation of the foot on the force platform during stance will not effect the joint moment arms and MPJ kinetics during stance, when the joint line is modelled as MTH1-5. This is because of the way the moment arm is calculated, as the shortest perpendicular distance of the CoP to the line between MTH1 and 5. Therefore if the foot rotates and the CoP moves then the joint arm also moves. If the foot is greatly rotated on the force platform, then using the X location of the CoP for deciding when to switch from joint axis MTH2-5 to MTH 1-2 may not be the best method. However, because the foot was aligned well with the FP axis, this method remained. It is estimated that the maximum error of using this method would be 1 or 2 frames off (i.e. switching from the MTH2-5 over to MTH1-2 at most 2 frames early or later) because we did not demonstrate large rotations of the foot on the FP and also because the CoP is quickly moving from the lateral to medial component.
It is not possible to represent how the XY locations of the MTHs and CoP (XY scatterplot) change throughout the stance phase, other than how it is done above.

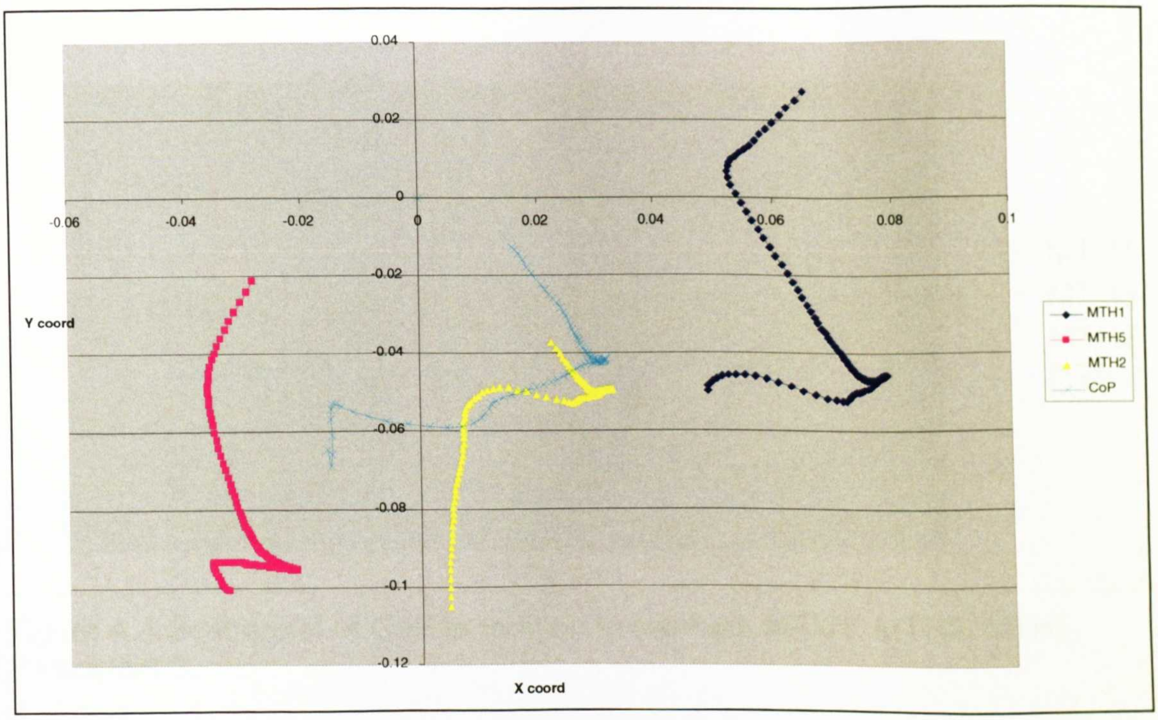


Figure A.2 Scatterplot of CoP in relation to markers MTH1, MTH2, MTH5. Participant 1.

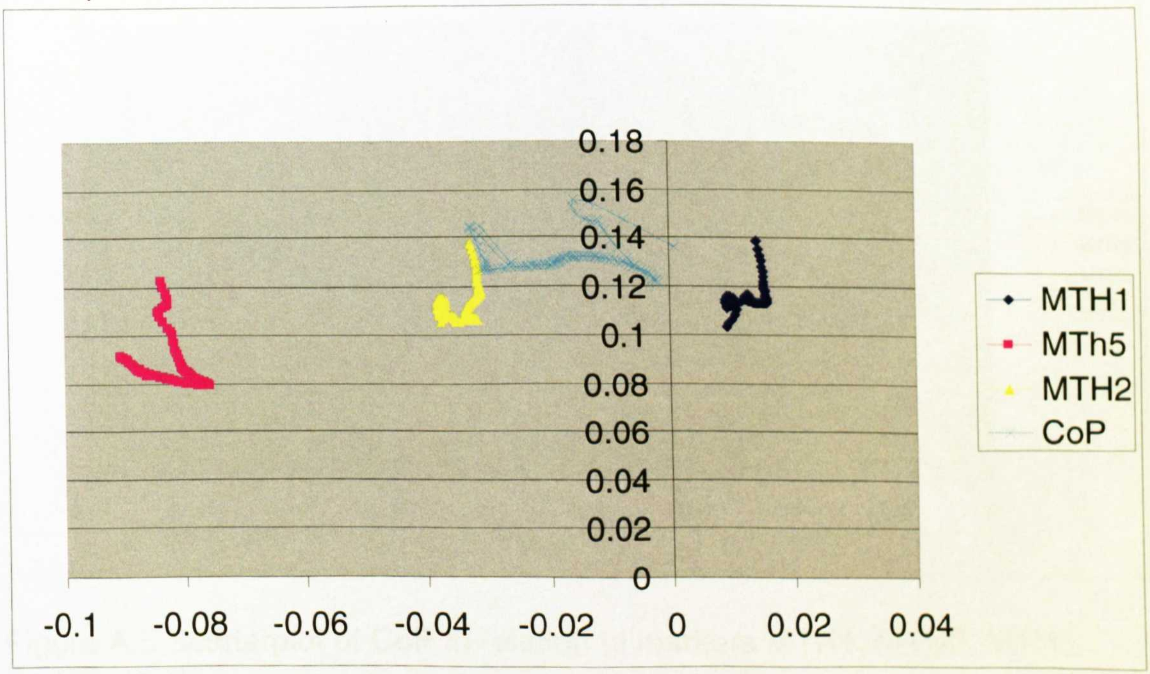


Figure A.3 Scatterplot of CoP in relation to markers MTH1, MTH2, MTH5. Participant 2.

