
**The kinematic effects of three quarter and full length foot
orthoses on anterior knee pain sufferers when walking
and descending stairs**

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Declaration

I declare that this thesis is entirely my own work, and is a true representation of the results obtained from the experiments undertaken at The University of Central Lancashire.

I declare that no material contained within this thesis has been used for any other academic award, and while registered for this research degree I have not enrolled as a candidate or student for another award at this University, or other academic or professional institution.

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Abstract

Background: Patellofemoral pain is a common disorder whose aetiology is complex often being described as multifactorial, increased load of the patellofemoral joint is often attributed to foot function. Foot orthoses are commonly prescribed for this condition; however the mechanisms by which they work are poorly understood. Previous studies using single segment foot models have hypothesised that it may be control of the midfoot which hold the key to understanding orthotic control. Over the last decade biomechanical analyses has advanced so it has become possible to divide the foot into segments, however no previous studies have investigated the use of orthoses on different segments of the foot when shod.

The overall aim of this study was to investigate the differences seen in the kinematics and kinetics of the lower limb between a patellofemoral pain group and a group of normals when using a standardised orthosis prescription during walking and descending a step.

Method: Initially fifteen healthy subjects had foot orthoses moulded to their feet, they were asked to walk at a self-selected pace and complete a 20cm step down; comparisons were made between sandals and shoes, plus two different orthoses. Kinematic and kinetic data were recorded using 10 Oqus cameras and 4 AMTI force platforms. The shoe data from the 15 healthy subjects was re-analysed and used as a control group to compare against 15 subjects diagnosed with patellofemoral pain. The foot was modelled using the calibrated anatomical systems technique (CAST) fixing the marker set directly on the feet and shoes of normal subjects which permitted comparisons of excursions between the shoes and sandals and the effects of the orthoses.

Results 1: Similar changes in the pattern of movement were seen between the shoe and the sandals conditions with and without the orthoses; the shoes reduced the excursions recorded except the transverse plane of the rearfoot. At the knee maximum extension was increased and maximum flexion at toe off was reduced by the orthoses.

Initial Conclusions: Expectedly the shoes reduced the range of motion over the sandal condition in most planes; however the similar effects seen with the orthoses in both types of footwear suggesting it was acceptable to use shoes in the later study.

Results 2: Significant differences were seen between the healthy subjects and the patellofemoral pain subjects at the foot and the knee. Both orthoses produced statistically significant results at the foot. In addition there was a significant reduction in the knee coronal plane moment range during the forward continuum phase of step down; this was attributed to a change in the ground reaction force as there were no changes reported in the kinematics of the knee.

Conclusions: The method of placement of the markers was able to detect small changes within the foot segments. This study identified potentially important differences between the patellofemoral pain subjects and the normals in both the knee and foot segments. However due to the lack of pain during the walking and step down trials it could not be determined if the changes were due to pain avoidance mechanisms or if they were causative factors. Many of the changes produced by the orthoses tended to be local to the foot, except for the knee coronal plane moment range during the forward continuum phase of step down. To the authors knowledge this work is unique in its investigation of the motion of foot segments while shod and confirmed the clinically held belief it is essential to consider footwear when prescribing orthoses to patients. The use of foot mechanics could be of interest to further research and may help to define sub-populations within this condition.

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Glossary of terminology

Closed Chain Motion – the distal segment of a joint is fixed while the proximal segment moves, such as during the stance phase of walking.

Concentric – where the muscle shortens during activity often associated with power production during gait.

Eccentric - where the muscle lengthens during activity, often associated with power absorption during gait.

Forefoot – used to denote front of foot can be used to describe plantar surface of the metatarsal heads in relation to the plantar surface of the heel. This study uses phalangeal segment in order to avoid confusion.

Going/tread – the horizontal surface of a step or stair.

Midfoot – often used clinically to describe midtarsal joint motion, used in this study to describe the centre segment of a 4 segment foot model which lies just proximal to the metatarsals.

Minimum clinical important change – is the smallest change in score that is noticed by individuals due to an intervention.

Minimum clinical important difference – is the smallest difference in scores of an outcome measure that is perceived by patients as being beneficial or harmful.

Minimum detectable change – is the smallest change that falls outside the level of measurement error.

Open Chain motion – the proximal segment of the joint is fixed while the distal segment moves, such as kicking a ball.

Pronation –motion occurring in all three cardinal planes simultaneously often associated with the subtalar joint it is a combination of abduction, eversion and dorsiflexion.

Q-angle – is the resultant angle between the anterior superior iliac spine to the mid-point of the patella and the mid-point of the patella to the tibial tubercle, (Herrington and Nester, 2004).

Rearfoot – used in this study to denote calcaneal segment motion with respect to the tibial segment.

Riser - the vertical component of a steps height.

Supination - motion occurring in all three cardinal planes simultaneously often associated with the subtalar joint it is a combination of adduction, inversion and plantarflexion.

Chapter 1. Introduction

This chapter examines the terminology and prevalence of anterior knee pain from previous studies; a brief outline of the function of the patella is presented. This leads into an overview of how foot function is thought to affect the patellofemoral joint and considers why foot orthoses are prescribed for this painful condition. To understand any pathology with an insidious onset the typical underlying mechanics must first be defined.

1.1. Patellofemoral Pain

Anterior knee pain is one of the most common lower limb disorders seen in musculoskeletal clinics (Messier, et al. 1991; Lafortune, et al. 1994; Heng and Haw, 1996; Powers, et al. 1999; Way, 1999; Duffey, et al. 2000; Brechter and Powers, 2002; Anderson and Herrington, 2003; Vincenzino, et al. 2008; Wilson and Davies, 2008; Barton, et al. 2009; Barton, et al. 2011). It is well discussed in the literature that anterior knee pain is a rather vague term for the condition; other terms used like chondromalacia patellae, retro-patella pain syndrome are no better; patellofemoral pain syndrome is a more favoured title and it has been suggested less misleading (Heng and Haw, 1996), hence will be used in this thesis.

Heng and Haw, (1996) suggested it is the reason for 20% to 40% of knee presentations in sports injury practice, while Brechter and Powers, (2002) suggested that one in four of the sporting population may be affected by this disorder, Callaghan and Selfe, (2007) conducted a review into the incidence of patellofemoral pain; they found only one study was conducted in the UK and most previous studies had recruited from sports clinics or military institutions. This led them to conclude that the prevalence of this condition in the general population is unknown. This did not prevent Aliberti, et al. (2010) stating patellofemoral pain affects 25-30% of the general population. At the second patellofemoral pain research retreat held in Ghent 2011 it was stated that 2.5 million runners will be diagnosed with patellofemoral pain in any one year (Taunton, et al. 2002).

Taunton, et al. (2002) proposed that the benefits of exercising on general health were promoted through the 1970s, and as a result many people started to exercise; running was one of the main regimes of choice due to its convenience and the fact it is free to participate. They noted that this increased the potential of being injured, their study looked at 2002 runners who were injured; 42.1% were found to have knee related pain, the most common diagnosis being patellofemoral pain syndrome, this condition was observed in 331 of the patients. Barton, et al. (2009) proposed it is essential for research to identify any differences in the kinematics of subjects with and without patellofemoral pain syndrome to aid the development of treatment and prevention of this debilitating condition.

Richards, (2008. Page 182) acknowledged that the precise role of the patella is still not determined with some authors claiming it acts as a “fulcrum for the extensor mechanism of the knee”, while others suggest it is there as a “balance beam for the patella tendon force and the quadriceps force”. Wilson and Davis, (2008) advocated that the patella is present to increase the moment arm of the quadriceps and as such improves the ability of the anterior thigh muscles to extend the knee. Heegaard, et al. (1995) also stated that the patella was present to increase the efficiency of the extensor forces throughout flexion, but they also suggested that the patella provides a gliding mechanism for the quadriceps muscles when the knee is flexing. The 2nd patellofemoral pain research retreat stated the “patella acts as a dynamic lever” which is put under the greatest loads during normal motion; this varies from around 0.5 times body weight for walking up to 7 times body weight for squatting (Powers, et al. 2012. Page A3).

Lee, et al. (2003) also noted that the patellofemoral joint has to deal with loads that exceed body weight and therefore it has the thickest covering of cartilage of any joint in the human body. Lee, et al. also stated that normal functioning of the joint is reliant on all the structures around the patella acting in balance, they described the quadriceps as active stabilizers while the congruency of the bony and cartilaginous surfaces, the peri-patella retinaculum and the patella tendon, were termed passive stabilizers. Richards, (2008) noted that during level walking the knee demonstrated 33% higher compressive forces than the hip, however during stair climbing it was noted this compression increased to approximately three times body weight.

1.2. Patellofemoral Pain and Foot Function

It is often accepted clinically that dysfunction of the foot can impart abnormal forces on the knee; it is usually suggested that excessive pronation of the foot causes the tibia to internally rotate as it follows the talus. Way, (1999) suggested that this rotation ultimately leads to femoral internal rotation which increases the Q-angle. Tiberio, (1987) explained that the gait cycle requires a coordinated sequence of lower limb joint movements. He proposed that during closed chain motion when the knee is flexed between 15°-20° extension must be accompanied by external rotation, while flexion must be associated with internal rotation, this phenomenon is termed “automatic rotation”, independent rotation can occur when the knee is flexed by more than 20°. This facilitates the argument that the sub-talar joint should pronate during the contact and early midstance phases of gait while the knee is flexing but then must re-supinate as the knee starts to extend. Any anatomical abnormality which causes excessive or delayed foot pronation will interrupt this; as the foot is held by friction on the supporting surface, the rotation of tibia is fixed.

Tiberio, maintained that as the tibiofemoral joint extends the transverse plane rotation must be furnished from a proximal source, this being internal rotation of the femur on the tibia. This would provide one convenient explanation for why the patella exerts an increased pressure on its lateral articulation with the femur. Tiberio’s theory is the explanation clinicians tend to relate to and underpins the basis of most conservative treatments

1.3. The Role of Foot Orthoses in Patellofemoral Pain

Pitman and Jack, (2000) suggested foot orthoses could be used as a first line treatment for patellofemoral pain, suggesting they could alter the functional Q-angle, however, this was not measured. Pain scales were used to confirm their hypothesis. Williams, et al. (2003) suggested that runners with excessive foot pronation (greater than 18°) demonstrated greater knee flexion and abduction, therefore, the target of foot orthoses was to realign the knee by altering foot placement. Powers, et al. (2012) noted that there is some evidence of foot orthoses being useful in the treatment of patellofemoral pain. However, it was suggested there was still a need to develop a model to detail exactly how foot function affects patellofemoral joint

function. Boldt, et al. (2013) noted that medially wedged orthoses are often prescribed to reduce retro-patellar stress by limiting calcaneal eversion and subsequent tibial rotation. Boldt, et al. did highlight that there were inconsistent results reported throughout the literature.

From clinical observations over many years it has been noted numerous young patients presenting with patellofemoral pain seem to demonstrate late stage pronation as the support limb enters single limb support, just prior to and during the propulsive phase of gait. This current work is aiming to uniquely demonstrate the action of foot orthoses on different sections of the foot throughout the gait cycle by using a three-segment foot model. The method proposed will not only allow comparison with previous barefoot studies but also measure the effects of wearing foot orthoses with and without shoes which will test the clinical hypothesis that supportive footwear is essential for orthotic therapy to be effective. This study aims to determine if these interventions are translated to the proximal joints in particular the knee. This will be the first time a multi-segment foot model will have been used to investigate what effect, if any, three quarter and full length foot orthoses have on knee joint function when walking and descending stairs. Full length wedging of the foot was used as it was felt this would provide greater control of foot position during step descent and the propulsive phase of walking. It is hoped the outcomes of this study could possibly inform current clinical practice by questioning the value of assessing patients barefoot, and influencing treatment plans and prescriptions of patients with patellofemoral pain who are deemed to require treatment with foot orthoses

1.4. Thesis Structure

To achieve the overall aims of this thesis it was necessary to conduct it as two separate studies, as such the structure was determined by the chronological order of the experiments:

Chapter 1 outlines the findings of previous work from patellofemoral studies and links this with foot function, to demonstrate how the present study will contribute to knowledge in this area.

Chapter 2 details the anatomy and function of the patellofemoral joint and the joints of the foot during walking. Step descent is then reviewed and foot motion during this task is discussed.

Chapter 3 considers how the foot has been measured previously, and looks how other authors have investigated footwear and orthoses. The latter half of this chapter examines how other authors have investigated the potential aetiologies of patellofemoral pain during both walking and step descent. The initial aims of the present study are stated.

Chapter 4 presents the methods used in the present study to compare both sandal and shoe data on a cohort of 15 normal subjects. The equipment specially developed for this work is detailed, before the procedures undertaken by the subjects are presented.

Chapter 5 displays the results of the normative data; comparing the effects of the shoes over the sandals, and highlights any systematic changes that the orthoses make during walking and step descent.

Chapter 6 discusses the results of the normal subjects, looking at kinematic results of the knee and the foot segments, plus the kinetics of the knee and ankle. A short initial conclusion section is included at the end.

Chapter 7 states the aims of the second experiment used in this work comparing the previous shoe data of the normal cohort with a group of patellofemoral pain

sufferers. A short method section details any differences from the initial part of this study.

Chapter 8 presents the mean results of the patellofemoral patients and compares these with the normal groups mean data, and highlights any effects due to the orthoses while walking and descending a step.

Chapter 9 discusses the results of patellofemoral pain subjects compared to the normal cohort; highlighting the differences between the groups and any changes in the kinematic and kinetic data produced by the orthoses.

Chapter 10 states the present studies limitations and considers where future research should be directed, before presenting the overall conclusions.

Chapter 2. Background

This chapter describes the anatomy and function of the patellofemoral joint and what is considered to be typical function of the knee. The functional characteristics of patellofemoral pain are described; highlighting that diagnosis can be difficult. Foot function is considered as one possible precipitating factor; hence this is a focus later in this thesis. The gait cycle and step descent cycle are described and the reasons for investigating step descent are indicated. The dearth of foot motion studies during step descent is detailed, highlighting the importance of further investigation.

2.1. Motion of the Patellofemoral Joint

The patella is described as a flat triangular bone which is the largest sesamoid bone in the body. It is situated in the quadriceps femoris tendon which is derived from rectus femoris and vastus intermedius. They insert into the superior border, the medial and lateral borders give rise to the vastus medialis and lateralis respectively. The apex on the inferior border provides the origin of the patella tendon (Fig.2.1.).

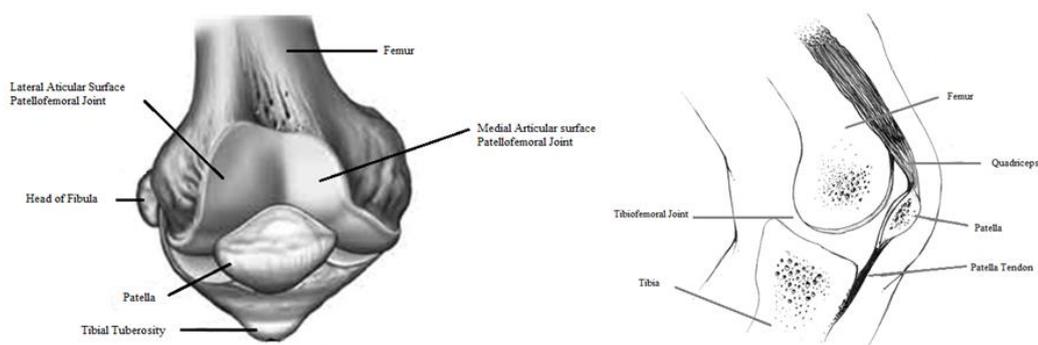


Fig.2.1 Coronal and sagittal plane views of patellofemoral joint
(Modified from <http://www.rahulpatel.net/knee-surgery/patello-femoral-instability.html>
<http://www.hopkinsortho.org/orthopedicsurgery/patellofemoral.html>)

The posterior surface of the patella is covered in thick hyaline cartilage which permits it to resist high compressive loads; there is a longitudinal ridge down the articular surface of the patella which splits it into the medial and lateral facets. The medial articular surface has a further division; the extreme medial border is sometimes referred to as the odd facet. Although the anatomy can vary between individuals, the lateral articular surface is usually larger. At full extension the patella

should mainly lie on the supratrochlea fat pad with only a small portion of the lateral articular surface in contact with the lateral femoral articulation. With flexion to 90° there is an associated internal rotation of the tibia which draws the patella in a medial direction. From 90° to 135° of flexion the ridge between the medial and odd facets contacts the lateral aspect of the medial condyle of the femur (Fig.2.2.). Beyond 135° it moves into the intracondylar notch, rotating and moving laterally, the odd facet then articulates with medial femoral condyle (Garth, 2001). Elliott and Diduch, (2001) reported that during flexion and extension of the knee the patella is piloted by active and passive soft tissue controls, these are arranged in a cross for the most effective anchorage.

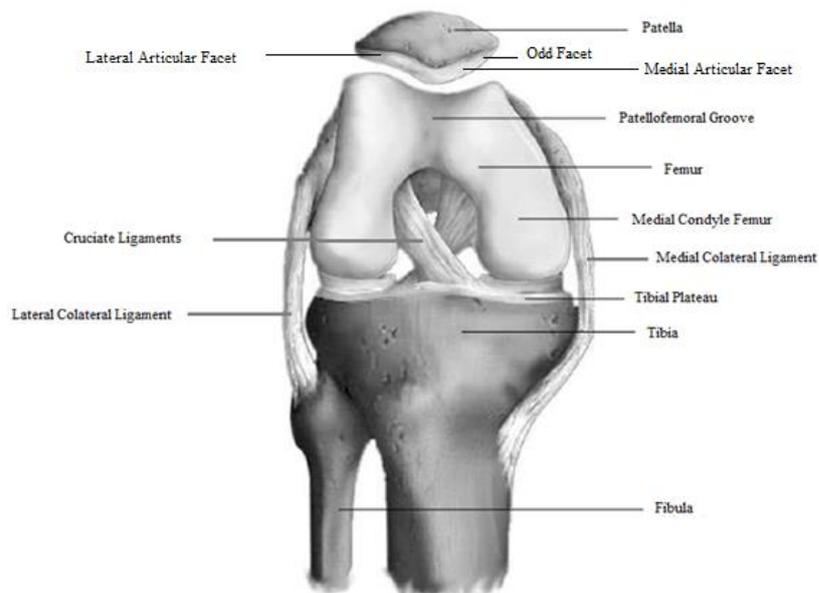


Fig.2.2. The knee flexed position (modified from <http://healthpages.org/anatomy-function/knee-joint-structure-function-problems>)

Heegaard, et al. (1995) described the kinematics of the patella in the cardinal planes; rotation in the sagittal plane was described as patellar flexion, while rotation within the coronal plane was called patellar tilt, motion in the transverse plane was termed patellar rotation. They noted that the patella also demonstrates translation in the sagittal plane and this was designated as patellar shift. Townsend, et al. (1977) stated that the patella was composed of cancellous bone that is arranged in sheets with connecting “struts”, they suggested the orientation of these sheets could contribute to the stiffness of the structure. They found that the direction of the sheets tended to

follow the direction of contact motion, though the structure under the central-medial facet was disordered.

Lafortune, et al. (1992) looked at five healthy male knees, using markers attached to intracortical pins, they reported that the tibiofemoral joint went through two internal rotations during the stance phase from heel strike and then again just prior to toe off, on both occasions it was reported as being less than 5°. Between these internal rotations the joint remained close to neutral, during swing phase the knee externally rotated to 9.4°(SD 2.7). They stated that the lack of external rotation during the stance phase is “logical” due to the action of the glutei producing an extensor moment this will also instigate an external rotation of the thigh, with the foot fixing the tibial segment; this gives a net internal rotation of the tibia. The authors noted that to insert pins into the bones is an invasive and “stressful” procedure. The necessity for infiltration of local anaesthetic at the insertion site alludes to the pain involved; probably altering the normal motion of the subjects.