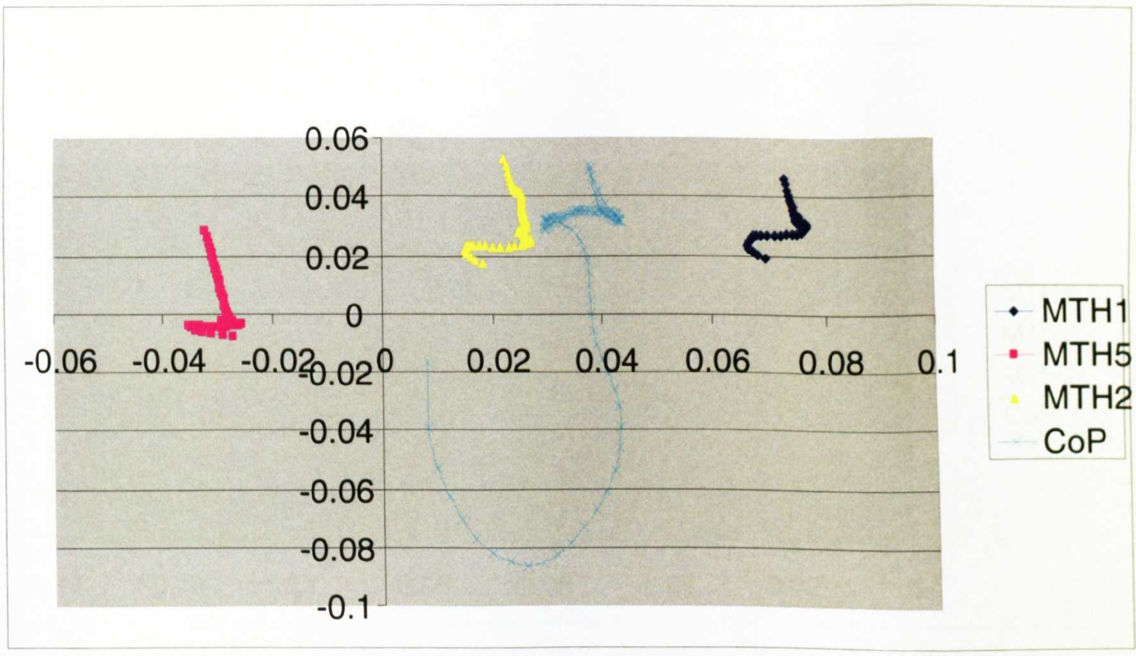


Figure A.4 Scatterplot of CoP in relation to markers MTH1, MTH2, MTH5. Participant 3.

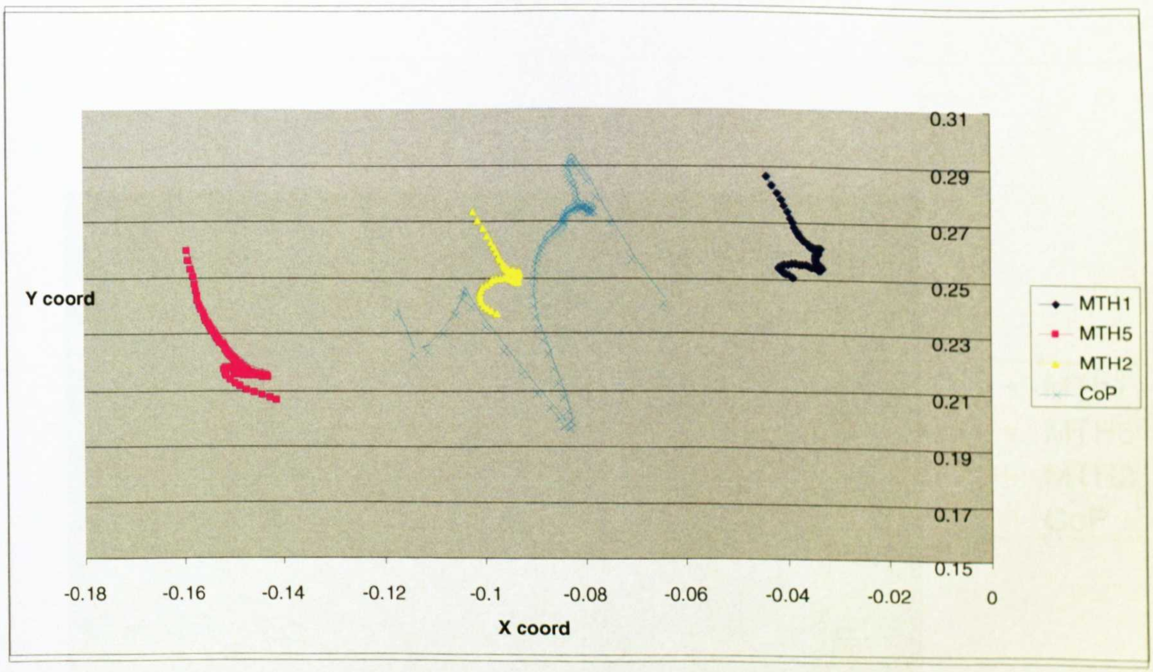


Figure A.5 Scatterplot of CoP in relation to markers MTH1, MTH2, MTH5. Participant 4.

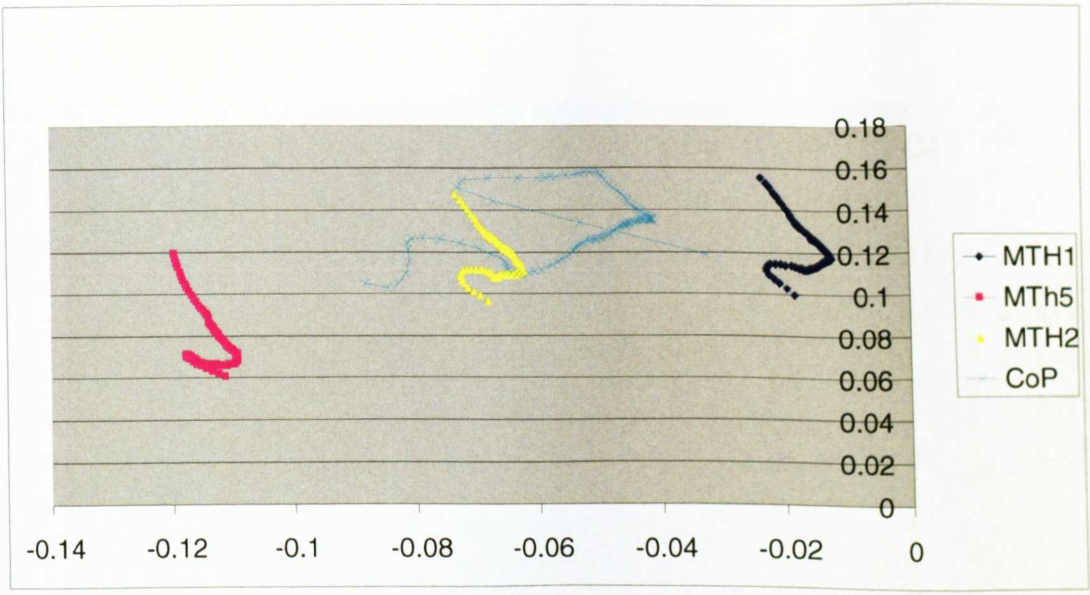


Figure A.6 Scatterplot of CoP in relation to markers MTH1, MTH2, MTH5. Participant 5.

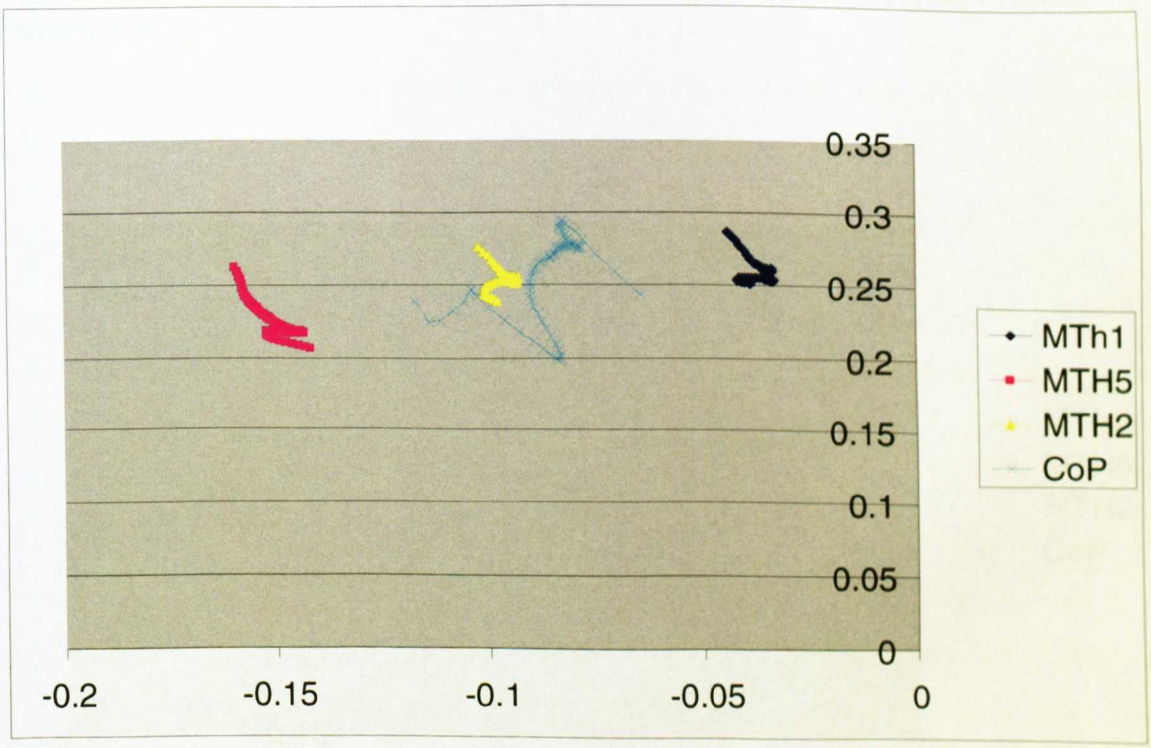


Figure A.7 Scatterplot of CoP in relation to markers MTH1, MTH2, MTH5. Participant 6.

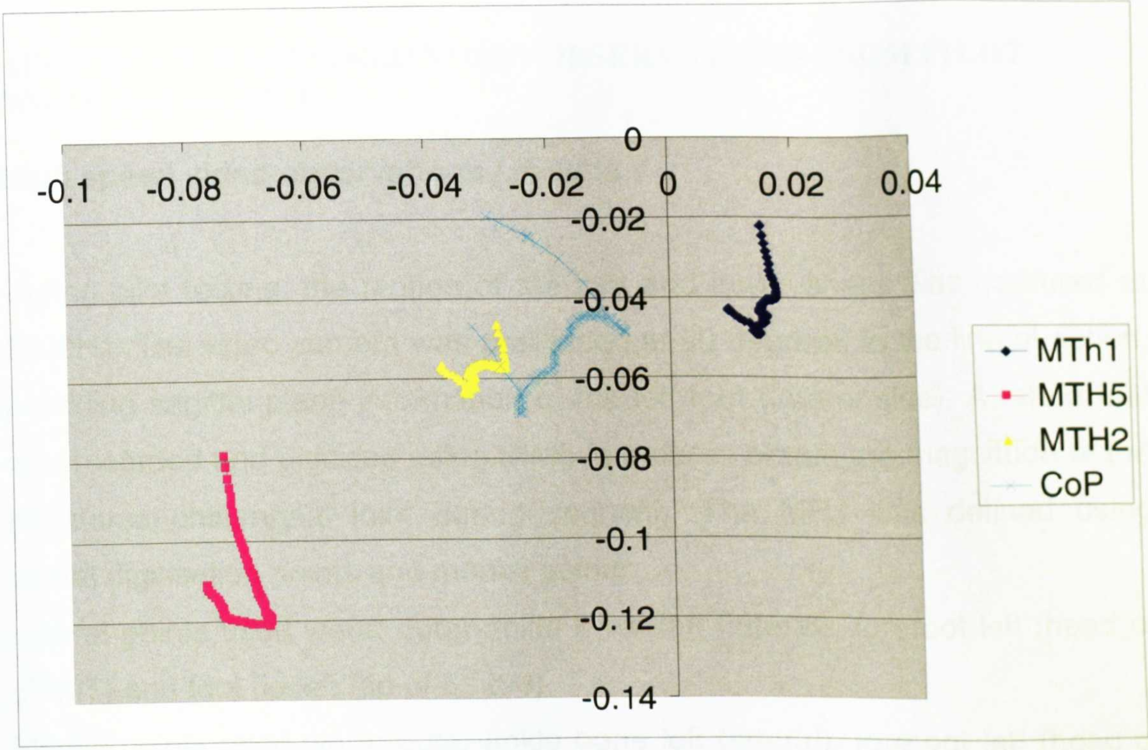


Figure A.8 Scatterplot of CoP in relation to markers MTH1, MTH2, MTH5. Participant 7.