Ramsey and Wretenberg, (1999) explained that due to the medial condyle being longer than the lateral; it induces a “helicoid” movement during flexion and extension which is often described as the “screw home mechanism”. It is usually accepted that knee extension is accompanied by external rotation of the tibia on the femur; this is reversed when the knee flexes, they suggested this provides increased stability over a simple hinge system.

Goldblatt and Richmond, (2003) suggested that successful treatment of knee pain is inextricably linked to knowing the anatomy and function of the structures that make up the joint. They noted that the tibiofemoral joint was the largest joint in the human body, and the condyles of the femur are “cam-shaped” when viewed laterally. They articulate with the corresponding surfaces of the tibial plateau, the menisci being present to improve the congruency of the joint. Goldblatt and Richmond, stated the medial condyle has a larger radius than the lateral and is generally larger in both anterior/posterior and proximal distal directions. They maintain this allows the medial condyle to rotate in all planes and translate, to a limited extent, on the tibial plateau; the lateral condyle is able to translate freely in the anterior/posterior

direction but can only rotate in the transverse plane when the knee is close to full extension.

This highlights the importance of understanding the anatomy of the tibiofemoral and patellofemoral articulations and why they are essential when considering the functional characteristics of the joints during gait and other daily activities.

2.2. Functional Characteristics of patellofemoral pain

Patellofemoral pain syndrome is characterised by pain in the retro-patella area or pain around the medial and lateral margins of the patella when undertaking activities that increase patellofemoral joint loading (Brechtler and Powers, 2002; Selfe, et al. 2007; Wilson and Davis, 2008; Barton, et al. 2011). Powers, (2003) noted that patellofemoral pain is commonly diagnosed in a wide variety of individuals; it is more common in physically active subjects. Patients have more problems with squatting, climbing and descending stairs (Callaghan, et al. 2009; Anderson and Herrington, 2003), or with prolonged sitting; Brechtler and Powers, (2002) used the term “movie-goers knee” for this latter problem. Many of the individuals seeking help are adolescents or young adults; there appears to be an increased incidence of female sufferers (Heng and Haw, 1996; Barton, et al. 2009).

Merchant, (1988) proposed a very detailed clinical classification system which separated patellofemoral disorders by their underlying cause such as trauma (single major or repetitive), or patellofemoral dysplasia, which then had further subcategories. He did suggest that chondromalacia should never be used as a diagnosis without further qualification and should only be used as a descriptive term for a lesion of the articular cartilage. In contrast Way, (1999) also recognised the shortcomings of diagnosis, leading to the statement that knowing the specific cause of the syndrome may not be critical to successful rehabilitation.

Elliott and Diduch, (2001) stated that historically therapy has been used for the non-operative treatment for patellofemoral pain; muscle length and strength were recognised as contributing factors. The use of taping and bracing was recognised as being useful treatments; similarly foot orthoses were also mentioned. Gross and Foxworth, (2003) noted that clinically foot function is often cited as a cause of patellofemoral pain but the experimental evidence for using foot orthoses was circumstantial at best. Laprade and Lee, (2005) stated that clinically it is usual to

undertake an assessment of both patellae, although it is often the case that one patella is more symptomatic than the other or in some cases the pain can be unilateral. By using this technique any asymmetry present may point the clinician to the best therapy for that particular patient.

Lowry, et al. (2008) suggested that there was no “gold standard” for the correct diagnosis of patellofemoral pain; they recognised that due to this there is no agreement on the most appropriate treatment. Callaghan, et al. (2009) suggests the syndrome presents in the absence of trauma or pathological disease and has a gradual build up prior to the patient complaining of symptoms. They stated that it has been generally accepted that pain or stiffness which is linked to certain activities has been used to aid diagnosis of the condition, however due to other knee conditions producing similar symptoms this may not be helpful; they concluded that patellofemoral pain syndrome is a “diagnosis of exclusion”.

Powers, et al. (2012) discussed both proximal and distal contributing factors; thirteen points were suggested for future investigations which were linked to foot function. They proposed that it is imperative for a model to be devised to demonstrate how altered foot function affects the patellofemoral joint. Powers, et al. suggested that further studies should be directed at investigation of midfoot. They stated that there is much work to be done to evaluate when orthoses should be prescribed, how they should be used and how they interact with footwear.

In a comprehensive review carried out by Mills, et al. (2012) peak rearfoot eversion, eversion velocity, lower limb rotations and shock attenuation were discussed. They suggested from their comparisons; orthoses that were contoured or moulded to the subject’s feet were more effective at reducing the loading rate and vertical impact; leading them to conclude future research needs to be directed at “neuromotor control effects”, there was no mention of any midfoot control within any of the studies they reviewed.

This section highlights that although the symptoms of patellofemoral pain are well recognised throughout the literature; there is still a debate on the best nomenclature. The description of the numerous underlying problems that may be responsible for this painful condition, may contribute to the lack of understanding, hampering a

functional definition of the underlying cause. One such problem is understanding the effect foot structure has on the proximal joints.

2.3. Foot Structure

Root, et al. (1977) suggested that joint morphology is genetically governed; this defines the axis of motion of a joint which enables the movement to be described relative to the cardinal body planes. They proposed that this morphology can be altered by “functional adaptation” which was explained as a response to a joint being forced to move in any direction which is not perpendicular to its axis.

Bruckner, (1987) suggested the foot was capable of being both a flexible adapting structure as well as a rigid lever capable of propelling the body forwards; this was attributed to a number of interactions including joint geometry; joint axes and soft tissue integrity in combination with ground reaction forces. This study concentrated on the anatomical variations of the subtalar joint; from the 32 bones that were studied 12 had 3 articular surfaces while the remainder had 2. The calcanei with only 2 subtalar joint facets had significantly greater surface area and smaller joint angles compared to the three facet specimens. It was hypothesised that subtalar joints with only two articulations may have lower subtalar joint axes and thus clinically may demonstrate a wider range of motion.

In 1999b Leardini, et al. undertook an experiment to establish what effects the ligaments have on passive motion of the ankle and subtalar joints. They did this by stripping soft tissues and baring the ligaments of seven amputated legs; these were then fixed to a “flexing rig”; intra cortical pin marker clusters were used to record the passive movement of each. They suggested that the joints have a “preferred path” and this is under the influence of the joint geometry and the ligaments; at the ankle some ligaments are there to guide passive motion while others are there to limit maximum position. The subtalar joint was different in the fact motion was only possible by the ligaments stretching and lengthening. Levinger, et al. (2010) stated that the human foot structure varies considerably between subjects, their foot posture being dependant on the alignment of the bones of the foot. They suggested that variation away from normal foot posture has been blamed in the literature for triggering lower limb injury.

2.4. Motion of the Foot Joints

2.4.1. Ankle Joint

The ankle or talocrural joint occurs between the distal end of the tibia and its malleolus, the fibula malleolus (clinically referred to as the ankle mortise) and the body and trochlear surface of the talus (Williams and Warwick, 1980) see Fig.2.3. Lundberg, et al. (1989) recognised that the talocrural joint is often described as a uni-axial hinge, demonstrating mainly sagittal plane motion; however when they measured eight healthy subjects with roentgen stereophotogrammetry, following the insertion of 0.8mm radio-opaque beads into the tibia talus, calcaneus and medial column bones of the right foot, they found in most subjects it actually demonstrated internal rotation with 30° to 10° of plantarflexion and then from 10° of plantarflexion to 30° of dorsiflexion it was accompanied by external rotation. They did state that there was considerable variation between subjects in the amount of rotation seen in the transverse plane. In the coronal plane they noted that there was a small amount of inversion exhibited with dorsiflexion from a maximum plantarflexed position into maximum dorsiflexion. When the leg was rotated the talocrural joint exhibited dorsiflexion which was contrary to the other joints of the foot, however this was the level at which most rotation was seen (Lundberg, et al. 1989c). Clinically the axis is orientated between the centres of the malleoli; externally rotated and tipped in a distal lateral direction (Czerniecki, 1988).

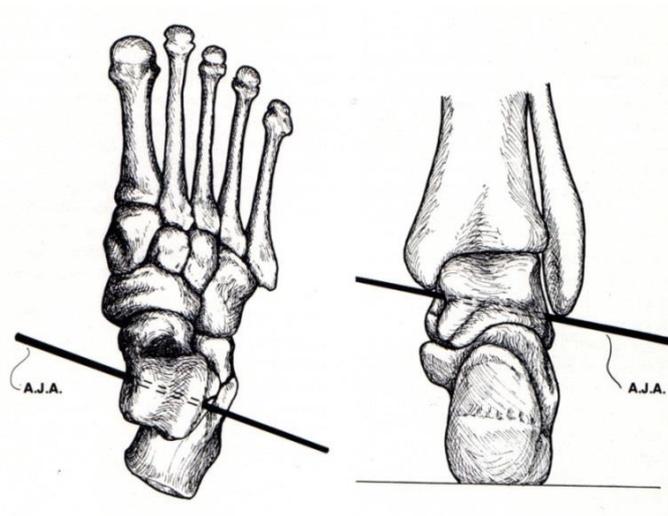


Fig.2.3. Diagram representing average ankle joint axis A.J.A. (Root, et al. 1977)

2.4.2. Subtalar joint

The subtalar or talocalcaneal joint is the articulation between the inferior surfaces (anterior, middle and posterior) of the talus and the superior surfaces of the calcaneus (Root, et al. 1977). The articulation is a modified multi-axial type (Warwick and Williams, 1980) its axis crosses all three body planes, therefore demonstrating tri-planar motion or pronation and supination (Fig.2.4.). Nester, (1998) stated that pronation is a single motion which combines abduction, eversion and dorsiflexion, while supination is a combination of adduction inversion and plantarflexion.

Classically the average axis for the subtalar joint was estimated to be 42° from the transverse plane and 16° from the sagittal (Root, et al. 1977), and was described as running from the posterior-lateral corner of the calcaneum to the dorso-medial aspect of the neck of the talus (Czerniecki, 1988).

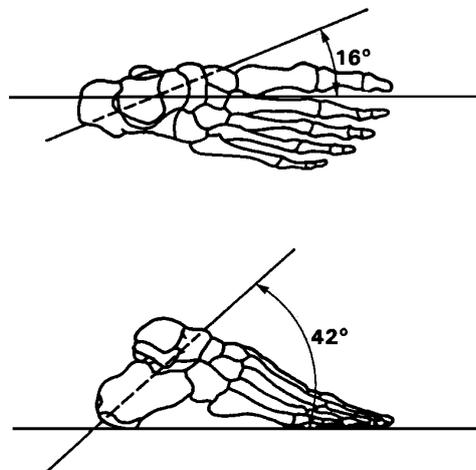


Fig.2.4. Diagram representing average subtalar joint axes (Czerniecki, 1988).

Motion at the subtalar joint and the ankle can allow the leg to rotate internally or externally while the heel is fixed on the floor. Rotation torque within the leg can instigate rearfoot motion; due to this action it has been termed a directional torque transmitter (Czerniecki, 1988). Lundberg, et al. (1989b) explained this process by recognising that the pelvis and leg demonstrates internal and external rotation during gait while the foot is generally fixed on the floor and it has therefore been assumed that this transverse plane motion is converted into pronation and supination of the foot. Scott and Winter, (1991) concurred with these statements noting that the ankle and subtalar joints are separated by the talus but the fact that it has no visible

landmarks makes it difficult to ascertain its position relative to the leg and calcaneus. Traditionally, when the foot is weight bearing, motion at the subtalar joint is termed “closed chain”, during supination the talus abducts, dorsiflexes and everts relative to the calcaneus, in closed chain pronation the opposite movements are seen (Nester, 1998).

Clinically the axis is classified as being high, low or normal, a high axis demonstrates less inversion and eversion, while a low axis exaggerates the coronal plane motions (Anthony, 1991). Kirby, (1987) acknowledged there is great variability in axis orientation from individual to individual; he recognised the axis can also be deviated medially. When this medial deviation occurs it increases pronatory torque from the forefoot, this is because the forefoot is on the lateral side of the axis producing a longer lever arm, it is therefore commonly seen in hyperpronated feet. Nester, (1998) also recognised the position and orientation of the subtalar joint axis is important, as these factors will influence the ability of the musculature to control subtalar joint motion.

The inclination angle of the subtalar joint has implications; Root, et al. (1977) suggested that when a foot with a high angle is moved in the coronal plane it will be associated with more rotation of the shank, conversely a foot with a low subtalar joint axis will demonstrate a wide range of coronal plane motion without the same scale of shank rotation. In 1998 Nester, returned to this phenomenon stating the orientation of the axis away from the transverse plane influences tibial rotation, the steeper the axis, the greater the amount of tibial rotation per degree of coronal plane foot motion. To complicate matters further it would also appear that as the subtalar joint moves the axis orientation changes; the axis was angled nearer the sagittal plane as the joint was supinated. Lundberg, et al. (1989b) investigated 8 feet being tilted from 20° of eversion to 20° of inversion resulting in a mean tibial rotation of 8.1° though they made no mention of inclination angle. In a later paper (Lundberg, et al. 1989c) 30° of tibial rotation yielded an average coronal plane motion of 6.1° though most of this was seen from when the tibia rotated from its neutral position into external rotation. They also noted that this was accompanied by plantarflexion and internal rotation of the foot.

2.4.3. Midtarsal Joint

The midtarsal joint is made up of two joints: the calcaneocuboid joint and the talonavicular joint. Williams and Warwick, (1980) described this articulation as multi-axial, producing a considerable range of gliding and rotation. Traditionally it is taught that the midtarsal joint has two triplanar axes: the long axis, which produces mainly inversion and eversion, transverse and sagittal plane motion being almost insignificant, and the oblique axis which provides adduction/abduction and plantarflexion/dorsiflexion seen when assessing the midfoot clinically (Root, et al. 1977).

Nester, et al. (2001) conducted a kinematic study of the midtarsal joint has thrown into doubt the validity of there being two axes able to function simultaneously at one joint. They suggest a single axis, its motion taking place through all three cardinal body planes; however they do admit this was only shown statically in stance and not dynamically in gait, though further investigations were recommended.

The fact that the midtarsal joint is made up of the talonavicular joint and the calcaneocuboid joint becomes significant when considering the stability of the foot. In 1960 Elftman, studied the axes of these joints and noticed that when the subtalar joint pronated they lined up allowing an increased amount of motion at the midtarsal joint, while supination lead to an obliquity of the axes and resulted in a locking of the midfoot. Lundberg, et al. (1989b) found when the foot was tilted from 20° of pronation to 20° of supination the talonavicular joint demonstrated continuous plantarflexion, it adducted in the transverse plane, while in the coronal plane talonavicular rotation exceeded the amount of motion demonstrated by the subtalar joint. Lundberg, et al. (1989c) recognised that in the foot and ankle there are a number of joints working together in close proximity which has led to problems in tracking actual joint motion.

2.5. Summary

Normal foot structure and joint motion needs to be defined in order to describe what is pathological. An understanding of the motion of each joint is relatively straight forward, however it becomes complex when considering the overall motion of the

foot during gait and other functional tasks. An appreciation of which joint is responsible for a given motion is imperative in order to describe the movement during such tasks.

2.6. Gait Cycle

It is customary for walking to be described in terms of the gait cycle; one complete gait cycle can be defined as the interval between two consecutive heel strikes of the same foot (Gage, 1990), Levine, et al. (2012) noted heel strike is the most convenient event but any point could be used. This stride can be divided further into stance phase and swing phase, during normal walking the stance phase takes up approximately 60% of the gait cycle (Gage, 1990; Abboud, 2002), however this is speed dependant (Richards, 2008. Page 54) The stance phase is further subdivided into the contact phase which starts as the heel strikes the ground to a point where the forefoot contacts the supporting surface; from this foot flat position to where the heel lifts is termed midstance, propulsion starts at heel lift and ends at toe off (Richards, 2008). On occasions some authors will split the propulsive phase into terminal stance and pre-swing; terminal stance is the period from the heel lifting until the heel of the swing limb hits the floor, pre-swing is from the point the contralateral limb contacts to toe off (Levine, et al, 2012).

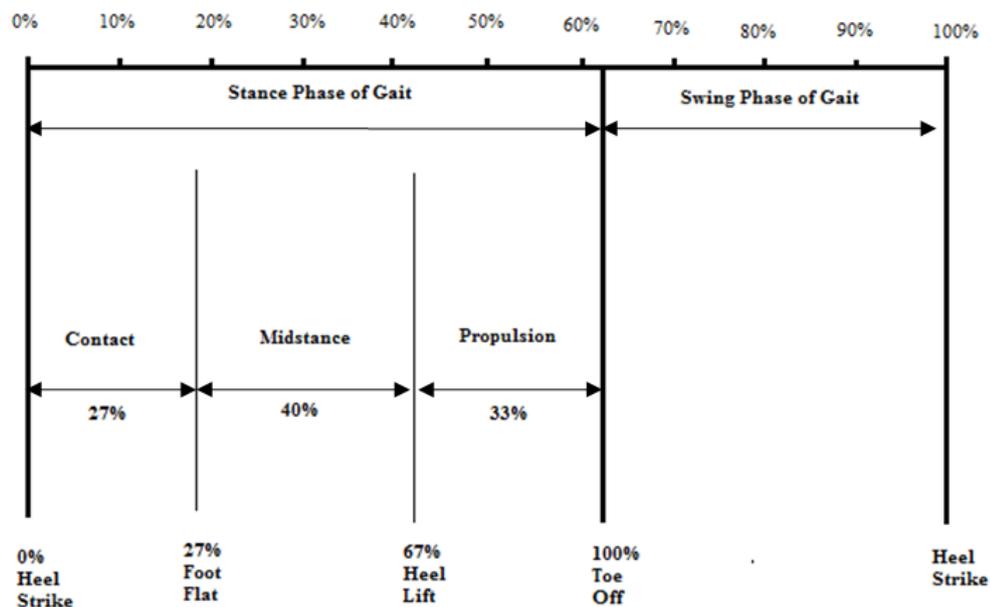


Fig.2.5. The gait cycle (adapted from Root, et al. 1977; Page 128)

Abboud, (2002) discussed that the foot is the only part of the body to contact the ground and as such provides support and balance when standing and has a major role in stabilizing the body when walking. Root, et al. (1977) suggested the foot pronates through the contact phase which is instigated by internal leg rotation; this action is slowed by eccentric action of tibialis posterior and other supinators of the foot until the forefoot loads smoothly on the supporting surface. At midstance the foot has to convert from an adaptive mobile structure into a rigid lever for propulsion, this is achieved by supination of the subtalar joint and is driven by external rotation of the leg and the action of the calf muscles. The propulsive phase still requires the subtalar joint to supinate which preserves the rigid properties of the foot and is aided by intrinsic muscle contraction. Root, et al. maintained there is a slight movement towards pronation just before toe off as the body weight is transferred from one limb to the other. Abboud, (2002) stated that if the foot is still pronated in the propulsive phase the plantar structures are put under an increased amount of tension, hence intrinsic muscle effort is directly linked to supination of the foot.