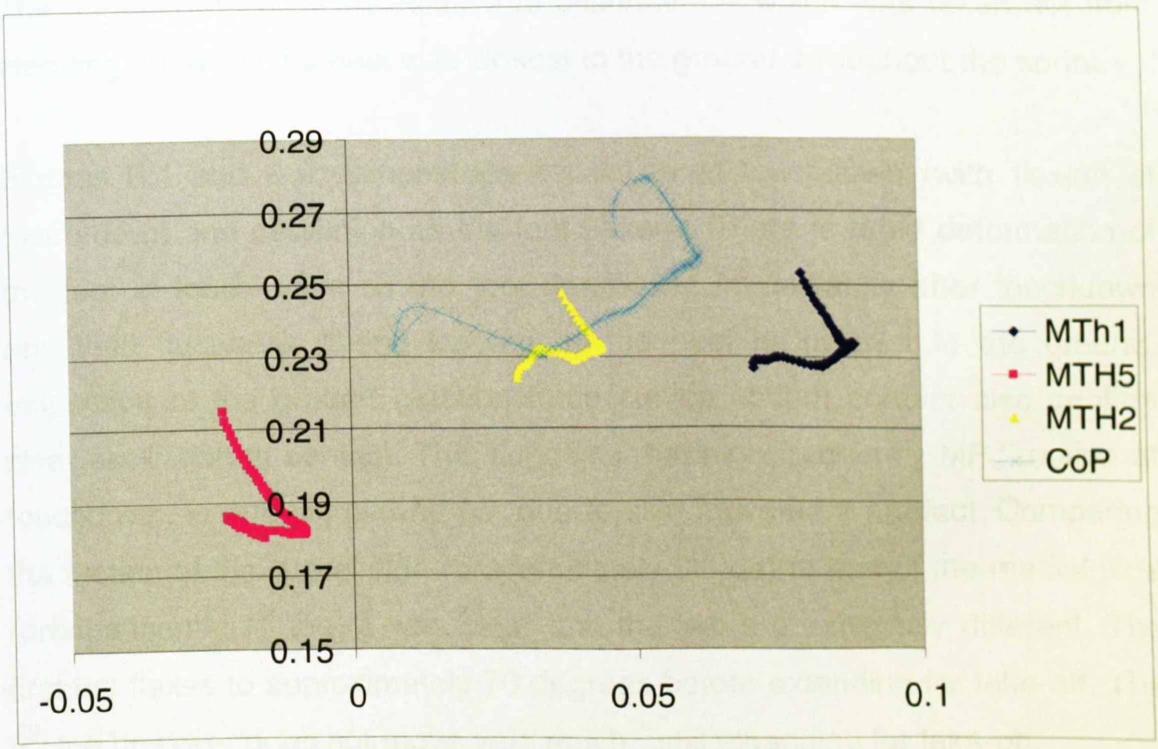


Figure A.9 Scatterplot of CoP in relation to markers MTH1, MTH2, MTH5. Participant 8.

APPENDIX B. HIGH SPEED VIDEO OBSERVATIONS FROM PILOT WORK (CHAPTER 3)

High speed video observations / results

During pilot testing, the motion of the foot and lower shank was captured at 2000Hz. The video camera was positioned at 90 degrees to the line of action, providing sagittal plane information of the left foot (lateral side). A shod trial was recorded and digitised using SIMI, in order to obtain the magnitude of the metatarsal-phalangeal joint during sprinting. The MPJ was defined using lateral digitisation points and medial points:

Lateral points used were: outer-ankle bone left (lateral), forefoot left (head of 5th MT) and foot tip left (tip of 5th toe)

Medial points used were: outer-ankle bone left (lateral), forefoot left (head of 1st MT head) and foot tip left (head of 1st toe)

The angles presented are relative to calibration – which was taken not from standing but when the heel was closest to the ground throughout the sprint.

Figures B.1 and B.2 demonstrate oscillation at touchdown (with flexion at touch-down and oscillation as the foot flattens. There is rapid deformation of the foot at touch-down as the foot dorsiflexes immediately after touchdown and then fluctuates during loading as the heel is lowered to the ground. Inspection of the ground reaction force curves at foot contact also confirm clear oscillation at contact. This suggests that high frequency MPJ motion at touchdown, is true signal and not due to skin movement artefact. Comparing the motion of the lateral side (predominately 5th toe) to that of the medial side (predominantly 1st toe) it was clear that the two are extremely different. The first toe flexes to approximately 70 degrees before extending for take-off. The 5th toe however does not move very much, until extending for take-off.

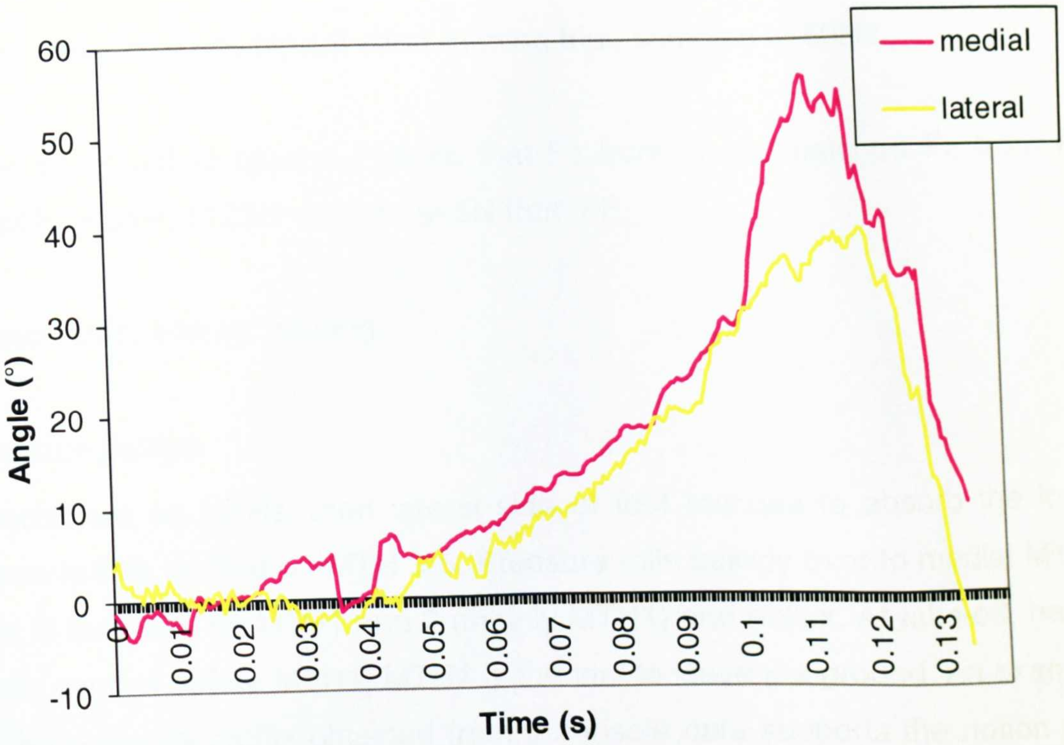


Figure B.1 Raw MPJ angle for one sprinting participant (shod) from high speed (2000 Hz) 2D video data

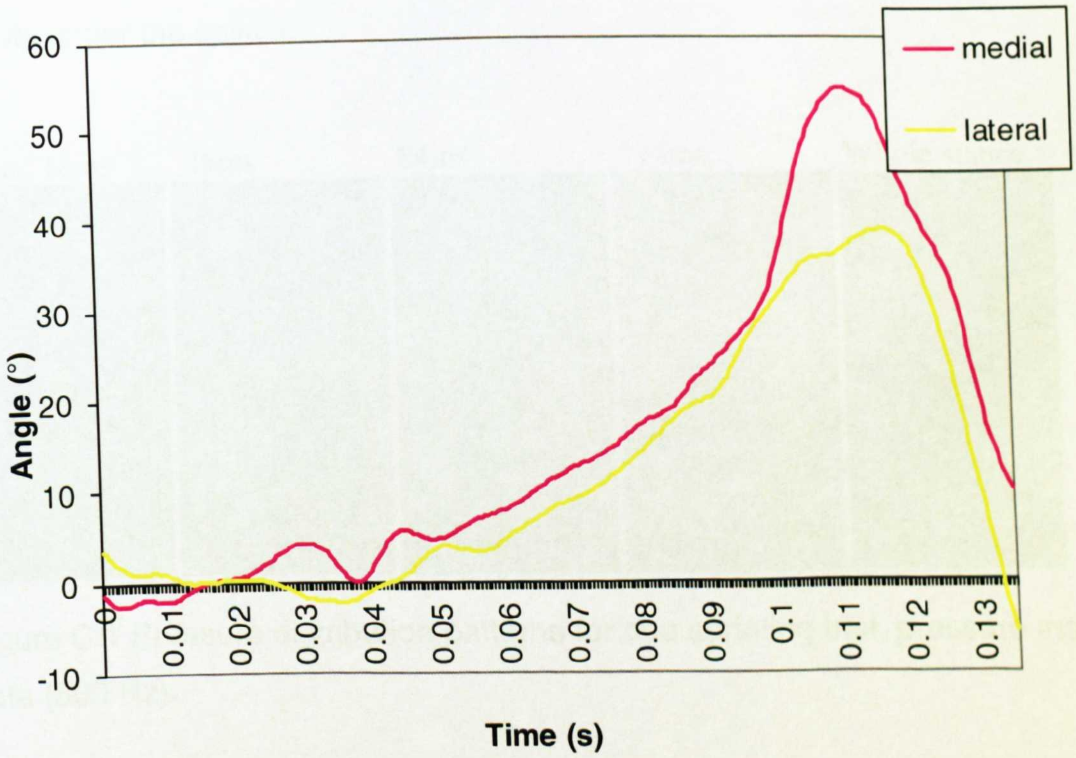


Figure B.2 MPJ filtered using Fc = 100 Hz, one sprinting participant (shod) from high speed (2000 Hz) 2D video data

APPENDIX C. INSOLE PRESSURE DATA

Insole data from one typical shod running trial, samples at 50Hz.

Gain increased to approx.. 1.5 so that Fz from insole matched Fz from FP (max Fz insole: 1923N versus 1945N from FP_

Stance time: 148ms: running.

Pressure pattern

Touchdown on MTHs, then lateral side of foot touches to absorb the load. There is little loading on MTH 3-5. Pressure rolls quickly over to medial MTHs and is focussed on MTH1 and 2 (mainly MTH1) and hallux. At take-off, hallux loses contact before MTH1, MTH1 is the last to leave the ground. An example of the pressure profile obtained from the insole data supports the notion of a reduced lateral to medial transition when wearing sprint spikes. This suggests that the different functions of the forefoot and loading profiles in barefoot and shod data are not just due to artefact generated because of the high pressure points under the spikes.