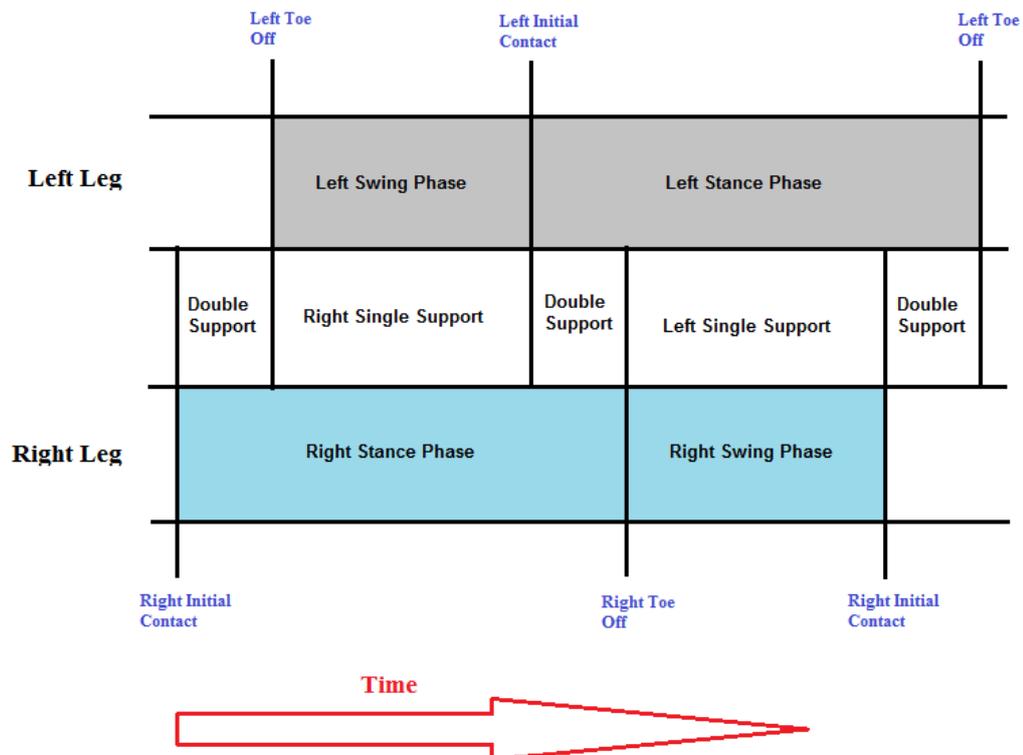


Fig.2.6. Gait cycle highlighting when single and double limb support phases are in effect (Levine, et al. 2012)

2.6.1. Step Descent

Andriacchi, et al. (1980) stated that the mechanics of stair use are very different from level walking and this is demonstrated by the changes seen in the range of joint motion, the maximum forces and moments acting on the joints plus the changes in the phasic muscle action. They found that during descent the hip of the stance limb remained close to full extension before flexing on average to 13.4° . The knee flexed to 68.9° just prior to toe off and therefore they suggested that it was under the control of the knee extensors; this was also the point at which the ankle dorsiflexed towards its maximum, they recorded a mean value of 24.7° .

Andriacchi, et al. (1980) stated the flexion moment of the knee was calculated to be three times greater when descending steps compared to walking over a level surface leading them to suggest this was the activity which induced the most stress on the patellofemoral joint. Maximum hip moment was seen during ascent but the ankles sagittal plane moments were similar whether ascending or descending. In the coronal plane they found that the hip adducted during the stance phase, the maximum adduction moment being produced during step descent to the floor. At the knee the maximum adduction moment occurred during descent to the floor, they found there was an inverting moment at the ankle for the whole of the stance phase. They did measure the transverse plane moments but they noted the results were low and rather variable, however they stated that there was a predominant internal moment at the hip and ankle and an external moment at the knee for both ascent and descent.

McFadyen and Winter, (1988) described the phases of descent as: weight acceptance as the foot first contacts the step; forward continuance and controlled lowering. They found during forward continuance there was a brief period where the muscles acted concentrically as the knee extended slightly (to a flexed position around 20°) this instigated forward progression and a slow rise of the body. It was proposed the controlled lowering phase was the major part of descent and was under the eccentric control of the quadriceps at the stance knee and soleus at the ankle for a shorter time. McFadyen and Winter, reported that hip power was minimal until the last part of the lowering phase, when there was a concentric contraction which was suggested was to aid in “pulling the leg off the step”.

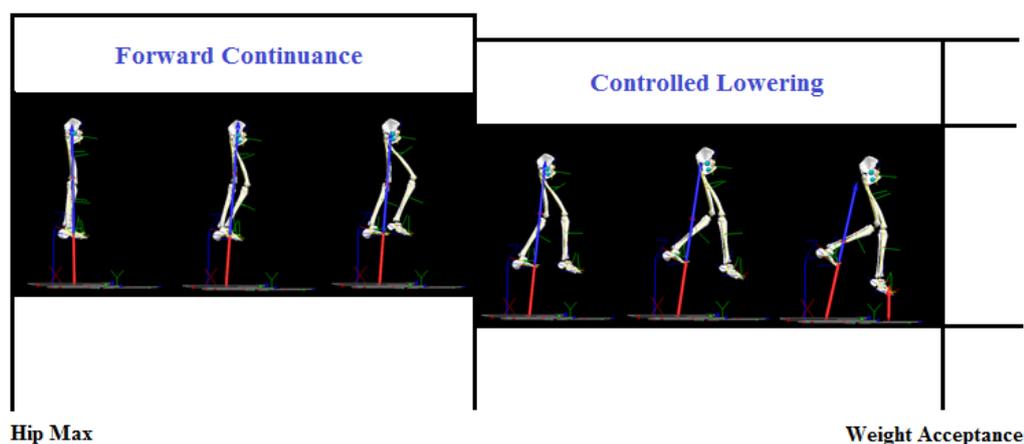


Fig.2.7. Step cycle depicting phases of descent described by McFadyen and Winter (1988)

Kowalk, et al. (1996) looked at the coronal knee moments when ascending and descending stairs in ten normal subjects. Their cohort produced greater maximum extension moments during descent rather than ascent, it was also noted that the maximum moment occurred much later in stance. The adduction moment demonstrated a similar biphasic pattern for both ascent and descent but statistically smaller than the extension moment. This led Kowalk, et al. to conclude that the sagittal plane moments are predominantly the most important, but the coronal moments cannot be ignored as they contribute to both medio-lateral stability and propulsion in ascent and decent. They rationalised this by highlighting the fact there were two periods - one in late stance during ascent and another early in stance during descent - where the extension moment was close to zero but the adduction moment was between 25 and 40Nm.

Riener, et al. (2002) hypothesised that the inclination of the staircase could affect both the kinematics and kinetics of ascent and descent. They constructed a set of steps that could be inclined from 24° to 42° this varied the height of the steps from 13.8cm to 22.5cm. They found as the stair inclination increased stance duration decreased. Although they concentrated on the impact of descent, the graphs they produced would suggest hip and knee flexion increased during the stance phase with the higher inclination but ankle dorsiflexion decreased. These alterations may not have been statistically significant as they only noted that the ankle remained in a

dorsiflexed position for most of the stance phase; they did not comment on this outcome.

In the UK staircases are categorised into three types by the building regulations; institutional, private and other stairs. The latter category is basically a building which does not fall into the other two categories. The maximum riser in a private house can be 220mm with a minimum 220mm going (42° pitch); while in a public place the maximum riser is 180mm with a minimum 280mm going. Flats and other buildings can have a maximum riser of 190mm.

When a group of subjects with anterior cruciate ligament deficiency was compared with a control group when ascending stairs, Thambyah, et al. (2004) found the external knee flexion moment was significantly reduced on the affected limbs as were the ground reaction forces, though the latter results were highlighted as small (0.1x body weight).

Protopapadki, et al. (2006) looked at 33 healthy subjects when ascending and descending steps; they recorded maximum hip flexion at 39.9° maximum knee flexion at 90.5° and maximum ankle dorsiflexion at 21.1° during descent. They found that there were higher flexion angles at the hip and knee on ascending (65.1° and 93.9° respectively) however maximum ankle dorsiflexion on ascent was only 11.2° almost half of that recorded for descent. In contrast to other studies (Andriacchi, et al. 1980, Kowalk, et al. 1996, McFadyen and Winter, 1988) Protopapadki, et al. found the mean maximum external knee extension moment was less in descent compared to ascent (0.40Nm/kg to 0.58Nm/kg) they suggested the variability between studies could be due to different subject heights, marker placement, and the methods used to calculate the joint moments. It was proposed that trunk position may have some influence over the lower limb moments and recommended this is an area that may require further research.

Selfe, et al. (2007) noted that step descent is mechanically more challenging than ascending; they explained this by recognising the centre of mass is carried forwards before the eccentric muscular action of the quadriceps has to resist the bodies mass plus gravity during the lowering phase. It was also observed that as the knee flexes

the external flexion moment increases and this needs to be balanced by increasing the eccentric muscle contraction. Age is of interest to researchers when considering stair descent due to the higher incidence of falls in individuals over the age of 65 years (Mian, et al. 2007). For this reason Mian, et al. conducted an experiment to compare the kinematics of 34 older adults (aged 69-82 years) with 23 younger adults (aged 20-29 years). After withdrawals, 14 older adults were placed in a group that received a generalised training program for muscle weakness and balance while another 14 were placed in a control group. A few differences were seen between the older and younger groups, the older subjects did descend at a slower speed and peak knee flexion was reduced. In the coronal plane pelvic obliquity was increased, the hip was more adducted during the latter half of the stance period. Transverse plane pelvic motion demonstrated no significant differences between the groups though the hip rotation in late stance was larger in the older group. The generalised training showed no significant changes; the authors accepted that the exercise program they used lacked specificity.

Novak and Brouwer, (2010) looked at sagittal and coronal plane moments in younger (mean age 23 ± 3 years) and older people (mean age 67 ± 8.2 years) during stair ascent and descent. They found cadence was group dependant, the younger adults demonstrated faster descent and shorter stance time; however side dominance was not a significant factor. They recognised that moments are influenced by speed and as such they performed their calculations with cadence as a covariate. It was reported that the eccentric plantarflexor moment was less in the older group but this was combined with a higher knee extensor moment and a small hip flexor moment. The authors proposed this increased the overall extensor support and could be a mechanism to increase stability. It was during descent that the coronal plane moments were greatest; the results demonstrated that the knee and hip abductor moments dominated, this was attributed to the body having to maintain the body's centre of mass within a small base during the controlled lowering phase.

It was noted by Whatling and Holt, (2010) that both stair ascent and descent had been regularly used to investigate a number of conditions due to its reduced variability over walking and the increased activity of muscles needed. Many experiments have been published including stability with age, ligament deficiency,

osteoarthritis, surgical interventions and fall prevention. They also stated that motion analysis had been employed to quantify normal knee function and the effect of step height. The main focus of their study was to investigate what effect the stair gait cycle had on the kinematics and kinetics of the knee during ascent and descent. Ten subjects walked up and down a four step staircase which were situated over a force plate, the 16cm high steps were moved between trials to allow the recording of ground reaction forces from different steps (either step one or step two- see Fig.2.8.). The stair gait cycles for the descent trials were: right foot leaving step three to the same foot leaving step one, and the right foot leaving stair four to the same foot leaving step two. The kinematics revealed no significant differences; however the moments were more sensitive depending which step they were measured from. Whatling and Holt, reported that there was a significant increase in the second flexion moment peak and the initial adduction moment peak for the stair gait cycle ending on step two compared to the cycle ending on step one. They suggested that it was important to consider these differences when designing an experiment and further studies are needed on the hip and ankle.

Richards, et al. (2010) proposed that descending stairs was more demanding than ascent and this was primarily due to the muscle control required to perform this task. It was suggested that the most unstable point of the step cycle is when the hip knee and ankle are flexed on the single support limb. They measured 6 pain free subjects walking, ascending and descending stairs; it was reported that the knee flexed more during ascent (110°) less during descent (106.4°) but maximum flexion was 70.6° during walking. Richards, et al. reported that the ankle also produced a greater range when on stairs; during walking their group recorded mean maximum dorsiflexion of 10.5° but ascent produced 15.9° while descent was 24.1° . They suggested for successful rehabilitation of patients it is necessary to understand the biomechanics of stair ascent and descent. Previously there has only been one study which has considered multi-segment foot motion during step descent (Rao, et al. 2009) this was completed bare foot, the present study will be the first to review foot motion during shod step descent.

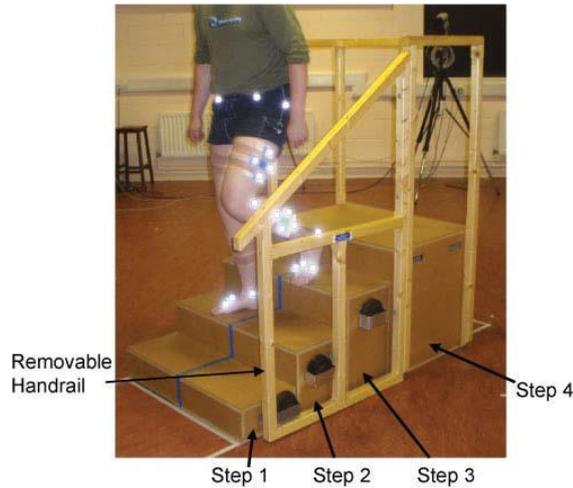


Fig. 2.8. Photograph of stairs used in Whatling and Holt, (2010) experiment positioned over a force plate.

Cluff, et al. (2011) proposed that stair descent poses little problems for healthy individuals but can be a hazardous task when motor function is impaired in the elderly or injured. It was stated that though many studies have looked at this task there have been a number of different methodologies used which makes comparisons between studies difficult at best. It was hypothesised that the step cycle would influence both the magnitude of the joint mechanics and a single step descent may not be enough to attain a “steady-state” of descent. Seventeen healthy subjects descended five steps ten times to obtain a complete step cycle for each limb at a self-selected pace. It was reported that progression velocity increased to step four where it plateaued; this was defined as “steady state”. Cluff, et al. highlighted that higher velocities in descent requires increased energy dissipation at contact, it was the ankle that demonstrated increased demand minimising any increases that were seen at the hip and knee. Previous studies have tended to only report on sagittal plane ankle joint function rather than foot function during step descent, hence there is a limited pool of knowledge to draw from.

2.6.2. Foot Motion during Step Descent

It is obvious from the literature that little kinematic data has been reported on foot motion when using a multi-segment foot model on stairs; presently the only study to consider this was Rao, et al. (2009) who were looking at individuals who had been diagnosed with midfoot arthritis. They stated their reasoning for looking at this task was due in part to the fact patients reported that stair descent gave them increased

pain, possibly due to the greater amounts of motion and the fact that stair descent is accepted as a more functionally challenging everyday activity as compared to walking. This declaration was as a result of ankle joint motion from previous studies. It was recognised no other studies had considered foot function during step descent. All subjects were measured barefoot (30 with midfoot arthritis and 20 control subjects) they reported that step descent required increased amounts of first metatarsophalangeal joint dorsiflexion, first ray plantarflexion, ankle dorsiflexion and forefoot abduction when compared to walking in both groups. The subjects with arthritis demonstrated greater metatarsal plantarflexion and rearfoot eversion compared to the control group during step descent, but no differences were seen between the groups during walking.

Aliberti, et al. (2010) also recognised that foot motion may be of importance when descending stairs but adopted a different approach; they recruited seventy four volunteers for their study, thirty of which were diagnosed with patellofemoral pain. Rather than looking at foot motion they used plantar pressure distribution to investigate differences between the groups during step descent. They hypothesised that the symptomatic cohort would have higher pressure and larger contact area on the medial border of the foot; the steps they used only had a riser of 16cm. The knee pain subjects did have a significantly larger contact area at both the rearfoot and the midfoot but the pressure time integral and the peak pressure did not elicit any significant differences between the groups during step descent. Aliberti, et al. highlighted that plantar pressure provides an indirect way of measuring foot rollover, but suggest the increased medial contact area may be associated with increased eversion of the rearfoot. However, this could not be attributed to be the cause or the consequence of the patellofemoral pain. It was suggested that modelling the foot as a single rigid segment may mask some of its flexibility during motor tasks.

2.7. Summary

It is of paramount importance that more studies are carried out to investigate segmental foot motion in both stair ascent and descent. In addition, no previous studies have looked at multi-segment foot motion when using shoes. Although some studies exist that have measured barefoot motion, these take no account of any stability imparted by shoes.

Chapter 3. Literature Review

This chapter discusses how gait analysis has developed, noting some of the shortcomings and difficulties in modelling the foot and lower limb during gait. A chronological list of some multi-segment foot models is presented summarizing the findings of each. The repeatability of different foot models is then described, highlighting how other authors have dealt with the problems of recording foot motion in footwear and how the use of foot orthoses has been undertaken in previous studies. This chapter then describes the signs, symptoms and aetiology of patellofemoral pain. The use of step descent is presented as a useful assessment due to the demanding nature of the activity for this disorder. Finally, previous studies, which have looked at how foot orthoses affect knee joint function, are presented.

3.1. Literature Review the Foot

In the clinical situation foot motion is assessed by patients being observed walking away and towards the clinician, either over ground or on a treadmill; this logically focuses the attention on the coronal plane. It was probably this assessment technique that was responsible for a number of studies focussing on 2D motion, positioning markers on the back of the shank and the heel modelled the rearfoot (Clarke, et al. 1983; Stacoff, et al. 1991; McPoil and Cornwall, 1994; Brown, et al. 1995; Messier, et al. 1995; Stell and Buckley, 1998). However McClay and Manal, (1998a) also proposed the low cost and the ease of use as contributing factors to its popularity. In another paper that year McClay and Manal, (1998b) stated measuring foot motion is not straight forward due to the difficulties in attaching external markers to the talus; this means standard motion analysis techniques cannot be employed.

Clarke, et al. (1983) acknowledged that sub-talar joint pronation was triplanar in nature. However, they argued dorsiflexion and abduction always accompanied calcaneal eversion and therefore by only measuring the coronal plane motion, the movement in the other planes could be predicted. Moseley, et al. (1996) noted 2D studies had been used to estimate 3D motion and therefore the displacements in the transverse plane had only been deduced and not measured. Soutas-Little, et al. (1987) suggested the 2D method only accurately described coronal plane motion when the foot and shank were in the vertical plane; this was corroborated by McClay

and Manal, (1998a) as they stated when the foot was not aligned in the sagittal plane or the transverse plane it magnified differences between the 2D and 3D data, due to the incorrect assumption that the frontal planes of the shank and rearfoot are orthogonal to the optical axis of the camera. McClay and Manal, demonstrated that abduction of the foot from the midline of the laboratory introduced significant differences between 2D and 3D in the measurement of propulsive phase eversion and time to peak eversion.

In 1990 Kabada, et al. employed a marker system that they hoped would be adopted by other clinical centres in an attempt to standardise 3D results. The attached markers modelled the foot as a single segment, combining the movements of the ankle, sub-talar, and mid-tarsal joints. Areblad, et al. (1990) suggested there were three ways of describing 3D rearfoot motion by measuring rotation around known axes if the axes can be established; by using clinical rotations in the sagittal, coronal and transverse planes, or by a non-anatomical description of rotation and translation around a screw axis.

Other methods have been employed when using 3D data to try and circumvent the problems of placing non-invasive markers. Mueller, et al. (1993) used navicular drop as a determinant for unshod pronation, while Nigg, et al. (1993) placed 3 markers over the lateral side of the foot and modelled them as a triangle, this showed foot eversion was related to internal rotation of the leg, but rather unexpectedly maximum eversion did not significantly influence arch height. It is generally accepted that transverse rotation of the shank is linked to pronation and supination of the foot, McCulloch, et al. (1993) stated there was a 1:1 relationship, Nester, et al. (2000) suggested shank rotation could be used to describe the motion at the ankle, sub-talar and mid-tarsal joints, which they termed the rearfoot complex. However experiments carried out by Holden, et al. (1997) using a percutaneous clamp and Reinschmidt, et al. (1997) using bone pins both suggested skin markers overestimated the total amount of transverse plane motion.

The most accurate way of measuring motion is to use bone pins (Fuller, et al. 1997; Karlsson and Tranberg, 1999; Ramsey and Wretenberg, 1999; McClay, et al. 2000) and this technique is held as the gold standard. However there are many ethical

considerations and this method is not suitable for most clinical studies with larger sample sizes, it is far more popular to use external markers attached to the skin with double sided tape. Though this is ethically more acceptable the movement between the skin and the underlying bone is well documented and termed “skin movement artefact” (Angeloni, et al. 1993; Reinschmidt, et al. 1997; Fuller, et al. 1997; Karlsson and Tranberg, 1999; Ramsey and Wretenberg, 1999; McClay, et al. 2000). This phenomenon obviously hinders accurate measurements of certain segments, Holden, et al. (1997) suggested that if the error size could be established, the soft tissue motion could be taken into account reducing errors to an acceptable limit. Ramsey and Wretenberg, (1999) noted it was during the recording of dynamic activities with increased impact that the greatest errors could be induced.

3.1.1. Foot Modelling

Scott and Winter, (1991) recognised that previous studies had mainly assumed that sagittal plane motion was furnished at the ankle, while motion in the other two planes was as a result of rotation at the sub-talar joint. They stated that both joints were known to be triplanar providing movement in all three planes simultaneously. From this Scott and Winter concluded that any motion in any plane was due to the combined motion at both joints, their study set out to try and isolate the kinematics and kinetics of the ankle and sub-talar joints, they modelled each joint as monocentric with a single degree of freedom, acknowledging that this may be an over simplification. From their three subjects they observed that positive rotation at one joint is often associated with a negative rotation at the other, though this was not always the case, they suggested that this may help maintain the foot in the plane of progression. At the ankle joint they reported 12°-20° of plantarflexion following heel contact then a gradual dorsiflexion of 20°-28° before a rapid plantarflexion of 22°-28° at push off. The subtalar joint everted by 10°-17° before 20% of stance, maximum inversion was reported at 12°-18°.

Kidder, et al, (1996) reported the lower limb had been modelled by defining the major segments and articulations, but the foot had been studied a “single rigid body”; there had been no attempt to track foot segments during either the stance or swing phases of gait. To achieve this they calibrated a small capture area (0.5m x 0.6m x 2.4m – WxHxL), and attached twelve 15.9mm retro-reflective markers (which

included a marker triad on the hallux) to a single subject's lower leg and foot to split the foot into three segments plus a fourth to depict the shank (see Fig. 3.1.). Each segment was defined using three markers; Kidder, et al. reported motion of the tibia to the laboratory axis, hindfoot to tibia, forefoot to hindfoot, and hallux to forefoot. Kidder, et al reported on motion of their segments in all three planes, they maintained that the rearfoot motion was similar to previous studies, but they took care to emphasize motion of the midfoot had not been reported previously in the coronal or transverse planes. The single subject used demonstrated less than 10° total range of motion in the sagittal plane, while in the coronal plane they suggested that the forefoot was in a varus position at contact, this diminished into midstance when a valgus position was achieved. This was maintained until toe off when there was a rapid rotation into varus. In the transverse plane it was reported the midfoot was in an adducted position through contact and midstance before moving rapidly towards neutral at toe off. No ranges of motion were reported but it was stated that collecting data from only one subject limited any clinical conclusions.

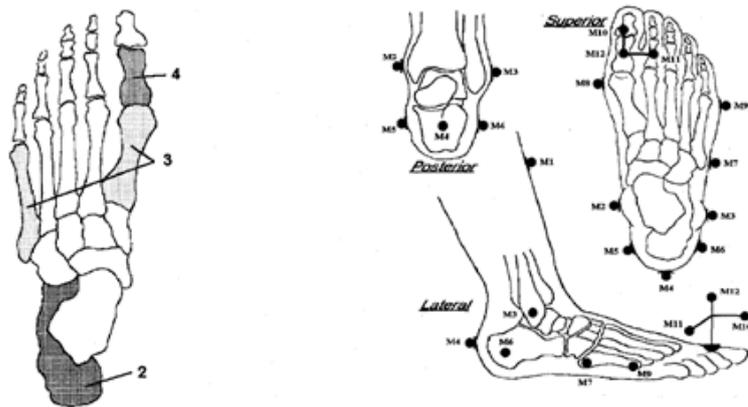


Fig.3.1. Diagrams taken from Kidder, et al. (1996) depicting foot segments and marker placement

In 1999a Leardini, et al. postulated it was imperative that during dynamic evaluation the shank and foot should be detailed as a “multi-joint mechanism” and proposed this may establish the origin of many foot and lower limb pathologies. They suggested at initial contact the motion of the ankle and sub-talar joint were paramount, but during early midstance it was the deformation of the mid-foot that would be of cardinal importance, while during propulsion they emphasised the

motion of the first metatarsophalangeal joint. The clinical relevance of dynamic measurement was also emphasised to distinguish between normal and pathological movement. To test their theory they recruited nine asymptomatic subjects and used a four segment foot model plus the shank. Leardini, et al. used four markers per segment but these were mounted on rigid Plexiglas plates, this modelled the calcaneus, midfoot, first metatarsal and hallux (see Fig. 3.2.). There was no attempt to use any markers to define the proximal and distal edges of the segments; instead sixteen anatomical landmarks were recorded in a static trial by using the tip of a pointer. The authors stated that this was a more accurate way of reconstructing the anatomical reference system.

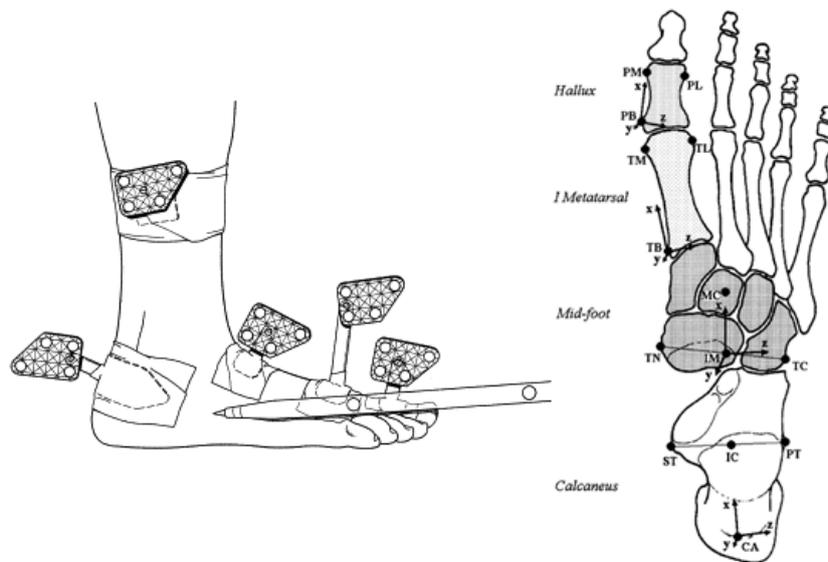


Fig.3.2. Diagrams taken from Leardini, et al. (1999a) showing Plexiglas marker plate positions with a representation of the pointer to define anatomical landmarks and foot segments

Leardini, et al. (1999a) reported the minimum and maximum angles (see Table 3.1.) for each segment plus the total range. They recorded that there were acceptable levels of repeatability between subjects when looking at sagittal plane motion of the calcaneal segment and the coronal and transverse plane motion between the first metatarsal and the hallux and slightly less when looking at the sagittal and coronal motion between the calcaneal and midfoot segments. No other repeatable patterns were seen; this led them to conclude that data comparisons between less repeatable joints must be “handled with great caution”.

In the same year Rattanaprasert, et al. (1999) undertook a comparison of 10 asymptomatic subjects with one who had dysfunctional tibialis posterior muscle; they compared the kinematics of the leg, rearfoot and forefoot and hence used a three segment foot plus the tibial segment, however, they only reported on rearfoot motion with respect to the leg and forefoot motion with respect to the rearfoot (see Table 3.1.). They used a total of ten markers, using three markers to define each segment. The patient demonstrated reduced rearfoot sagittal plane motion, in the coronal plane the range was similar but the patient demonstrated more inversion at heel strike. The midfoot mean sagittal plane range was more in the symptomatic subject over the normals but less in both the coronal and transverse planes. Rattanaprasert, et al. reported that most of the major differences between their normal group and the patient happened from just before heel lift toe off; they stated that this is the part of the gait cycle that needs a stable arch. It was highlighted that by modelling the rearfoot and midfoot separately the experiment added quantifiable results to the clinical diagnosis.

Carson, et al. (2001) stated that a foot model has to simplify the anatomy of the twenty eight bones of the foot; the aim of their study was to develop a three segment foot model (plus the leg) to evaluate the reliability of both the method and the model itself. Their segments included the hindfoot, forefoot and hallux, plus the shank (see Fig. 3.3. below) they again reported on the tibia with respect to the floor, and then each subsequent segment was reported with respect to its proximal neighbour. It was highlighted that the protocol was not invasive and was not dependent on x-rays which most previous studies had resorted to.

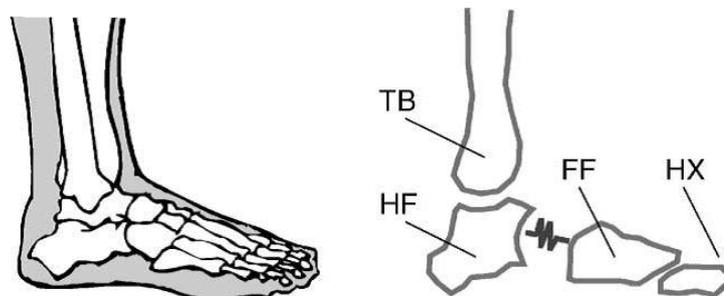


Fig. 3.3. Diagram from Carson, et al. (2001) depicting the segments of the foot.

The four segments in this model (Carson, et al. 2001) each used a ZXY embedded coordinate system, some of the anatomical markers used to define the segments were removed after the static trial and this was done to reduce the chances of skin movement artefact in areas where there was an increased risk of excessive skin motion. It was noted that three markers were required for tracking the movement of each segment (see Fig. 3.4. below). No mean data was submitted, however there were graphs published to show the motion traces of each segment; an estimation of the ranges seen in these traces is in Table 3.1. Carson, et al. concluded that this method produced acceptable repeatability when tracking “subtle patterns of foot motion”.

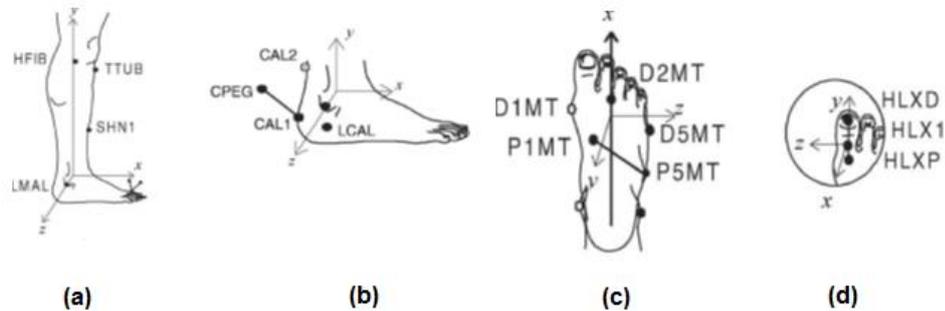


Fig.3.4. Diagrams taken from Carson, et al. (2001) showing marker names and placements plus how the segment coordinate systems are aligned.

In the same year Cowley, et al. (2001) also highlighted the fact that many previous studies had modelled the foot as a single segment only using two or three markers to determine ankle kinematics and kinetics. They detailed a nine segment foot and ankle model which they suggested would be more useful clinically to plan pre-operatively or determine the outcomes of surgery in more detail. MacWilliams, et al. (2003) also adopted this methodology and highlighted it is essential that multi-segment foot models are devised to establish what effects the mid foot and forefoot has higher up the limb and stated that simple single segment modelling can lead to incorrect assumptions being made. They used nineteen 10mm reflective markers to

create an eight segment foot model plus the shank on eighteen paediatric subjects. From their results they suggest that some foot joints are generating power while others are simultaneously absorbing power. It was also stated that motion of the foot excluding the rearfoot mainly happens in the sagittal plane; they did concede that not all articulations could be isolated some of which may be of great importance. It is obvious from Fig. 3.5. that not all segments had three markers to model them, therefore it would be impossible to model in six degrees of freedom..

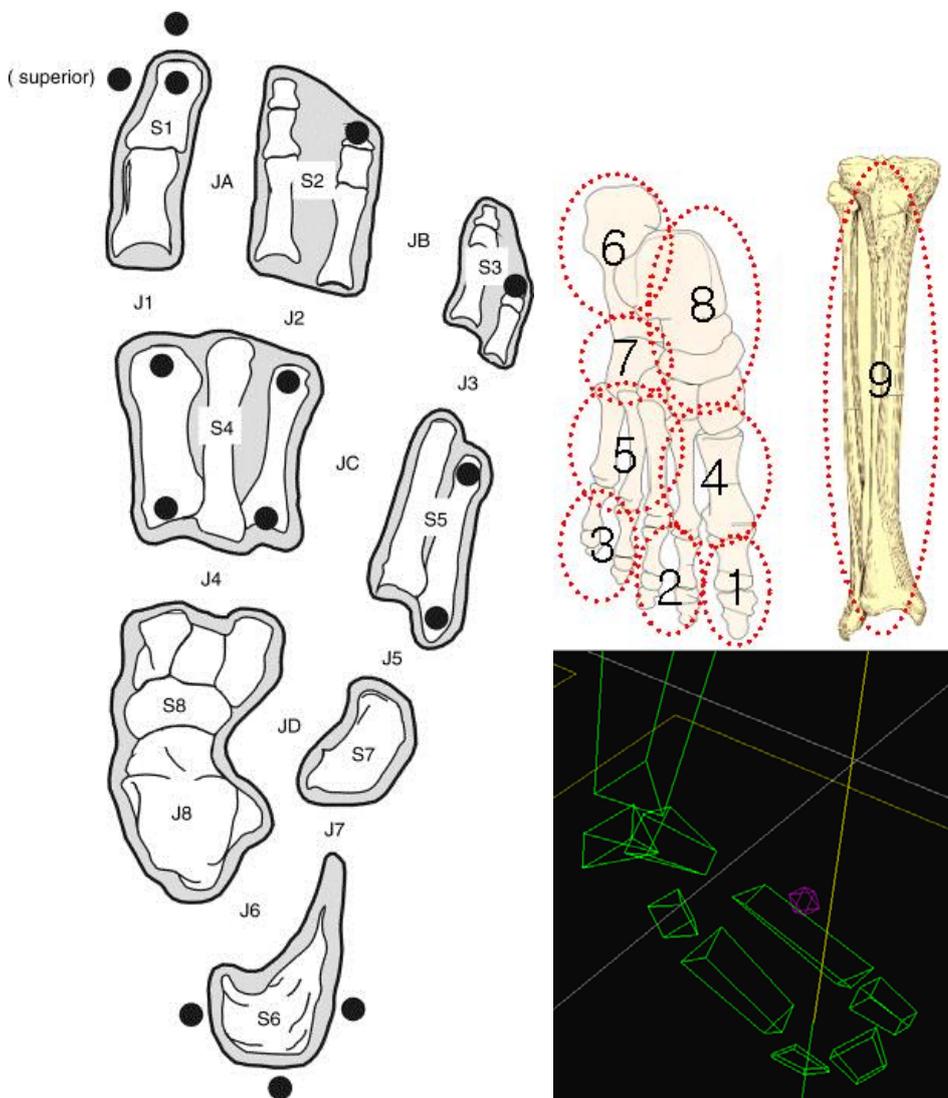


Fig.3.5. Diagrammatic representation of MacWilliams, et al. foot segments from Buczek, et al. (2006) and similar segmentation described by Hwang, et al. (2004) plus the modelling Bodybuilder™ (motion analysis software)

Hwang, et al. (2004) remarked that the foot was essential for dynamic balance, but many previous studies have only presented motion of the “ankle joint” in the cardinal planes. It was stated that most joints in the foot are not simple hinges but have multi-plane axes. Though the marker lay out was identical to Cowley, et al. (2001) and MacWilliams, et al. (2003) neither was cited (see Fig.3.5.). Interestingly Hwang, et al suggested they were modelling a talus segment as opposed to a combination of talus/navicular/cuneiform stated by the others, but there was no marker on that segment as it is recognised in the literature that there are no surface landmarks to attach to. Five normals’ had both kinematic and kinetic data recorded including EMG; graphs were presented for all segments in all planes but only the ankle and STJ joints were discussed in the sagittal and coronal planes and the lateral MP joints in the transverse plane. It was concluded that ankle joint power calculated from a single segment foot model may overestimate compared to a multi-segment model.

Table 3.1. Comparing mean angle data from different multi-segment foot model experiments.

Authors	Plane	Rearfoot Motion	Midfoot Motion	Forefoot/Hallux Motion					
Scott and Winter (1991) (rotation around a single axis)	Sagittal	30° ankle							
	Coronal	39° STJ							
	Transverse								
Kidder, et al. (1996) Only resolution reported approx. data from graphs of a single subject	Sagittal	13.0	10.0	35.0					
	Coronal	10.0	35.0	15.0					
	Transverse	20.0	15.0	8.0					
Leardini, et al. (1999)	Sagittal	12	8	11.5	25.6				
	Coronal	9.5	4.4	14.3	7.6				
	Transverse	10.7	2.8	6.9	15.7				
Rattanaprsert, et al. (1999) (mean Data 10 normals)	Sagittal	20.2	11.9						
	Coronal	13.7	10						
	Transverse	10.2	15						
Carson, et al. (2001) (Data from single subjects graphs)	Sagittal	14	16	50					
	Coronal	10	10	7					
	Transverse	8	5	5					
MacWilliams, et al. (2003) (Single paediatric subject graphs)	Segment	Ank	STJ	Calc	Med	Lat	1st	mid	5th
	Sagittal	12	5	7	15	8	45	40	35
	Coronal	6	5	5	3	3	10	8	8
	Transverse	8	10	5	8	3	7	5	8
Woodburn, et al. (2004) (5 Normals)	Sagittal	19.1(2.1)			12.3(2.0)		38.7(4.1)		
	Coronal	12.5(2.8)			10.0(4.5)				
	Transverse	12.9(2.6)			7.2(3.8)				
Simon, et al. (2006) (ROM mean 10 subjects SD= inter-day difference)	Sagittal	22.2(1.8)			42.1(1.1)				
	Coronal	10(0.3)			11.5(0.8)				
	Transverse				12(0.6)		3.2(0.5)		
Stebbins, et al. (2006) (ROM 15 Children)	Sagittal	24.2(2.7)			20.8(2.7)				
	Coronal	10.8(2.2)			8.8(1.6)				
	Transverse	12.3(2.7)			9.9(2.4)				
Wang, et al. (2010) (Control Group coronal plane calculated as only avg. reported)	Sagittal	21.9(4.9)			15.4(3.6)		27.9(5.0)		
	Coronal	10.9(5.0)			6.6(5.4)		19.2(6.5)		
	Transverse	9.3(2.6)			8.5(2.6)				

Woodburn, et al. (2004. Page 1918) suggested that traditional motion analysis techniques are unable to be adopted when measuring the motion of the small joints of the foot, they therefore highlighted that groups of bones had to be combined into “functional units”; they used a modified version of Carson, et al. (2001) model. Their study was to compare a normal cohort with a group of patients with rheumatoid arthritis (RA) it was suggested a multi-segment foot approach would help interpret which joints tend to be affected which may help guide clinical interventions. It was

found the RA patients had limited plantarflexion of the rearfoot segment in the pre-swing period; they also operated about an everted position in the coronal plane and were abducted in the transverse plane. Forefoot motion was found to be restricted compared to the normals in all three planes, the hallux was also restricted, Woodburn, et al also reported on navicular drop which was also reduced in the patients. They concluded this method gave a more intricate picture of how RA affects the smaller joints of the foot but that in the future it should be refined further.

In 2006 Baker and Robb, noted that it had only just become possible to track the foot markers with the rest of the lower limb, previously it had been necessary to use a different camera set up for each operation. Earlier the same year Simon, et al. (2006) recognised that the use of multi-segment foot models was still under development and as such they tried to develop a system that could be adopted by more centres in an attempt to standardise the results. The rearfoot was modelled as two anatomical hinge joints but chose to describe the forefoot and mid-foot motion from the chosen landmarks; they called this the Heidelberg foot measurement method. Baker and Robb, (2006) acknowledge this could be a good method of describing foot deformities but postulated that segment based models could better depict the underlying biomechanics.

Also in 2006 Stebbins, et al. suggested if the foot was modelled as a single segment this would not provide enough information to investigate treatments which were specific to the foot. They stated their specific interest was looking at paediatric feet with the future progression looking towards the investigation and quantification of pathological deformity; cerebral palsy was an interest cited. It was proposed this provided more problems due to the smaller surface area and greater variability between subjects. The model suggested by Carson, et al. (2001) was used as a basis but the tibia was redefined using the knee joint centre. The hindfoot was modelled independently of its adjoining segments which were judged to be important for looking at foot deformity. At the forefoot the hallux marker was placed medial to the extensor tendon leading to the anterior/posterior axis being altered. Five different variations were then recorded on fifteen healthy children; this was done by either eliminating specific markers or using different markers to track specific movements, determining the variability in the results.