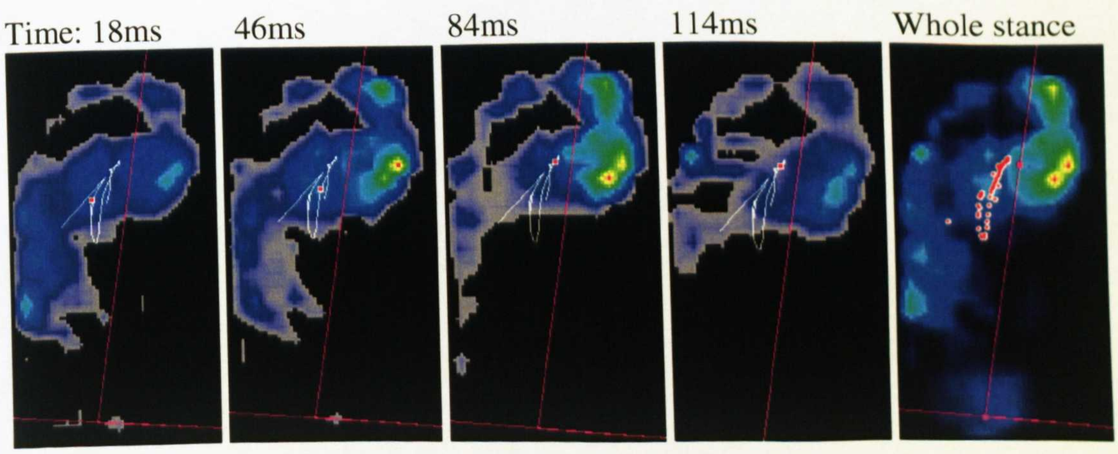
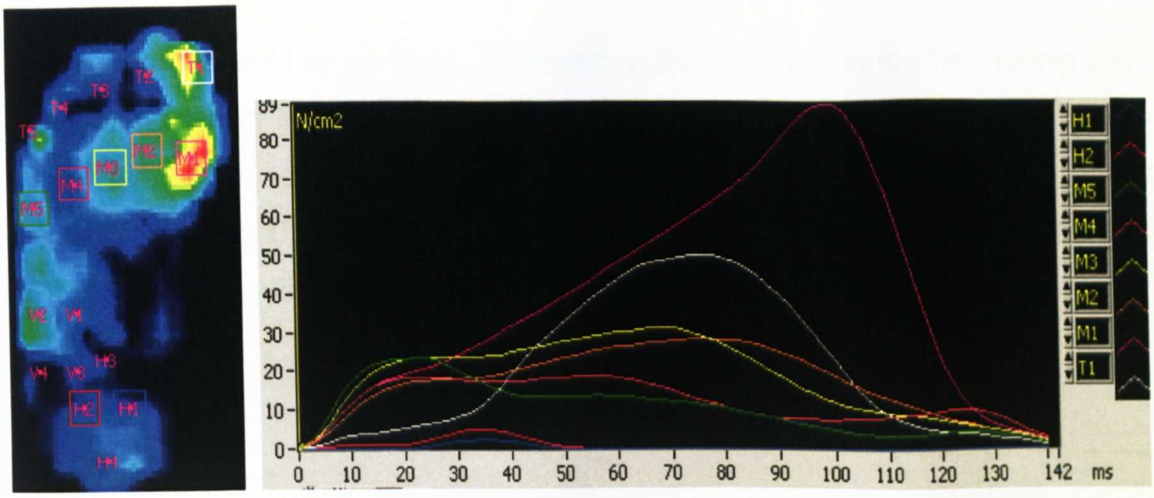


Figure C.1 Pressure distribution patterns for one sprinting trial, pressure insole data (500 Hz).



Peak Pressures

Left foot	X (pixels)	Y (pixels)	Size square (pixels)	Peak1(N/cm2)
H1	4	7	2	2.33
H2	4	10	2	4.98
M5	16	13	2	23.46
M4	18	10	2	18.77
M3	19	8	2	31.25
M2	20	6	2	28.20
M1	20	2	2	89.08
T1	25	2	2	50.17

Figure C.2 MTH loads for one sprinting trial, pressure insole data (500Hz).

APPENDIX D. MPJ ANGLE DATA, CHAPTER 5

Raw kinematic MPJ angle data, collected at 600 Hz was smoothed using a 4th order Butterworth low pass digital filter at different cut-off frequencies. From visual inspection of the resulting MPJ angle and angular velocity curves (Figures D.1 and D.2) the cut off frequencies of 60 Hz for the MPJ angle and 30 Hz for the angular velocity were chosen.

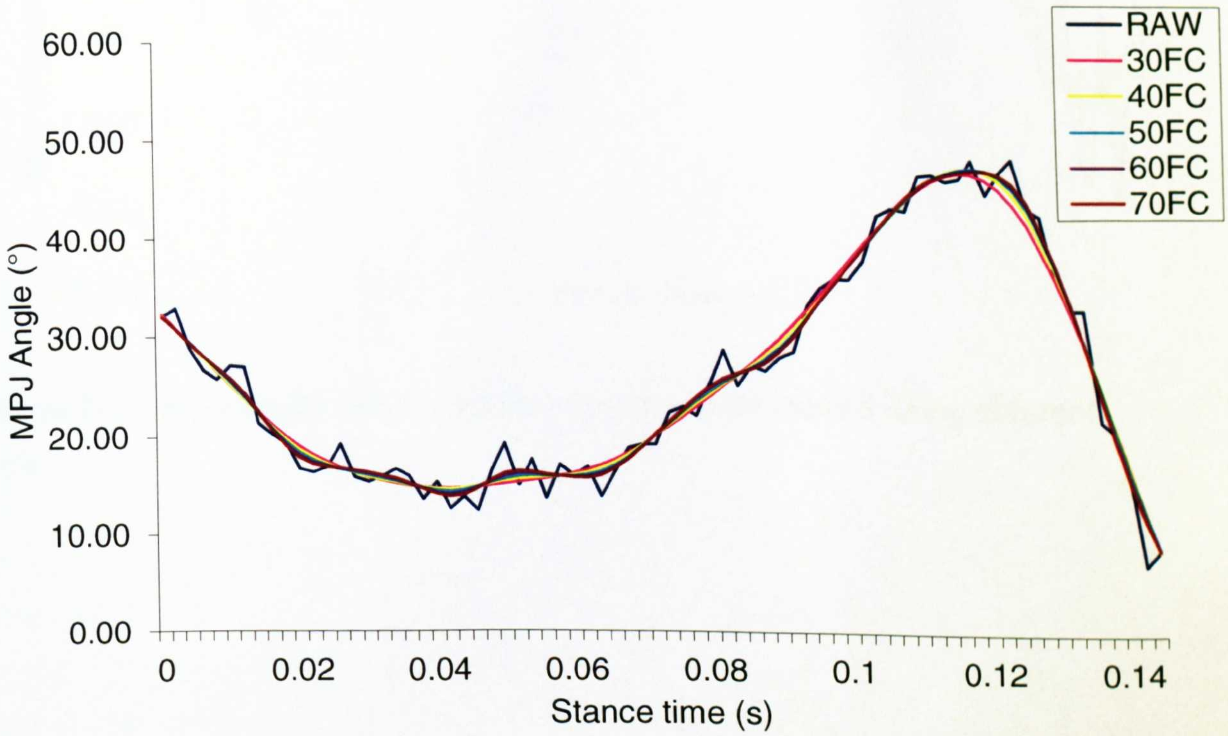


Figure D.1 MPJ angle for one sprinter, data filtered using different Fc's.