Stebbins et al. concluded that marker placement was critical to produce a valid study, the model they recommended was named the Oxford foot model and it was said to produce results that were consistent with other studies carried out on adults. They found the sagittal plane was the most consistent while the transverse plane was the least repeatable, though eliminating the wand maker on the heel did improve this (see Fig. 3.6. below). It is noted in the Visual 3D™ online tutorial that the rearfoot definition is a little ambiguous as it could be interpreted in more than one way, therefore it is left up to the user to decide “the best definition for their needs”. The creators of Visual 3D also note the hallux is problematic as there is only one marker, in the original set out by Carson, et al. (2001) there was a marker triad on the big toe; this however seems to have been omitted in the “adoption”.

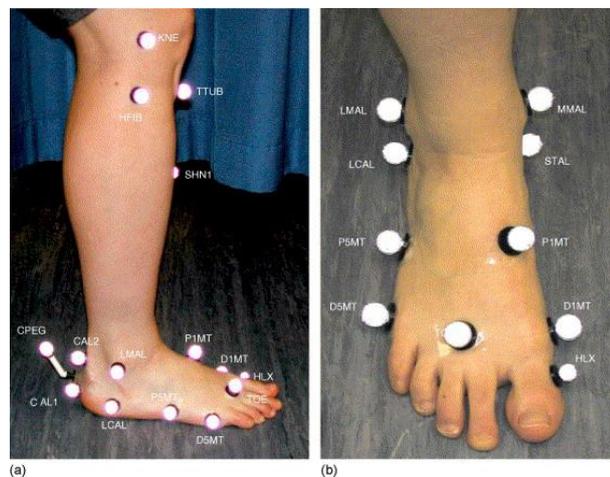


Fig.3.6. Photograph of marker placement Stebbins, et al. (2006)

Most authors agree that it is very important to establish the normal motion of the mid-foot and forefoot (Leardini, et al. 1999; Carson, et al. 2001; MacWilliams, et al. 2003; Woodburn, et al. 2004; Simon, et al. 2006; Stebbins, et al. 2006), however as yet there is little agreement on the best model. Leardini, et al. (2007), Woodburn, et al. (2004), and Jenkyn, and Nicol, (2007) suggested a 5 segment foot including the shank, Hunt, et al. (2001), Carson, et al. (2001) and Theologis, et al. (2003) preferred a four segment model. While Cowley, et al. (2001) and MacWilliams, et al. (2003) reported using nine segments, Buczek, et al. (2006) recognised the work done by MacWilliams, et al. but stated even though the foot had been split into multiple segments it was still possible for there to be substantial tissue forces between the

medial and lateral forefoot sections and at the calcaneocubiod joint which had not been investigated previously.

Lundgren, et al. (2007) highlighted that even though the foot was finally divided into several sections there was still opportunity for the bones to move within the segments, it was stated that this could miss some important motion at certain joints or even erroneously assign motion to an incorrect joint. They therefore drilled nine intra-cortical bone pins per foot with a marker triad on each, into six subjects, this was in an attempt to avoid skin movement artefact; however some bones still had to be grouped together in small segments. Lundgren, et al. reported on the movement of eleven joints including the distal end of the tibia and fibula though the usual arguments of pin impingement on tissue and normal motion while under local anaesthetic have to be taken into account. Nester, et al. (2009) investigated this phenomenon further suggesting the movements within the rigid body segments could mask some important joint movements, they proposed a three segments mid and forefoot model but stated that the model used in any study should be matched to the research being undertaken.

It was also noted by Caravaggi, et al. (2010a) modelling the foot as a single segment “oversimplifies” its function through gait, it was also recognised that there was a considerable number of multi-segment foot models that had been proposed over the preceding years. It was proposed that the lack of standardization between the models makes comparisons difficult at best, sagittal plane rotation at the rearfoot and first metatarsal were the only motions that demonstrated repeatability between models. It was highlighted that speed obviously affects how the segments of the foot function so that was the focus of their study, they used a four segment foot model on ten subjects, they found sagittal plane motion was affected by increased speed in two ways initially dorsiflexion was increased during the first half of stance then plantarflexion was increased during the latter half. In the coronal plane the calcaneal segment was more inverted on the shank from contact to propulsion when it everted while the midfoot was the mirror image, it being more everted at push off. This was presumed to instil more rigidity into the structure of the foot as forces were increased.

This section illustrates that there are a number of foot models which have good points, while others have less successful elements. This study set out to record foot motion within shoes using a modified Carson, et al. (2001) model. In the original model the hallux was recorded separately, however, in this study, there was no attempt to record the hallux as it could not be separated when confined within the toe box of a shoe.

3.1.2. Repeatability of Multi-segment Foot modelling

Caravaggi, et al. (2010b) were more concerned with intra-session and intra-examiner repeatability, they used two subjects the markers being located by four different examiners with different levels of experience. The walking trials were repeated within two to four weeks of the original. They found that inter trial variability was small, but the inter session variability was influenced by experience of the examiner; this was attributed to the very close proximity of the three markers on the calcaneus. Variations in the positions of these markers were recorded by taking photographs of the markers at all the trials; it was presumed that small deviations in marker position induced larger digression of the calcaneus reference frame; this having a knock on effect on the joint kinematics.

Saraswat, et al. (2012) were also concerned with validation of this technique; they stated that if a multi-segment foot model was to be used for clinical decision making there must be meticulous testing of its repeatability. They used a modified Shriners Hospitals for Children Greenville (mSHCG) foot model when considering pronated feet in children; this model split the foot into three segments and investigated inter- and intra-clinician variability. They found that their measurements on the subjects with pronated feet were statistically no worse than their control group, though repeated placement of the markers did introduce about 4° of error; this was irrespective of this being between day or between clinician.

Bishop, et al. (2012) recognised that during the preceding ten years foot and ankle kinematics had been better described with a number of “pioneering” bone pin studies and many multi-segment foot models. It was suggested that these high-tech investigations had led to a better understanding and quantification of foot function compared to the more traditional single segment foot models. Bishop, et al. reviewed

twenty six articles all of which had reported using various foot modelling, they looked for five standards within these articles: 1. Definition of marker placement and accuracy, 2. Definition of segments, 3. Definition of segment co-ordinate system, 4. Report joint parameters by defining joint rotations and degrees of freedom, 5. Reliability of joint kinematics. It was noted none of the articles satisfied all of the suggested standards; the two models most often used were the Oxford foot model (see Stebbins, et al. 2006) that was cited more than the original work of Carson, et al. (2001), and the Milwaukee foot model (see Kidder, et al. 1996).

Repeatability of foot modelling has been a concern of many authors. In an attempt to maintain the highest possible standards, all the markers in this study were applied by the same experienced clinician. It was also the intension of this study to look at range of motion, as well as maximum positions as these can minimise recording errors (see 3.2.6. clinical relevance, Wilken, et al. 2012) and possibly help expose abnormal functioning of the foot.

3.1.3. Measurement of Abnormal Foot Motion

Liu, et al. (1997) explained that the three-degrees of freedom model that describes movement in the three cardinal planes could be enough to describe normal foot motion but would be inadequate for pathological feet. They laid out that that the usual plantarflexion/dorsiflexion, inversion/eversion and abduction/adduction had to be considered in conjunction with compression/distraction, medial/lateral shift and anterior/posterior translation; therefore requiring a six degrees of freedom model. This view was echoed by Ramsey and Wretenberg, (1999) who stated that for “complete kinematic analysis” it was necessary to model joints in six degrees of freedom; they described this as three rotations and three translations.

Theologis, et al. (2003), investigated the outcome of surgery on congenital talipes equinovarus using a three segment model compared to a group of age matched normal. They recognised that previous studies had only used single segment models reporting on sagittal plane motion of the ankle and rotation of the whole foot with respect to the tibia; while they documented clinically it was known that stiffness within the foot is one of the major post-surgical problems. They found that the

talipes group had significantly greater movement between the hindfoot and forefoot which was surmised as increased dorsiflexion, there was also a trend towards reduced hindfoot dorsiflexion.

Jenkyn and Nicol, (2007) used rigid clusters with three markers each to define their four segment model; they acknowledged there was no “gold standard” method recognised for reporting 3D foot motion. By attaching the clusters to over the talonavicular joint and the lateral side of the calcaneus it was claimed that motion of the subtalar joint could be measured. They justified this model by arguing the medial and lateral split of the forefoot segment allowed them to report on coronal plane twisting (see Fig. 3.7. below) it was also stated that without a tarsal segment motion the adjacent proximal and distal segments could not be captured simultaneously, but there was no clear explanation offered as to why.

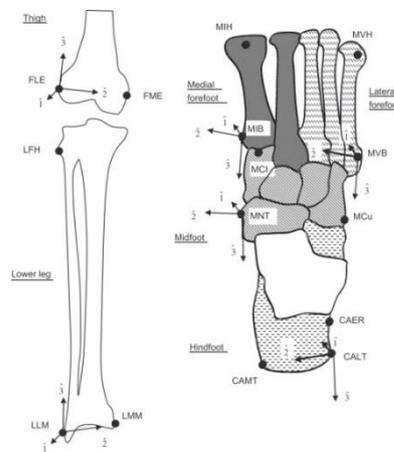


Fig.3.7. Jenkyn and Nichol, (2007) foot segments description.

Levinger, et al. (2010) used the Oxford foot model to compare the foot function of a group of 19 subjects, 10 with normal arched feet and 9 with flattened arches; they classified the foot posture with radiographic assessment. The group with the flattened arches exhibited significant greater rearfoot internal rotation with respect to the tibia and a trend towards greater rearfoot eversion in the coronal plane ($p=0.06$) though this was not significant. The forefoot showed significantly greater plantarflexion in late stance and greater abduction with respect to the rearfoot in the flat arched group. They concluded that their findings reinforced the clinical idea that flat arched feet are associated with greater pronation during gait.

Wang, et al. (2010) also used the Oxford foot model to investigate any alterations in barefoot function following operative repair on unilateral ankle fractures. They found that the injured side demonstrated less rearfoot plantarflexion, but there were no differences in the coronal or transverse planes, the midfoot of the affected sides showed significantly less transverse plane range but the sagittal and coronal planes were not altered. The hallux showed less dorsiflexion on the injured side, it was reported this was possibly a result of residual joint stiffness following the injury. Wang, et al concluded that this technique gave a quantitative value to post-operative outcomes which would detect small differences, this was adjudged to be an improvement over the normal clinical test of passive ankle range.

Hyslop, et al. (2010) looked at repeatability that measured the outcome of an intervention such as a drug used in the treatment of psoriatic arthritis; a before and after design has to be adopted. They used a seven segment foot model though they counted the shank as one segment and a single foot segment. They found low trial to trial variability but greater day to day disparity - this was attributed to marker placement altering the anatomical axes of the segments. It was also concluded that some of the segments demonstrating smaller ranges of motion seemed to be the least repeatable especially in the coronal and transverse planes. Tulchin, et al. (2010) reported that walking speed can affect sagittal plane motion in a multi-segment foot; they therefore argued that using a treadmill can easily provide governed speeds. Several significant differences were reported but dismissed as not clinically significant due to them being less than the inter-day differences.

Dubbeldam, et al. (2012) also utilised the Oxford foot model to look at coupling relationships between leg and foot segments. They looked at 14 healthy subjects walking and found that there were consistent movement affiliation between hindfoot inversion/eversion, forefoot abduction/adduction and hallux plantar/dorsiflexion. Dubbeldam, et al. reported a strong relationship between hindfoot inversion/eversion and hallux plantar/dorsiflexion, they noted this has been postulated before but not observed. This was attributed to the fact that previously only hindfoot eversion had been looked at. It was reported there was no consistent pattern between hindfoot inversion/eversion and rotation of the shank, it was noted that there was a strong

relationship between leg rotation and medial arch motion; again they noted this had not previously been reported. Dubbeldam, et al. did not find a consistent pattern of motion between the hindfoot and midfoot, they did note there were large individual variations and when the foot is planted on the floor independent coronal plane motion between these segments is possible.

Saraswat, et al. (2012) reported that the mSHCG model produced similar results when looking at ankle eversion and sagittal plane midfoot motion between pronated and control groups. However, the Oxford model did not find any differences between the groups in the ankle sagittal plane whereas the mSHCG model reported a reduced amount of motion in the pronating group; this was attributed to the Achilles tendon being tight - a common finding in pronated feet. They also reported further differences between the models. The Oxford model produced greater transverse plane rotation in the pronating group which was not found with the mSHCG model. The latter model was reported to have smaller standard deviations which it was suggested may indicate improved repeatability, though the Oxford study looked at thirty eight feet (10 normal and 9 pronated subjects) whereas the mSHCG study used ten subjects (20 feet) five in each group, which could easily account for the greater difference seen.

It is essential to be able to record foot motion accurately. Small abnormalities of foot motion have been a preoccupation of many studies due to the number of lower limb pathologies being attributed to excessive pronation. However, it must be noted that excessive supination is also thought to contribute other pathologies.

The foot models used in the previous studies have been developed to record bare foot motion (Table 3.1.) most rely on palpating anatomical landmarks to place the markers, Richards, (2008) noted this would make studying the effect of footwear and orthoses “impossible”, he suggested that the Carson, (2001) model could be modified to produce a three segment foot model which allowed placement of the markers on the shoe. It was highlighted that this model could not predict the foot/shoe movement artefact but it would allow modelling in 6 degrees of freedom. Because the present study was primarily investigating the effects of foot orthoses this method of foot modelling was adopted.

3.1.4. Footwear

There are a number of studies that have looked at the effect of footwear on foot function (Mann, et al. 1981; Clarke, et al. 1983; Stacoff, et al. 1991) though due to the limitations of modelling the foot as a single segment they were only able to focus on rearfoot movement. Mann, et al. (1981) suggested the term “heel counter support” to describe a shoe's ability to resist excessive contact phase pronation, while Clarke, et al. (1983) adopted the term “rearfoot control” for the same phenomenon though they were actually measuring heel eversion. Stacoff, et al. (1991) stated that footwear could contribute to injuries in runners and that the torsional stiffness of the shoe could be important in decreasing excessive pronation.

It is well recognised that there is potential for the foot to move within the shoe (Clarke, et al. 1983; Brown, et al. 1995; Milani and Hennig, 2000) a point not ignored by Stacoff, et al. (1991) who suggested that even though experimental verification had been limited, the fact that many runners get blisters highlights the fact there is friction between the foot and the interior of the shoe, they suggested that not only could this be a problem of slippage but also torsional movements. Reinschmidt, et al. (1997) compared bone pins with surface mounted markers to try and establish the range of skin movement artefact during walking (see Fig. 3.8.). They suggested that movement of the shoe with respect to the heel could induce errors of up to 7° though they admitted to have no way of separating the errors of skin movement or shoe slippage, and further discrepancies could have been induced by the cutting of windows in the heel counter to allow the accommodation of the bone pins.

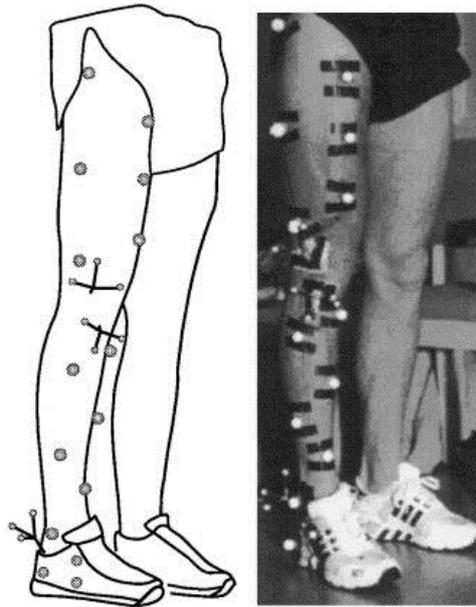


Fig.3.8. Picture from Reinschmidt, et al. (1997) showing external marker and bone pin placement to accommodate the calcaneus pin the lateral part of the Achilles tendon protector was removed plus a 12mm semi-circular cut out of the heel counter.

In 1999 Oeffinger, et al. used a single foot segment model with a cohort of fourteen “able-bodied” children with and without shoes. They noted a significant reduction in foot rotation with the shoes at midstance and mid-swing in the transverse plane and a small decrease in knee flexion and ankle plantarflexion in the sagittal plane, Oeffinger, et al. highlighted that there were no coronal plane changes seen and a significant increase of stride length was discussed in detail. They suggested that shoes have minimal effects on the normal gait of children; the statistically significant changes may not be clinically relevant

The majority of studies to date using multi-segment foot models have only considered barefoot walking due to the intricacies of applying the markers. This method cannot be used to investigate the effects of orthoses, or resolve what influence footwear has on the function of the foot. Wolf, et al. (2006) recognised this and used a relatively simple multi-segment foot model (see Fig. 3.9) to examine foot motion in eighteen children, to determine if footwear were inducing foot deformities. They used flexible and “normal” shoes, dividing the foot into three segments; windows were cut into the shoes to accommodate the markers. Wolf, et al. reported the shoes significantly increased ankle joint dorsiflexion and decreased

medial arch length, hallux flexion, foot torsion and forefoot width. Wolf, et al. concluded that it was possible to measure the effect of shoes on the foot; commercial shoes significantly influenced the motion of the forefoot, however the eight markers they used to define the three segments did not allow them to model in six degrees of freedom.

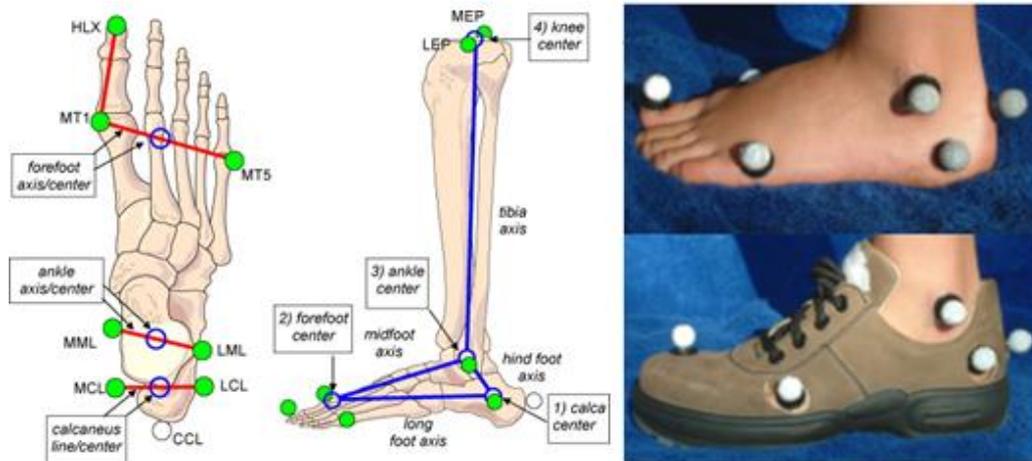


Fig.3.9. Pictures from Wolf, et al. (2006) detailing how the foot and lower leg was modelled and marker placement on both the foot and shoe.

Shultz and Jenkyn, (2011) noted there had been a number of attempts to use modified shoes or sandals to record the effect of footwear on the function of the foot, but due to the open nature of the footwear, to allow attachment of the tracking markers, they could not be representative of a supportive running shoe. They recognised that some other studies had cut holes in shoes to allow for the fitting of markers or allowing bone pins to protrude, but there had been no research to determine if this practice compromised the structural integrity of the shoe, or indeed at what point this happens. Using a single subject design they looked at three shoe types: a stability shoe; a motion control shoe and a cushioning shoe. They looked at oval holes cut into the shoes from 2.3x1.9cm to 3.5x2.8cm in four steps. It was reported that a hole of less than 1.7x2.5cm did not significantly alter any of the shoes function (though this size hole was not listed in the table of hole sizes) and that the maximum diameter holes listed previously did not affect the motion control shoe which was by definition the strongest shoe. Shultz and Jenkyn, concluded that the size of the hole that will not affect the integrity of the shoe will be dependent on the robustness of the shoe. It was stated that having the markers attached to the skin

must more accurately represent foot motion than shoe markers however skin movement artefact will still play a part.

Bishop, et al. (2013) noted that most multi-segment foot models had been developed to specifically study barefoot movement, it was highlighted that most daily activities are conducted while wearing shoes. They recognised that some previous studies had attempted to fix markers to the skin through modified shoes and that applying markers to the outside of a shoe may not strictly represent the movement of the foot within the shoe. They suggested that maybe shoe segments could be used to model the “foot-shoe complex”; they noted that this was a logical step but no palpable anatomical landmarks would be available to define the segments. A heel counter, mid-sole and toe box model was suggested, with a proviso that there could be no guarantee that the anatomical frame would relate to the bone embedded frame.

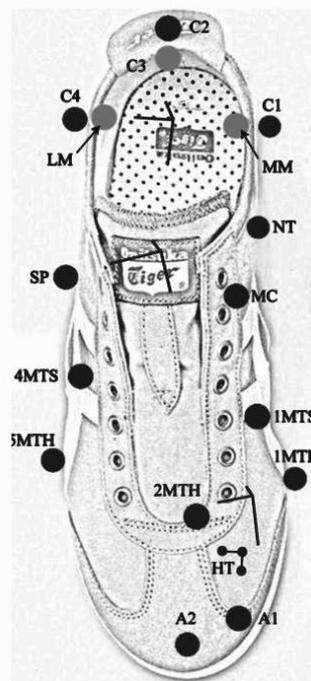


Fig.2.10. Diagram taken from Bishop, et al. (2013) demonstrating the marker positions of the “Foot-shoe complex”

To test their hypothesis Bishop, et al. (2013) conducted three experiments to test reliability, accuracy and the models minimal detectable difference. The first experiment used twelve subjects with two investigators, one with more than five years’ experience the other with less than five years. Each investigator applied the

marker set seen in Fig. 2.10. to each subject twice, a week apart. It was reported that for the recorded static trials the intra-rater reliability was better for the more experienced investigator and that the mean difference in marker placement was less than 8.0mm; this was described as good to excellent.

The second experiment used twenty participants. A marker set was applied to the right foot of each subject by the experienced investigator; this was then x-rayed using medio-lateral and anterior-posterior views. Those markers were then removed before a similar set was applied to a shoe; the x-rays were then repeated to compare the accuracy of the marker placement. The accuracy of the shoe marker placement was ≥ 6.7 mm when looking at all planes, the greatest variability was seen over the toe box, this was explained due to the fact the toe box does not touch the foot at this point.

Bishop, et al. (2013) noted from the final part of their experiment that the minimal detectable difference was $< 5^\circ$ in some instances, but this led to them to question if the foot-shoe model is able to detect small changes in joint kinematics, as any results will be prone to motion of the vamp material, tissue artefact and measurement noise. It was concluded that by placing the markers on the shoe this does not represent foot motion but movement of the foot-shoe complex and this will differ with the type of shoe that is used. It was noted that further studies are required to describe how the foot and shoe surface interact together during differing tasks.

This study set out to investigate the effect foot orthoses have on a group of patellofemoral pain patients with pronated feet. As clinicians will tend to recommend and fit these devices into supportive footwear, it seemed essential that this advice was followed in the experiment otherwise it would fail to inform clinical practise.

3.1.5. Foot Orthoses

Foot orthoses are regularly prescribed for numerous lower limb problems including patellofemoral pain associated with walking and running, however it is well documented in the literature that the mechanisms by which they work are poorly understood (Kilmartin and Wallace, 1994; MacLearn, et al. 2006; Chen, et al. 2010),

investigations into how they affect subjects are hampered by the fact that the response to a particular prescription can be specific to one person (Payne, et al. 2002) this could disguise any trends within the data recorded.

Foot orthoses can be classified in different ways, sometimes they are described due to the properties of the materials used i.e. soft, semi-rigid or rigid, or to the type of procedure used to construct the appliance i.e. moulded or non-moulded (Lockard, 1988). There are varying definitions of what foot orthoses do, Anthony (1991. Page 142) suggested they are “orthopaedic devices designed to promote the structural integrity of the foot and lower limb,” while Root, et al. (1977. Page 73) advocated that they “assist in controlling foot geometry and force direction stabilising joints and reducing muscle contractions.” Most orthoses are used to control excessive pronation (McCulloch, et al. 1993) but it must be noted there are also anti-supinatory orthoses (Nester, et al. 2001; Kilmartin and Wallace, 1994).

In their review Kilmartin and Wallace, (1994) concluded that foot orthoses are clinically useful in reducing lower limb pathologies but the way in which they work is still a point for debate. Chen, et al. (2010) stated that flatfoot deformities are often treated with some form of orthotic device in an attempt to increase stability and restore the arch of the foot. They noted that the use of foot orthoses to control flat feet was widespread, but reiterated that the “biomechanical effects” of this intervention is still not understood. To investigate the alterations produced by shoes and shoes and orthoses they placed markers (Helen Hayes marker set) on the feet of eleven excessively pronated subjects and then fitted a similar marker set to the vamp (uppers) of bespoke shoes they had fabricated for the subjects. It was reported peak knee flexion and peak ankle dorsiflexion were significantly increased, while peak ankle plantarflexion was significantly reduced when wearing both shoes and shoes with insoles compared to walking barefoot. Chen, et al. were more concerned with the overall effect that foot orthoses had on the proximal joints of the lower limb, they proposed that many orthoses studies only focus on foot function and as a result may miss the overall function of the lower limb. They recorded no significant differences in the kinetic data at the hip, knee or ankle between the three conditions, it was highlighted that coronal plane motion of the knee and ankle was not significantly

altered, it was also reported that stride length and support time were significantly increased in both shod conditions compared to barefoot walking.

It was suggested by Stacoff, et al. (2000) that externally mounted markers could overestimate the movement of the underlying bone, they therefore used two bone pins on five subjects; one in the lateral tibial condyle and the other in the lateral side of the calcaneus. There were three markers were glued to the shoe to model the rest of the foot, the heel counter of the shoe had to be cut away so as not to impinge on the distal bone pin. They then tested two types of orthoses; a medial wedge on the heel and a standard arch support, while the subjects ran down the laboratory. It was found the orthoses made small but significant reductions in tibial rotation (four out of the five subjects) but maximum eversion was not significantly reduced even though the same four subjects demonstrated some reduction. The shoe marker movement to the tibia tended to show a larger range compared with the bone pin data. They concluded that medial foot orthoses did not substantially change tibio-calcaneal motion, but differences between individuals were larger than between the orthoses, they hypothesised that mid and fore-foot motion may hold the key to understanding orthotic effect. This view was echoed by Church, et al. (2006) they drew attention to the fact that single segment foot models would probably “misrepresent ankle joint function” and can only provide limited information when considering foot deformity, their study looking at pronated feet found they demonstrated greater dorsiflexion of the medial forefoot. They concluded that this movement would not be seen in the single segment model.

Zifchock and Davis, (2008) recognised that looking at single segment foot motion may not be representative of what is actually happening. They therefore set out to establish what happens to the rearfoot when custom and semi-custom orthoses are introduced into shoes of subjects with low and high arches. To achieve this they cut holes in the heel counters of the shoes used in the tests, to allow the markers to be placed on the heels of the individuals (Fig.3.11.).



Fig. 3.11. Photograph from Zifchock and Davis, (2008) showing marker placement and windows cut in shoes.

Zifchock and Davis, (2008) did not elucidate how they removed the shoes around the makers, they did report that eversion velocity and excursion were significantly reduced by both orthoses, but peak eversion angle was unchanged.

Most previous studies that have looked at the effect of orthoses have tended to use a single segment foot model mainly due to limitations in technology. Now it is possible to divide the foot into smaller segments it would seem a logical progression to apply this technique to investigating the effects of orthoses. The problems arise when considering how to hold the orthoses onto the feet, to date few studies have attempted to report the effects of orthoses within the different phases of gait due to this obstacle. Cobb, et al. (2011) looked at sixteen asymptomatic subjects with low mobile foot posture, they used a four segment foot model plus the leg and referred to each segment as: 1. rearfoot complex 2. calcaneo-navicular complex 3. medial forefoot and 4. first metatarsophalangeal complex (see Fig.3.12. for marker placement) each movement was described between the distal segment with respect to its adjacent proximal segment; however rotation of the leg was defined from the calcaneus. The orthoses were fitted into a sandal using double-sided tape. The static trial for modeling the leg/foot was conducted with the subject sitting. The calibrated

area was 0.5m x 0.4m x 0.9m for the dynamic trials, it was stated that they aimed to use five trials for each subject but due to marker “drop out” on some subjects, only 3 trials were used. From the photographs in Fig.3.12. it would appear that some of the midfoot markers would impinge on the distal strap. There was no mention of how this was avoided in the text; they did state that the lateral forefoot was not presented due to “reconstruction difficulties”.

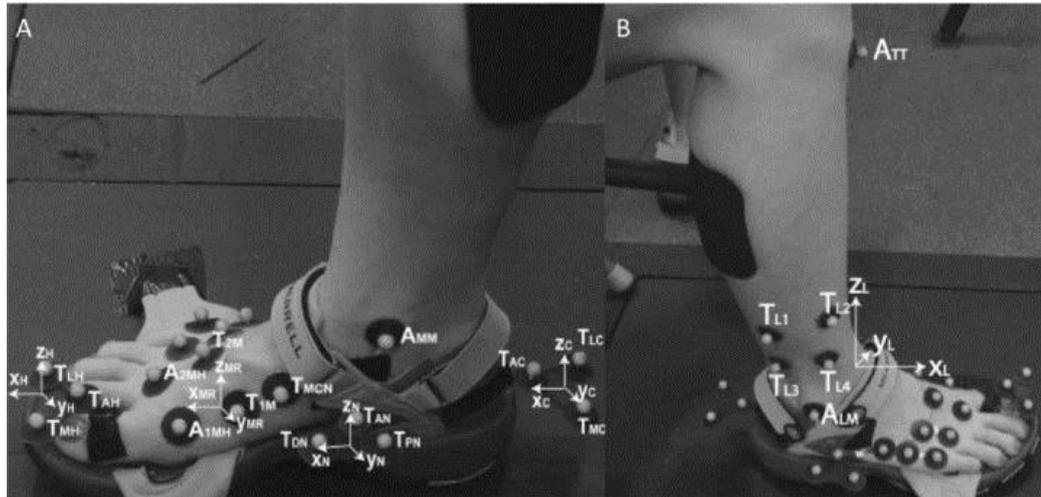


Fig. 3.12. Photographs of medial and lateral marker placement from Cobb, et al. (2011)

Cobb, et al. (2011) randomly assigned two different types of orthoses to their subjects; they were asked to use them for two weeks prior to testing, they reported their results for each of the functional articulations during each of the phases of stance. No angular data was reported, but rather the results of repeated-measures analysis of variance (ANOVA). It was reported that there was a greater rearfoot dorsiflexion displacement during midstance with the orthoses than without. There were no significant differences seen in the medial forefoot in any phase, but there was a difference found between the orthoses for the first metatarsophalangeal complex during terminal stance, though this was not repeated when looking at the orthoses condition compared to the no orthoses condition. Cobb, et al. did not discuss the issue of footwear aiding the support of the foot and that it is not a regular clinical practice to fit orthoses into sandals, though it may be done on occasions where there is no choice. They did however postulate there may have been more significant results if the structures had been fatigued, it was also noted that there was

no way of knowing if using the orthoses for two weeks prior to the testing had altered the “no-orthoses” trials in some way.

Only two other studies have attempted to report on any changes in multi-segment foot motion when using orthoses. Ferber and Benson, (2011) stated that many foot orthoses consist of or include an arch support suggesting that this minimises strain on the plantar structures in some way. They took twenty healthy subjects with a normal arch height index and looked at three footwear conditions: 1. Footwear only 2. Non-moulded orthoses 3. Moulded orthoses. To allow the markers to be attached to the skin there were holes cut into the shoes at strategic points (see Fig. 3.13.) the investigators drew round the bases of the markers so they could be removed and replaced between the different footwear conditions. The subjects then walked on a treadmill and ten continuous footfalls were recorded, gait events of heel contact and toe off were determined by using a kinematic velocity-based algorithm which was applied to the medial calcaneus marker and the toe box marker.



Fig. 3.13. Photographs from Ferber and Benson, (2011) depicting marker placement, shoe cut outs and how the medial longitudinal arch was modelled and plantar fascia strain were calculated.

Ferber and Benson, modelled the medial longitudinal arch as a triangle (see Fig. 3.13.) while the distance between the markers on the medial side of the calcaneus and the first metatarsal head was purported to represent the strain on the plantar fascia. It was reported there were no significant changes in peak rearfoot eversion,

peak tibial internal rotation or medial longitudinal arch angle between the no orthotic and orthotic conditions, however the moulded orthoses did significantly reduce the strain on the plantar fascia ($p=0.03$) over the no-orthoses condition. This was not the case with the non-moulded orthoses. They concluded that this was the first study to investigate the effect of an orthoses on midfoot biomechanics when walking. Heat moulding an orthotic device does not induce any measurable effect. No attempt was made to try and model the forefoot in the coronal or transverse planes, though the lack of markers on the lateral side of the foot would have made this impossible, this would suggest there is still much to learn as to how arch support orthoses effect the midfoot during gait.

Chevalier and Chockalingam, (2012) looked at the effect of eleven sets of custom foot orthoses on one female (elite rower) subject with a six month history of right medial knee pain; each pair of orthoses was made by a different practitioner. The different orthoses were placed in plimsolls with holes cut to accommodate an Oxford Foot Model marker set; no explanation of how the shoes were removed and replaced between trials was given. They presented the minimum and maximum angles plus the angle at heel strike for the knee, rearfoot to tibia and forefoot to rearfoot in all three planes. With the orthoses their subject demonstrated mean reductions in the kinematics of the left knee but not the symptomatic one. The right foot demonstrated significant decreases in peak dorsiflexion and sagittal plane angle at heel strike of the hindfoot to tibia and an increase in peak eversion. The forefoot to hindfoot was significantly less plantarflexed on the rearfoot, but there was an increase in sagittal plane angle at heel strike. The left foot did not demonstrate any significant changes at the rearfoot but peak plantarflexion and pronation was reduced at the forefoot while sagittal plane angle at heel strike was increased. It was stated that though the clinical goal of most foot orthoses was thought to concentrate on the reduction of coronal and transverse plane motion this was the most variable, leading them to declare that foot orthoses may primarily act on the sagittal plane. They concluded that inter-practitioner variability is an important factor when looking at the effects of custom made foot orthoses, so much so that it was recommended to use caution when drawing general conclusions from such devices.

3.1.6. Summary

- The attributes of different foot models were discussed in this chapter, most were designed to look at barefoot motion. As the main aim of the present study was to investigate the effect of orthoses within shoes, a modified version of the Carson, (2001) model was adopted.
- Previous studies, looking at abnormal barefoot motion, were examined; these concluded it was important to model in six degrees of freedom. It was also noted that only recording barefoot motion would not make it possible to determine the effect of foot orthoses
- It was demonstrated that previous studies had tended to cut holes in shoes to record shod foot motion, however, this practice did not consider the effect on the support of the shoe. Shultz and Jenkyn, (2011) did consider this problem and concluded that the size of the holes that could be used depended on the rigidity of the shoe.
- Studies looking at the effect of orthoses were highlighted; however, none had used normal supportive footwear without cutting holes into the upper of a shoe. It was stated that this would be the first study to investigate the effect of foot orthoses on a three segment foot model in unmodified supportive shoes.

3.2. Literature Review Patellofemoral Pain

3.2.1. Anterior knee pain definition

Grelsamer, (2009) stated that a syndrome is not a diagnosis but rather a group of signs and symptoms that often occur together, this lead him to suggest this label was merely a repetition in medical terms of what the patient complained of. It was described as one of the most common but least understood joint pathologies by Townsend, et al. (1977). Though the syndrome is well researched the aetiology has not been attributed to any single source and is thought probably to be multifactorial in origin with both intrinsic and extrinsic factors being blamed (Way, 1999; Alberti, et al. 2010; Barton, et al. 2011) these factors are described in the next section.

3.2.2. Aetiology Patellofemoral Pain

Garth, (2001) suggested that normal movement of the patella should see it move in the patellofemoral groove in an open crescent shape the tips facing laterally. He noted that there is a common tracking problem called J-tracking where there is an observable lateral displacement of the patella during terminal extension. The motion of the patella against the anterior articular surface of the femur is commonly termed patella tracking (Laprade and Lee, 2005). Mal-tracking of the patella is often blamed for the predisposition of patellofemoral pain syndrome; however the normal tracking of the patella is difficult to define due to individuality of knee anatomy.

Heng and Haw, (1996) proposed the patella should run centrally in the trochlear groove of the femur but this relationship can be interrupted by intrinsic geometry of the bony or soft tissue surfaces of the patellofemoral joint, or the extrinsic Q-angle. However it is difficult to directly measure patella motion, Laprade and Lee, (2005) stated that intracortical pins can and have been used for functional evaluation but the invasive nature of the procedure limits the sample size, they did discuss the use of cadaveric knees but concluded the experimental setup could be responsible for the wide variety of results found with this technique, they also highlighted the fact that there were no ground reaction forces and muscle activation was not possible.

Increased lateral patellofemoral joint stress is often cited as an aetiology of patellofemoral pain (Heng and Haw, 1996; Gross and Foxworth, 2003; Herrington and Nester, 2004; Laprade and Lee, 2005; Barton, et al. 2009; Alberti, et al. 2010), though there are a number of anatomical functional factors that are attributed to this condition. It is often assumed to be a result of an increased Q-angle affecting the coronal plane alignment of the patella, where contraction of the quadriceps pulls the patella onto the lateral articular surface of the femur. This excessive force on the lateral articular surface is also blamed on the internal position of the femur in the transverse plane which may be due to the structural anatomy of the femur or from the kinematic control from the hip (Barton, et al. 2009; Alberti, et al. 2010). Another hypothesis for internal femoral position is excessive rearfoot pronation; foot position is thought to lead to internal rotation of the tibia which consequently induces an internal rotation of the femur (Way, 1999; Alberti, et al. 2010).

3.2.2.1. Genuvalgum

Pitman and Jack, (2000) stated that subjects with a genuvalgum tended to have an increased Q-angle. Increased valgus knee angle was also cited by Powers, (2003) as another anatomical presentation that may increase the Q-angle, however it was stated that this could be due to the femur being adducted on the pelvis (coxavarum) or the tibia being abducted on the femur. It was also noted that it may be a transient position due to dynamic instability and it was suggested that this may be due to weakness in the hip abductors especially gluteus medius. A wider than average pelvis was also attributed for moving the centre of mass more medially away from the hip joint centre; this will increase the adduction moment and hence put the hip abductors under greater strain.

Paoloni, et al. (2010) were a little more specific suggesting patellofemoral pain may be linked to kinetic and kinematic changes in the coronal and transverse planes and these may be more pertinent to the clinician. In their experiment they compared nine knee pain sufferers with nine matched controls. When walking the only significant kinematic variable was peak knee adduction during the loading phase of gait. The kinetics revealed further significant differences the hip abductor moment; the knees abductor; external rotator and extensor moments were all greater in the patient group during the loading response. During terminal stance it was only the hip abductor moment and the knee extensor moment that were significant. They concluded that these differences may identify some of the biomechanical risk factors for the development of patellofemoral osteoarthritis.