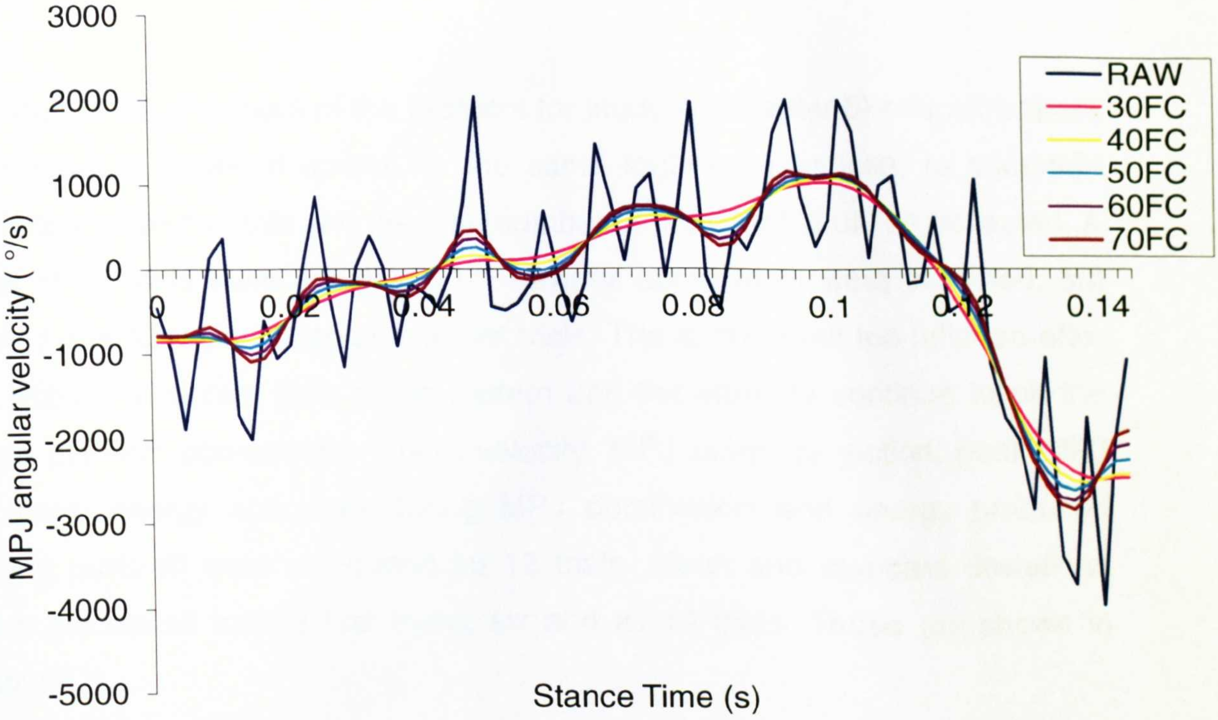


Figure D.2 MPJ angular velocity for one sprinter, data filtered using different Fc's.

APPENDIX E. Chapter 6 protocol – trial variability

During the development of the protocol for study 4 (Chapter 6) one participant performed a series of sprints, in the same footwear condition, to ascertain variation between trials and the total number of trials that could be collected. A total of 12 trials were analysed, 14 maximal sprint trials were collected, but from these 12 were acceptable sprint trials. The sprinter felt too fatigued after 14 trials to maintain their stride pattern and therefore to continue to hit the force platform consistently. Sprint velocity, MPJ range of motion, peak MPJ moment, energy absorbed during MPJ dorsiflexion and energy produced during push-off were calculated for 12 trials. Mean and standard deviations were calculated for the first three, six and all 12 trials. These are shown in Table E.1

Table E.1 Selected biomechanical variables for one participant performing twelve trials in the same footwear condition.

	N = 3	N = 6	N = 12
Sprint velocity (m/s)	7.85 ± 0.07	7.85 ± 0.06	7.88 ± 0.13
MPJ ROM (◦)	37.3 ± 2.9	36.9 ± 3.8	38.3 ± 3.6
Peak MPJ Moment (N·m)	72.2 ± 3.4	70.7 ± 5.3	71.5 ± 6.3
Energy absorbed during MPJ dorsiflexion (J)	-35.3 ± 1.2	-34.9 ± 4.2	-35.5 ± 3.4
Energy produced during push-off (J)	0.8 ± 0.9	0.6 ± 0.8	0.7 ± 0.6

The coefficients of variation for sprinting velocity ranged from 0.8% (six trials) to 1.6% (12 trials). For MPJ kinematic and kinetic variables the CoV for twelve trials was 9.3% for MPJ ROM, 7.5% for MPJ Moment and 9.5% for MPJ energy absorption, however for energy produced during push-off the CoV was 125%.

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APPENDIX F. PARTICIPANT QUESTIONNAIRES

F.1 Study 3 (Chapter 5) Subject post-testing Questionnaire

1. Please rank the sprint shoes in order of preference, which shoe you believe you performed fastest in. Write the colour of the sprint shoe in the space provided.