3.2.2.2. Femoral Rotation

Ramsey and Wretenberg, (1999) undertook a review on the methods used to describe the tibiofemoral and patellofemoral joints, they noted there was not one single “perfect” method for measuring knee joint motion. They discussed that by fixing the markers to the skin there is the potential for considerable movement to occur between the soft tissue and the bony landmarks they are placed over. The review then looked at a number of studies that had used intracortical pins to overcome this phenomenon, however even this invasive method was perceived to have problems

with possible impingement of the structures overlaying the bones; especially the iliotibial band. It was also noted that there are different methods for measuring three-dimensional kinematics; using helical axes, or joint co-ordinate systems amongst others, all of which can affect the results and make comparisons between studies difficult. They concluded that some rotations could be influenced by cross-talk from the flexion/extension of the tibiofemoral joint.

In the same year Reischl, et al. (1999) noted that previously completed studies suggested that the hip and knee internally rotate during early stance. They highlighted that there were difficulties interpreting this data as rotations of a joint are commonly referenced to the proximal segment. Therefore a report of internal knee rotation could be as a result of the tibia internally rotating on the stationary femur, or the femur externally rotating on a stationary tibia. It was also suggested that both the femur and tibia could be externally rotating but if the tibia rotated less this would still be interpreted as internal knee rotation. In Reischl's study of thirty subjects it was reported that tibial rotation was similar between subjects, but femoral rotation was inconsistent. Eighteen exhibited external rotation in early stance while the rest were internally rotated, the group mean was 2.0° (SD4.9) of external rotation.

3.2.2.3. Tibial Rotation

Hefzy, et al. (1992) recognised that tibial rotation will affect the contact area of the patellofemoral joint when considering different degrees of flexion. They took four cadaveric limbs and calculated the patellofemoral contact area using a 3-SPACE system. It was concluded that tibial rotation did not significantly alter the total contact area, but medial femoral contact areas increased with internal tibial rotation at all flexion angles. External tibial rotation caused increased femoral contact on the lateral facet. They also found that the patella shifted medially with internal tibial rotation during flexion. Internal tibial rotation also caused the patella to tilt medially at near full extension and rotated more medially at full flexion, both these latter positions will cause an increase in the medial contact areas.

Lee, et al. (2003) undertook a review of the literature considering the influence of femoral and tibial rotation on the patellofemoral joint. They identified that rotations of both segments have a significant effect on the kinematics of the patella; ultimately it being the attachment of the patella tendon which influences the direction of patella

movement. The authors stated that internal rotation of the tibia causes a slightly increased pressure on the medial facet of the patella while the opposite was true for lateral rotation, however this is opposite when considering femoral rotation; internal rotation increases the compressive forces on the lateral side of the patellofemoral joint.

Although excessive internal tibial rotation provides a convenient explanation for the overloading of the patellofemoral joint, Levinger and Gilleard, (2005) found no significant relationship between maximum internal rotation and maximum eversion of the foot. When looking at two groups of subjects one with patellofemoral pain and the other being a control group with no pain, they suggested this may be to do with variability in the alignment of the sub-talar joint axes of the subjects. They did find a significant correlation between the timing of peak internal rotation and peak eversion but only in their group of knee pain subjects not in the control group.

Levinger and Gilleard, (2007) looked at two similar groups, stated that the group with knee pain reached peak eversion of the rearfoot later in the stance phase; they speculated this could be to aid shock absorption early in the stance phase and consequently may affect the propulsive phase. Conversely they did not find any prolonged internal rotation of the tibia this was thought to be due to the fact peak eversion and internal rotation did not correlate with each other.

3.2.2.4. Foot Pronation

In 1999 Reischl, et al. conducted an experiment on thirty non-symptomatic subjects to establish if foot pronation magnitude and timing was related to similar parameters of the tibia and femur. They stated that it is clinically accepted that foot pronation is linked to internal tibial rotation, but their cohort did not exhibit any statistically significant link. Reischl, et al. suggested that this lack of correlation may be due in part to the fact they modelled the foot as a single segment, the combination of rearfoot and forefoot motion may not be pertinent during early stance.

In contrast to this Curren, et al. (2011) investigated the effect foot posture had on knee alignment; this was done by recruiting 335 healthy individuals with pronated, neutral and supinated feet. This was measured using a static trial; foot pronation was associated with increased internal rotation of the femur and tibia, which in turn

caused medial displacement of the tibial tubercle. It was also found that this changed the patella alignment and tended to increase the valgus alignment of the knee itself.

Barton, et al. (2011) found their patellofemoral pain group demonstrated earlier rearfoot eversion than their control group. They suggest this may be due to the fact gait velocity was not measured in the previous experiments. They suggested reducing gait velocity is a recognised compensation mechanism for reducing knee pain and could be an explanation for the contradiction between these studies. They also noted that there may be sub-groups within the patellofemoral pain population, some with early peak eversion while some exhibit delayed eversion. They found that their knee pain group had a mean increase of sagittal plane rearfoot motion of 4.1° over the control group this was suggested to show the difference in gait velocity. However this was only evident when the rearfoot was related to the laboratory axis, when related to the tibia there was no significant difference found. Though they used the Oxford foot model which splits the foot into 3 segments they only reported on the differences in rearfoot motion. They proposed that to expose differences in walking tasks there may have to be a controlled pace on large numbers of subjects.

3.2.2.5. Knee Flexion

Townsend, et al. (1977) were interested in establishing the contact area of the patella's articular surfaces on the femur at different fixed flexion angles. This was achieved by introducing dye into the joint under a force of 50kg in eleven fresh cadaver limbs which were clamped in a jig. It was noted that during 50° - 60° of flexion, as is seen in stair climbing, the cancellous bone found in the central-medial facet is "overloaded". Townsend, et al. concluded that the structural makeup of the patella is directly related to the forces that it has to withstand.

Powers, et al. (1999) stated that it is during flexion of the knee in the loading phase of gait that the patellofemoral joint is under its greatest load, when the quadriceps are acting eccentrically to aid shock absorption. They investigated if a group of patellofemoral pain subjects demonstrated differences in the loading response when walking at a self-selected speed and a fast pace. The affected group demonstrated a significant reduction in both peak loading response and peak vertical ground reaction force compared to the controls during their self-selected speed. Due to the fact there were no kinematic differences observed this was directly attributed to the measured

reduction in gait velocity, cadence and stride length. During fast walking the patellofemoral pain group still managed to reduce peak loading and vertical ground reaction forces compared to the control group. Powers, et al. found significantly reduced loading response knee flexion and reduced step length. Due to the timing of maximum loading knee flexion, they proposed that peak loading response is controlled more by gait velocity than the position of the limb, but it could possibly be the eccentric action of the quadriceps that limits the peak vertical ground reaction force.

Duffey, et al. (2000) also found that patellofemoral knee pain subjects had reduced vertical impact forces over their control group yet there was no difference between the symptomatic side and the non-affected side. They suggested this questions the belief that higher impact forces can be related to over-use injuries and suggested that it may be necessary to look at joint moments to establish any differences. Garth, (2001) identified clinically that if the tibia is displaced posteriorly on the femur due to ligament laxity or a deficiency in the anterior cruciate ligament this will also have the effect of increasing the angle between the line of pull of the quadriceps tendon and the patella tendon; hence the patellofemoral contact force is also increased (see Fig. 3.14.).

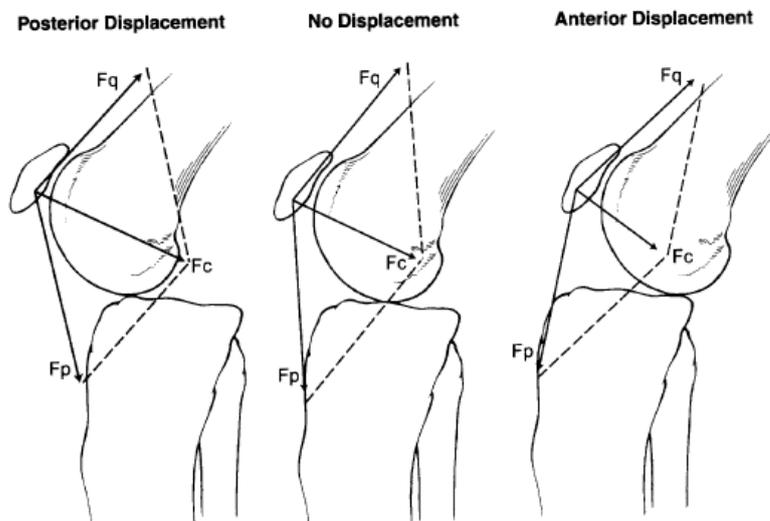


Fig. 3.14. Diagram from Garth, (2001) illustrating how position of femur on the tibial plateau in the sagittal plane will affect the patellofemoral contact forces.

Hunter, et al. (2007) conducted X-rays on 595 subjects (aged 70-79 years mean=73) with knee pain via the fixed flexion technique for the tibiofemoral joint and weight bearing skyline views for the patellofemoral joints. They carried out the X-rays at the initial assessment and after 2 and 5 years and recorded joint space narrowing giving it a score. Three anatomical measures were considered when looking at the patellofemoral joint: bisect offset of the patella from the centre line of the femoral condyles; the sulcus angle describing the depth of the groove in the femur that the patella articulates with and the angle of patella tilt (see Fig. 3.15.). None of the three measures could be related to the pain scores; however the offset produced the most noteworthy joint space narrowing.

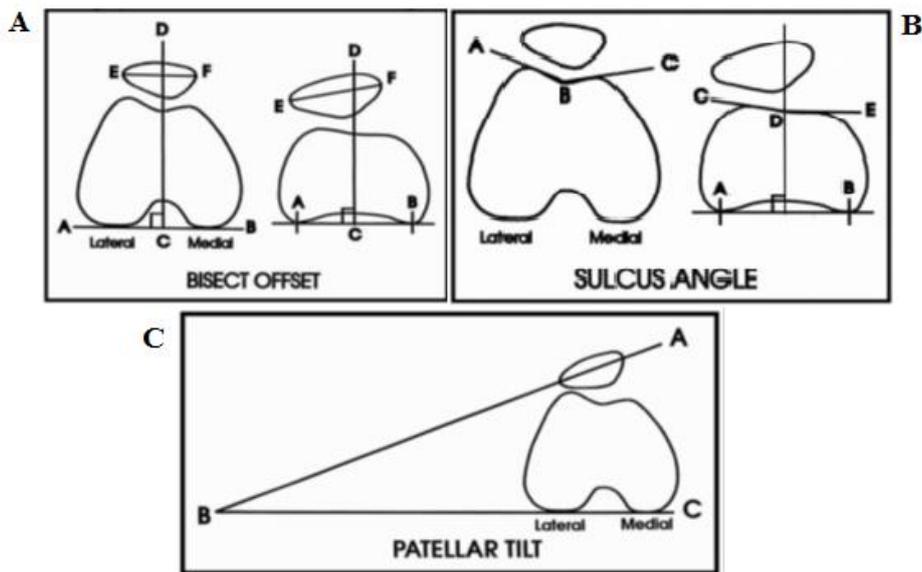


Fig. 3.15. Diagrams from Hunter, et al. (2007) illustrating the patella positions under scrutiny.

Quintelier, et al. (2008) stated that the at knee flexion angles of less than 60° the quadriceps lever arm works with a mechanical advantage, angles over this and the quadriceps has to work at a disadvantage. They looked at the pressure the patella exerts on the femur using a post mortem knee in an Oxford based test rig (see Fig. 3.16.) they found that the size of the contact area of the patella was related to the quadriceps force, and knee flexion. They stated that the patella never loses contact with the femur in full extension but as the knee extends from a flexed position it is

the superior lateral and the inferior medial areas of the patella that are subjected to the greatest pressure.

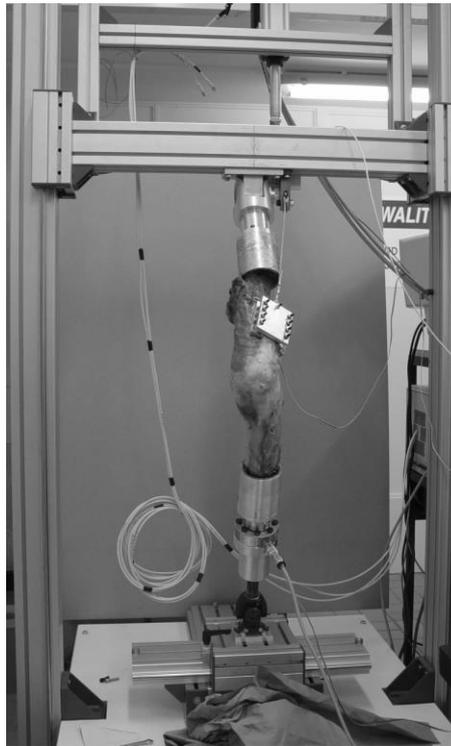


Fig. 3.16. Oxford test rig, picture from Quintelier, et al. (2008) showing how cadaveric limb was fixed for their experiment on patellofemoral pressures

Farrokhi, et al. (2011) acknowledged the value of the cadaveric studies in determining the stresses within the patellofemoral joint but questioned the fact this form of experimentation is unable to measure dynamic muscle loading. They suggested that it may be possible to assimilate subject-specific musculoskeletal measurements into a computer model to explore patellofemoral joint stress. Ten subjects with patellofemoral pain were compared with ten control subjects; all had a magnetic resonance image of their knees showing bone geometry and cartilage morphology; squat positions at 15° and 45° angles were recorded using 3D motion analysis with concurrent electromyography. All these measurements were incorporated into a computer model (finite element). Generally stress increased with increasing flexion angles though the highest peak and mean stresses were seen on the lateral side of the patellofemoral joint. It was reported that the symptomatic group

had significantly higher peak and mean stress levels within the joint leading to the conclusion that treatment strategies need to be aimed at reducing these stress levels.

This section outlines that there are a number of theories to the underlying cause of patellofemoral pain. Most patients presenting with this condition will probably have a number of aetiologies superimposed on top of one another. Due to previous studies not always reporting alterations in gait patterns between patellofemoral pain patients and control groups, it was deemed necessary to investigate a more demanding task. As highlighted in the background section, step descent is an everyday function which provides an increased challenge.

3.2.3. Step Descent and Patellofemoral Pain

Townsend, et al. (1977) stated that stair climbing requires higher flexion angles, the area of articular surface of the patella in contact with the femur at 50° to 60° is overloaded and hence at greater risk of cartilage erosion. Selfe, (2000) noted that the articular surface of the patella has to move proximally on the articular surface of the femur as the knee flexes, this is a consequence of the patella being anchored to the tibia via the “inelastic” patella tendon. He explained that the proximal movement of the patella has the effect of increasing its moment arm thus reducing the effectiveness of the quadriceps, this progressive reduction was illustrated by stating that the quadriceps are only half as efficient when the knee is flexed at 90°. Selfe, conducted an experiment on 100 subjects (200 legs) descending a step slowly and measured the knee flexion angle at which they lost control; this was termed the critical angle. The average result was 61.3° (SD8.9) but gender and limb dominance did produce statistically significant differences.

In 2002 Brechter and Powers, maintained that ascending and descending stairs is one of the most challenging activities undertaken by individuals with patellofemoral pain. It was reported that their group of 10 patellofemoral pain subjects had a significantly reduced cadence when descending steps compared to a control group (115.4 vs 153.6 steps/min). They found there was no difference between the groups with respect to patellofemoral joint reaction force during the descent but there was a trend towards reduced patellofemoral joint reaction force time integral.

Patellofemoral joint reaction force was explained as the product of the force calculated from the quadriceps contraction with the knee flexion angle. Brechter and Powers, concluded that their symptomatic subjects did not demonstrate increased patellofemoral joint stress during descent compared to their control group, but this was possibly due to the adoption of compensatory strategies.

Anderson and Herrington, (2003) conducted a similar study to Selfe, (2000) but chose to look at isokinetic torque production and velocity of the knee. They looked at twenty patellofemoral pain patients and twenty control subjects; 50% of the patients and 15% of the control subjects demonstrated a “break” in the torque curve. The same three control subjects had a similar interruption in the velocity curve while 60% of the patellofemoral pain patients displayed the break. Anderson and Herrington, concluded that the patellofemoral pain subjects could not produce a smooth controlled eccentric quadriceps contraction under load; they hypothesised the “break” in the contraction could be due to quadriceps inhibition.

Selfe, et al. (2008) highlighted the fact that researchers needed to find a suitable activity that would dynamically “challenge” the knee, but was also an activity that the patients would encounter on a regular basis thus avoiding injury. Step descent was proposed due to the increased eccentric control it requires over level walking as the centre of mass progresses forward and gravity has to be resisted during the controlled lowering phase. The twelve asymptomatic subjects were asked to descend a 20cm step as slow as possible while the eccentric knee had no intervention, tape or a brace applied. By using the Calibrated Anatomical System Technique (C.A.S.T.) it was reported that the coronal and transverse kinematics and kinetics were reduced, this was purported as an improvement in knee joint control. However this came with caveats that there was no way of knowing if a patient cohort would react similarly and the mechanisms by which the interventions worked was also unknown; although neuromotor and mechanical were proposed. Selfe, et al. concluded that coronal and torsional kinematics and kinetics must not be excluded when investigating step descent.

Grenholme, et al. (2009) compared a group of 17 females with patellofemoral pain with a control group during stair descent; they found only the angular velocity of the

stance limb displayed any significant difference between the groups. They hypothesised that the symptomatic group would exhibit decreased knee flexion this however had to be rejected. Grenholme, et al. did find that plantarflexion of the swing/impact limb was significantly increased; it was proposed this could account for reducing some of the final load on the patellofemoral joint at the end of the lowering phase. This mechanism was presented as an alternative to altering knee kinematics, hence could explain the lack of significant changes. It was also noted that individual anthropometrics may substantially contribute to the kinematics of step descent, and hence may need to be normalised.

In 2011 Selfe, et al. undertook a follow-up study to compare the effects of a brace and a simple taping technique on thirteen patients with patellofemoral pain. They suggested that previous studies which had focussed on the patellofemoral joint during step descent had only reported sagittal plane or used simple marker sets which had led to incomparable results. It was highlighted that patella bracing tends to direct the forces medially therefore measuring the sagittal plane would not be as valuable as the coronal and transverse planes. Subjects were asked to descend a 20cm step as slowly as possible with no intervention, neutral patella taping or a knee brace, each subject acted as their own control. It was reported that there was no difference in the maximum adduction angle however there was a marked difference seen between the patients. The coronal range was significantly reduced by both the taping and the brace (mean angle was reduced from 8.96° to 5.34° with the brace and 8.07° with the tape) the latter intervention also significantly reduced the range over the taping. It was similar in the transverse plane in that the peak angle demonstrated no significant change but again the range was significantly reduced by both treatments. There were no differences found between conditions when looking at the coronal and transverse plane moments. The authors suggested that a neutral taping technique could only be aiding the knees stability due to proprioception, though there was some uncertainty whether it was because the brace had an increased surface area therefore a greater sensory input, or whether the lateral buttress straps had some mechanical effect. It was recognised that some of the differences in the measurements were very small, but as they were reducing the compensatory movement of the knee they were probably of clinical relevance.

3.2.4. The Use of Foot Orthoses for Patellofemoral Pain

Lafortune, et al. (1994) inserted 3 Steinmann pins (tibia, femur and patella) into five normal male volunteers to assess knee motion, they were asked to walk with neutral, 10° varus and 10° valgus shoes to assess the influence motion of the sub-talar joint had on knee function. Coronal plane motion was found to be small the total range of motion was less than a degree. The mean difference between the footwear was minimal though all five subjects did show a similar difference when comparing the varus and valgus shoes at peak knee flexion. Transverse rotation of the knee was also subtly affected by the different shoes though it was at peak knee extension that all subjects showed the same difference in the same direction, the 20° of coronal foot wedging only induced 1.3°(0.8) of transverse plane motion. Tibial rotation was affected more by the footwear at both peak knee flexion and extension (4.3° and 2.7° respectively) though this rotation was with respect to the laboratory axis. They concluded that even though clinical evidence suggested that minor footwear modifications could induce changes at the knee joint the extreme modifications they used only induced subtle kinematic adjustments at the knee.