Ranking:

1 (fastest) ________________________________

2 ________________________________

3 ________________________________

4 (slowest) ________________________________

2. Explain why you gave the above rankings

____

____

____

____

____

3. Please rank the sprint shoes in order of stiffness (1 = most stiff, 4 = least stiff), writing the colour of the shoes in the space available. For each of your rankings indicate whether you thought the shoe stiffness was appropriate by circling too stiff / just right / too flexible

Ranking		Please circle:
1 (most stiff)	________________________	too stiff / just right / too flexible
2	________________________	too stiff / just right / too flexible
3	________________________	too stiff / just right / too flexible
4 (least stiff)	________________________	too stiff / just right / too flexible

F.2 Study 4 (Chapter 6) Subject during -testing Questionnaire

After trial 2 (compared to baseline previous trial)
Rate the stiffness (in terms of your perception of how much your forefoot was bending) of the condition relative to the baseline. Baseline has a value of 100.
Value = _________
Rate your performance (how fast you think you ran) for that trial. Baseline has a value of 100. Value = _________

After trial 4 (compared to baseline previous trial)
Rate the stiffness (in terms of your perception of how much your forefoot was bending) of the condition relative to the baseline. Baseline has a value of 10.
Value = _________
Rate your performance (how fast you think you ran) for that trial. Baseline has a value of 100. Value = _________

After trial 6 (compared to baseline trial previous trial)
Rate the stiffness (in terms of your perception of how much your forefoot was bending) of the condition relative to the baseline. Baseline has a value of 100.
Value = _________
Rate your performance (how fast you think you ran) for that trial. Baseline has a value of 100 Value = _________

After trial 8 (compared to baseline trial previous trial)
Rate the stiffness (in terms of your perception of how much your forefoot was bending) of the condition relative to the baseline. Baseline has a value of 100.
Value = _________
Rate your performance (how fast you think you ran) for that trial. Baseline has a value of 100. Value = _________

After trial 10 (compared to baseline trial)
Rate the stiffness (in terms of your perception of how much your forefoot was bending) of the condition relative to the baseline. Baseline has a value of 100.
Value = _________
Rate your performance (how fast you think you ran) for that trial. Baseline has a value of 100. Value = _________

After trial 12 (compared to baseline trial previous trial)
Rate the stiffness (in terms of your perception of how much your forefoot was bending) of the condition relative to the baseline. Baseline has a value of 100.
Value = _________
Rate your performance (how fast you think you ran) for that trial. Baseline has a value of 100. Value = _________

F.3 Study 4 (Chapter 6) Subject post-testing Questionnaire

1. Please rank the sprint shoes in order of preference, which shoe you believe you performed fastest in. Write the colour of the sprint shoe in the space provided.

Ranking:

1 (fastest) ______________________________

2 ______________________________

3 ______________________________

4 (slowest) ______________________________

2. Explain why you gave the above rankings

____

____

____

____

____

3. Please rank the sprint shoes in order of stiffness (1 = most stiff, 4 = least stiff), writing the colour of the shoes in the space available. For each of your rankings indicate whether you thought the shoe stiffness was appropriate by circling too stiff / just right / too flexible

Ranking		Please circle:
1 (most stiff)	________________________	too stiff / just right / too flexible
2	________________________	too stiff / just right / too flexible
3	________________________	too stiff / just right / too flexible
4 (least stiff)	________________________	too stiff / just right / too flexible

APPENDIX G. ETHICS FORMS

G.1 Ethical approval

Ref. 0753

Thursday 19th October 2006

Dear Grace,

I am pleased to inform you that the Ethics Committee has considered your application for approval of the project entitled:

Biomechanics of foot function in relation to sports performance

And I am happy to confirm that it has been approved.

The Ethics Committee approval is given on the understanding that:

(i) any adverse reactions/events which take place during the course of the project will be reported to the Committee immediately;

(ii) any unforeseen ethical issues arising during the course of the project will be reported to the Committee immediately;

(iii) any change in the protocol will be reported to the Committee immediately.

Please note that ethical approval is given for a period of five years from the date granted and therefore the expiry date for this project will be **October 2011**. An application for extension of approval must be submitted if the project continues after this date.

I am enclosing form EC5 and would be grateful if you could spare the time to complete the questionnaire and return it to me.

Yours sincerely,

Research and Graduate School Office

Top Floor, Rodney House, Liverpool L3 5UX Facsimile +44 (0)151-231-3724 http://www.livjm.ac.uk/research_and_graduate/

G.2 Example Informed Consent form

EC3

LIVERPOOL JOHN MOORES UNIVERSITY
FORM OF CONSENT TO TAKE PART AS A SUBJECT IN A MAJOR PROCEDURE OR RESEARCH PROJECT

Title of project/procedure: Biomechanics of foot function in relation to sports performance.

I, ... agree to take part in
(**Subject's** full name)*
the above named project/procedure, the details of which have been fully explained to me and
described in writing.

Signed .... Date ....
(Subject)

I, GRACE SMITH certify that the details of this
(**Investigator's** full name)*
project/procedure have been fully explained and described in writing to the subject named above and
have been understood by him/her.

Signed .... Date ....
(Investigator)

I, ... certify that the details of this
(**Witness'** full name)
project/procedure have been fully explained and described in writing to the subject named above and
have been understood by him/her.

Signed .... **Date** ....
(Witness)

NB The witness must be an independent third party.

* Please print in block capitals

G.3. Example Participant Information Sheet (Study 1, Chapter 3)

Participant Information Sheet

Name of experimenter: GRACE SMITH
Supervisor: DR MARK LAKE

Title of study/project: Biomechanics of foot function in relation to sports performance

Purpose of study: To develop a multi-segment foot model, including forefoot and rearfoot segments, and evaluate the role of the foot and footwear in high speed locomotion.

Procedures and Participants Role: You will be required to come to Henry Cotton Campus for one testing session. On arrival to Henry Cotton Campus, you will be escorted to the Biomechanics Laboratory. You should wear your own athletic training clothes and bring your own running shoes / spikes. At the laboratory your height and weight will be measured. You will be scanned using the DEXA scanner machine, to obtain body segment parameters of your foot. Following this, you will independently perform your own athletic warm up. Small reflective markers will then be attached to your shoe and lower body using double sided tape. These will remain in the specific anatomical position throughout the testing. Pressure insoles will be inserted into your running shoes. Carbon plate insoles will also be inserted into your running shoes You will first be asked to stand still on the force platform whilst a standing calibration takes place, where the cameras will obtain the position of the markers. You will then be instructed to begin a series of maximal sprints / jumps over the force platform, where force, pressure and kinematic (camera) data will record your movements. The experimenter will inform you of the number of trials you will need to successfully complete. You will obtain sufficient rest between trials. After successful trials have been completed you can remove the markers and pressure insoles, and complete your cool down. The whole procedure will take approximately 2 hours. You have the right to withdraw at any time.

Please Note:
All participants have the right to withdraw from the project/study at any time without prejudice to access of services which are already being provided or may subsequently be provided to the participant.