Klingman, et al. (1997) used X-rays to observe any changes in patella position when using medially posted foot orthoses, they recruited twelve asymptomatic females who demonstrated at least 6° of calcaneus eversion in relaxed stance. Each subject had a pair of orthoses fitted that positioned the posterior bisection of the calcaneum perpendicular to the supporting surface. They then had a skyline view of the patella taken while standing on a step with the knee in 45° of flexion; this was done with the subject barefoot then repeated wearing the orthoses. From the radiographs the lateral patella displacement was measured to establish any alteration in patellofemoral joint alignment each subject acting as their own control. They found all 16 knees demonstrated a reduced lateral patella displacement with the orthoses (mean difference 1.8mm SD 0.52mm) they concluded this difference was statistically significant but recognised that a static measurement may not be a good predictor of dynamic motion.

Way, (1999) used a more rigid design of foot orthoses on a single athletic subject; the activities he looked at were walking, stairs and squatting, running,

jumping/landing and twists. Overall the subject's pain improved within 2 weeks the author suggested that this could have been due to the orthoses being moulded to the subject's feet rather than just using external wedging and the rigidity of the devices. He also acknowledged that there are limitations in the conclusions that can be drawn from a single subject design. Way, did suggest that it may be possible to over correct subjects' feet by trying to obtain a position close to sub-talar joint neutral, a goal suggested by Root, et al. (1977).

In 2000 Pitman and Jack, suggested it was possible to use foot orthoses as a first line treatment for patellofemoral pain. They fitted fifty seven athletic subjects with symptomatic knees with foot orthoses then sent them a retrospective questionnaire to establish their pain levels following the intervention. Forty one subjects responded and the authors split the subjects into four groups' dependant on gender and age (older or younger than 20 years); all but two of the respondents reported some pain reduction. It was suggested that younger athletes experienced quicker and greater pain reduction than older individuals.

Duffey, et al. (2000) compared a large number of runners, 99 with knee pain and 70 without, they reported that the injured runners demonstrated less rearfoot motion and could only suggest that this is more likely to be related to the timing of joint actions. They concluded runners with higher arched feet and reduced pronation in the first 10% of the gait cycle are more likely to develop patellofemoral problems. Neptune, et al. (2000) took a different approach to look at the effect of foot orthoses and strengthening of Vastus medialis on knee pain. They recorded the kinematics and kinetics of nine healthy subjects running then applied a sophisticated dynamic musculoskeletal model with fourteen functionally independent muscle groups to their results. They found that the strengthening of the muscles had the most consistent effect in reducing lateral patellofemoral joint loading while the prefabricated soft foot orthoses they used did have a beneficial effect in some subjects but no effects in others, however, this could only be modelled as extra stiffness; no adverse results were reported.

Gross and Foxworth, (2003) noted that the experimental evidence for using foot orthoses to combat patellofemoral pain syndrome is “theoretical and circumstantial”. Thus they reviewed grouped experiments that used: pain outcome measures; relationship between foot and patellofemoral mechanics and the effects of foot orthoses lower limb mechanics. They found many studies which looked at using retrospective pain scales reported overall improvement. The mechanics with and without orthoses were more variable, however they concluded that clinically patients may benefit from the use of foot orthoses if they demonstrate a pronated foot position, or the orthoses help the anatomical alignment in the coronal or transverse planes.

Collins, et al. (2008) undertook a randomised control trial on 179 participants with patellofemoral pain; they were divided into four groups: one group received prefabricated foot orthoses; another had only flat insoles fitted; a third group received only physiotherapy treatment while the last group received both physiotherapy plus the orthoses. They found that the prefabricated orthoses improved the subjects’ pain scores in the short term over the flat insoles but there was no long term benefit in combining them with physiotherapy treatment or having physiotherapy on its own. They concluded that all groups did improve therefore it may be worth using orthoses to accelerate recovery.

Powers, (2010) identified that orthoses can be provided without considering underlying biomechanics leading him to conclude that using orthoses to treat patellofemoral pain is a trial and error treatment. He summarised that the link between patellofemoral joint function and foot pronation is tenuous and therefore it is still not certain whether foot orthoses have the potential to alter lower limb function. It was consented that there is a certain amount of circumstantial clinical evidence which supports the use of foot orthoses as a treatment modality but says it is essential for future research to be focussed on which subjects will respond to this form of treatment.

Barton, et al. (2011) used prefabricated orthoses on 60 participants with patellofemoral pain; they used the anterior knee pain scale and the lower extremity functional scale to determine their effects. The tasks set the participants were: how

many pain free 20 cm step downs could they complete; how many single leg raises from sitting could they manage and the change in pain of a single leg squat. All these tasks were completed at initial fitting; after 6 weeks and after 12 weeks with and without the orthoses. There were significant improvements in the number of pain free step downs and single leg raises that could be undertaken and the foot orthoses improved the pain scores significantly at 12 weeks when completing the single leg squats. They observed greater beneficial effects with the orthoses after 12 weeks of use; more than after initially fitted. There were however no improvements of perceived differences between the 6 and 12 weeks tasks suggesting there may be a plateau in recovery over time. To date there have been no studies conducted that report on the effects of foot orthoses during step descent.

3.2.5. Rational

Medially wedged orthoses are often used in clinical practice to reduce foot pronation which is purported to reduce internal rotation of the lower limb (Boldt, et al. 2013). It was felt that as this study was to attempt to measure midfoot motion it was imperative to mould the orthoses to the shape of the participants feet, hence all the orthoses were to be heat moulded and supported to that shape. To maintain consistency within the orthoses a 5° medial wedge (commonly used and commercially available) was to be placed under the insole. Often the wedge only extends to the metatarsal heads (3/4 orthoses) which aids fitment to the patients footwear. However, due to the step down trials being undertaken in the present study, it was considered that the lack of wedging under the metatarsal heads may not provide support during the lowering phase; hence a full length wedge replaced the 3/4 length wedge for sets of walking and step descent trials. Although this is done infrequently in clinical practise, it was considered to be the simplest method for keeping the prescriptions as constant as possible between the subjects.

3.2.6. Clinical relevance

Form a review of the literature; it is obvious that there is a lack of information regarding the amount of change which is clinically relevant. Dawson, et al. (2007) suggested that there are two methods of looking at small changes, the first being statistical; the minimal detectable change (MDC) was a method to describe the smallest change for an individual subject that is probably beyond the measurement

error and hence is likely to be actual change. They highlighted that it is not known whether this measurement would be clinically relevant. The second is the minimal clinical important difference (MCID), which is defined as the smallest change a patient would notice which may lead to a change in management, however this latter method is not related to the measurement error of the equipment hence clinically important change may be missed when investigating statistically.

Landorf and Radford, (2008) suggested that statistically significant change only considers the probability that two treatments are different, therefore there can only be a degree of conviction that the result is not just due to chance. It was also stated that just because an effect reached statistical significance, it does not mean that it will be clinically significant; in reality the change could be imperceptible to the subject. Landorf, and Radford, defined MCID as the smallest change which a patient perceives as being beneficial. Mathieson and Upton, (2008) stated that statistical significance and clinical significance are not always associated; they also suggested that even if the change is small, rejection must be made with caution if that change could be clinically relevant.

Keurentjes, et al. (2012) used a very similar definition however they noted that it could also be a detrimental change. It was stated that MCIDs are often calculated from pre- and post-operative scores where patients indicate they are “somewhat” better or worse. Copay, (2012) proposed that MCIDs were suggested as the threshold at which most patients would reach an improvement, however she noted that just describing a group average did not always reflect a patients experience and as such MCIDs failed to convey their definition in certain instances.

Wilken, et al. (2012) recognised that computer gait analysis is often used to determine the effectiveness of a treatment but the detection of actual change is limited due to the errors induced by the data collection process, which raises the question what is the minimum detectable change? Wilken, et al. recognised that minimal detectable change was reduced considerably when range of motion was investigated rather than looking at the maximum position. However clinically what we are trying to determine is the minimum clinically important difference between patients and normals, and the minimum clinical important change when a patient is

given an intervention. To date no threshold values have been published to identify the effect of orthotic management, however clearly this is an important area for further investigation in Biomechanics and Gait Analysis.

3.3. Initial Aims/Hypothesis

The aim of this study was to establish the optimal method for using the multi-segment foot model with footwear. It was also investigated whether it was possible to measure the effect of orthoses when using supportive shoes compared to sandals which allow placement of the markers on the skin.

Objectives

1. Compare the kinematics of the sandal and shoe conditions when walking and descending steps.
2. Compare the kinetics of the sandal condition to the shoe condition when walking and descending stairs.
3. Establish normative data for rearfoot, midfoot and forefoot segments when descending stairs.
4. Determine if standard three-quarter length and full length foot orthoses have the same effects when using sandals and shoes.

It was hypothesised that it was possible to measure foot motion within a shoe by comparing it to markers on a foot using a sandal and that changes that were produced by using orthoses would be similar within both footwear types.

Chapter 4. Method Non-symptomatic Subjects

4.1. Subjects Initial Study

Fifteen healthy subjects were recruited from a staff and student population of the University of Central Lancashire: 7 males and 8 females, (mean age 30.06 SD 9.99). The participants were included in the study as long as they had no lower limb pain or back pain at the time of data collection and had not suffered regularly with non-specific knee pain in the past. They had to have mild to moderate foot pronation (score +6 to +9) as defined by the using the foot posture index (Redmond, et al. 2006; Evans, et al. 2003). Permission to use the Foot Posture Index (FPI) was sought and granted by personal communication. Written informed consent was obtained from each individual prior to starting the trials; ethical approval was obtained from the Faculty of Health Research Ethics Committee University of Central Lancashire, Preston.

4.2. Footwear Conditions (Normal subjects)

All subjects completed the walks and step descents under the following footwear conditions.

- Sandals Only.
- Sandals with moulded orthoses and 5° three quarter wedge.
- Sandals with moulded orthoses and a 5° full length wedge.
- Shoes only.
- Shoes with moulded orthoses and 5° three quarter wedge.
- Shoes with moulded orthoses and 5° full length wedge.

The sandals were constructed specifically for this study by the author from medium density EVA (Ethyl Vinyl Acetate, shore 40) with a high density EVA base (Shore 60) they were constructed with a removable foot-bed; this enabled the sandal to be used as either a normal “flat” sandal with the centre portion in place, or as a holder for the orthoses with the centre removed (see Fig. 4.1.). Velcro straps were glued between the midsole and the sole and positioned so as not to impede the markers which were positioned directly on to the foot (see Fig. 4.2.); this was the main consideration behind the design of the sandals. The subjects were asked to wear their

normal training shoes which would accommodate the orthoses for the shod trials (see Fig.4.3.).



Fig.4.1. Photographs showing sandals with footbed in (top left) and without footbed (top right) Orthoses inserted (bottom)



Fig.4.2. Photographs depicting marker placement and sandals - constructed by the author



Fig.4.3. Photographs depicting marker placement on the shoes and lower leg cluster; note extra tape over the medial markers which were more at risk of collision when the subjects were walking.

4.3. Orthoses

The orthoses were custom fabricated by using a correctly sized pair of Slimflex™ insoles. These were heated with a hot air gun and moulded to the longitudinal arch profile of each subject; this shape was then maintained with low density EVA (shore 30) which was ground with no posting. To standardize the posting of the orthoses, 5° medially wedged EVA was used under the moulded orthoses. Either 3/4 which, is usual in the clinical situation, finished just proximal to the metatarsal heads, or full length wedges were used and placed under the moulded orthoses and held in place with double sided tape for the trials (Fig. 4.4.).



Fig.4.4. Photograph showing Slimflex™ insole moulded to subjects foot and supported with low density EVA

4.4. Equipment and Materials

Three wooden steps were constructed by the author using solid pine (legs; 94mm x 72mm, stretchers; 142mm x 46mm in section) and MDF (19mm), these were over engineered to guarantee their stability using large mortise and tenon joints between the legs and the stretchers, the tops were then glued and screwed with sixteen 40mm size 8 screws. The legs were designed to protrude beyond the surface area of the step as it was felt it would make the steps more stable if a subject was to inadvertently place their foot towards the periphery of the steps surface during the trials (See Figures 4.5. and 4.6.).

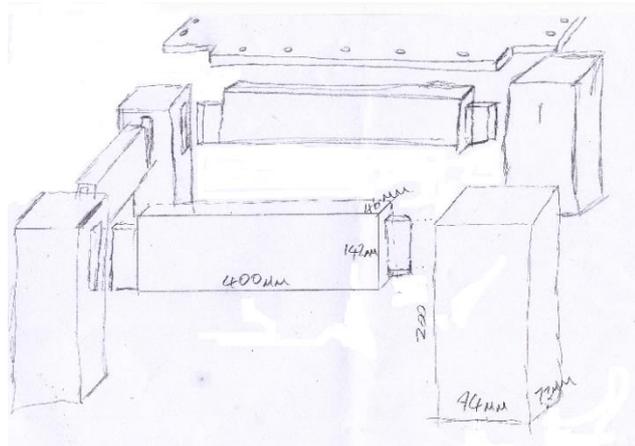


Fig.4.5. Initial “rough” design sketch of steps with right hand stretcher omitted

Two of the steps were constructed to have a height of 20cm the other had an overall height of 40cm. The two shorter steps were situated both sides of the taller one, the taller one and the “second” of the shorter ones being placed on force plates (see Fig. 4.6.).



Fig.4.6. Photographs of steps (group of 3 not positioned on force plates, photographic purposes only)

The markers that were to be placed on the feet were custom fabricated by the author by turning 12mm dowel into 10mm spheres with a plinth (Fig.4.7.) before being covered with 3M™ reflective tape. This was done as it was felt the “plinth” gave extra area for the adhesion of the markers to the skin or the shoe, the proprietary 10mm markers available at the inception of the study were more apt to fall off and deformed more easily as they were constructed from foam rubber.



Fig.4.7. Photograph of custom made 10mm wooden spheres with “plinth” to aid adhesion to subjects prior to covering in reflective tape

4.5. Methods of Analysis

4.5.1. Segment Modelling

Using a standardised protocol 52 passive retro reflective markers were attached to the pelvis, lower limbs and feet of each subject by the author. The pelvis was modelled from markers over the anterior and posterior superior iliac spines, the proximal end of the thigh was defined using the greater trochanter which in combination with the pelvic markers modelled the hip joint centre; the distal end of the thigh was marked at the epicondyles of the femur. The latter markers were also used to denote the proximal end of the tibial segment. The malleoli were used to establish the length of the tibia; these landmarks made up the anatomical markers for the standing trials (Fig.4.8.) and were removed before the walking or step descent trials were completed. A rigid plate with four non-collinear markers was placed centrally and laterally on each of these segments to act as the tracking markers (Calibrated Anatomical System Technique [CAST] Angeloni, et al. 1993; McClay, et al. 2000, Selfe, et al. 2011) the corners of the plates were marked using ink to check for movement of the plates during the trials.

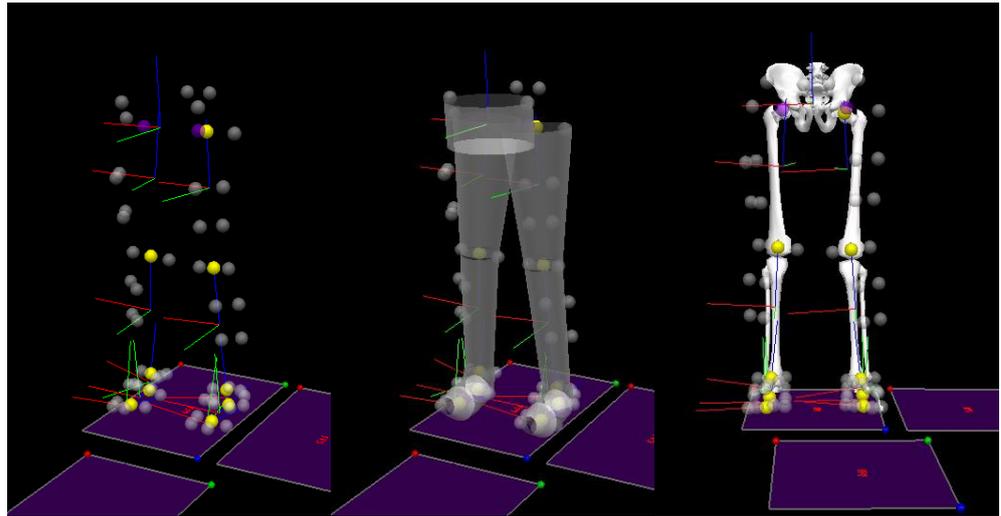


Fig.4.8. Static calibration showing marker placement, segment modelling and skeletal representation (left to right) Segment co-ordinate frames included in each picture

All foot markers were placed directly onto the skin for the sandal trials, or directly onto the shoes vamp over the corresponding areas described. The heel or calcaneal segment was modelled using two markers on the back of the heel; the lateral one being slightly lower. The distal margin was defined by two markers which were inferior and perpendicular projections from the malleoli. The anatomical markers on the malleoli defined the proximal end of the segment and the height of the segment; the projections acted as the distal anatomical and tracking markers.

The metatarsal segment was defined proximally by the projections from the malleoli, and distally by markers just proximal to the first and fifth metatarso-phalangeal joints. There were three tracking markers placed over the dorsum of this segment taking care they did not make a straight line, to prevent gimbal errors. The phalangeal segment used the metatarso-phalangeal joint markers as its proximal margin and then distally there were markers on or over the hallux and on or over the fourth toe (all marker placement is summarized in Table 4.1.). There was no attempt to model the hallux as a separate segment in this study as it was felt this would have little relevance when the foot was shod. Markers over the first and fifth metatarso-phalangeal joints were used in modelling the midfoot; it was felt this may have a considerable bearing on the coronal and transverse plane motion of the phalangeal segment with respect to the metatarsal segment, therefore only the sagittal plane motion was reported. Modelling each segment in this way allowed for them to be

tracked in six-degrees of freedom. A number of pilot trials were carried out before the main study established that the markers placed medially on the feet were prone to being caught by the other foot and knocked off. Therefore extra surgical tape was fitted over the markers and around the “plinth” which seemed to prevent this from happening (see Fig. 4.9.).

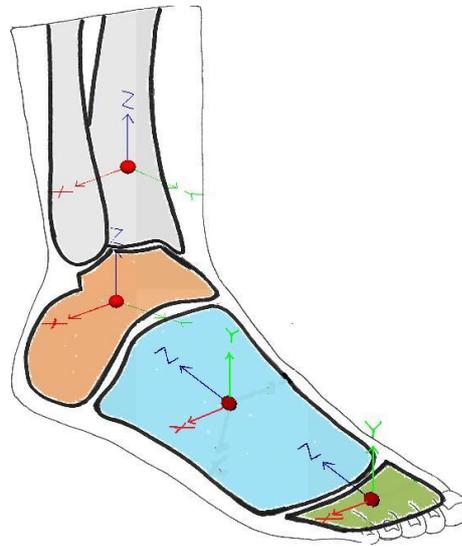


Fig.4.9. Drawing of the three segment foot model plus the shank and segment coordinates (the shank co-ordinate system positioned lower for diagrammatic purposes)

The proximal and distal ends of each segment were used to determine the orientation of the segment co-ordinate system (Brown, et al. 2009). The X-axis relating to sagittal plane motion, Y-axis coronal plane and the Z-axis transverse plane, similar to the description by Grood and Suntay, (1983). In the leg and rearfoot segments this gives flexion/extension adduction/abduction and internal/external rotation respectively. However, the segment co-ordinate system is rotated in the midfoot and forefoot segments as the long axis is at right angles to the leg, this leads to the movement in the Z-axis becoming inversion/eversion while the Y-axis demonstrates abduction/adduction; the X-axis remained constant and still displays dorsiflexion/plantarflexion (Fig. 4.9.).

Table 4.1. Positions of markers defining which segments and their type.

Segment	Marker location	Type
Pelvis	Anterior superior Iliac Spines (ASIS)	Dynamic
	Posterior superior Iliac Spines (PSIS)	Dynamic
Thigh	Greater Trochanter	Static
	Mid-Thigh Cluster	Dynamic
	Medial and lateral epicondyles of the Femur	Static
Shank	Medial and lateral epicondyles of the Femur	Static
	Mid-Shank Cluster	Dynamic
	Medial and Lateral Malleoli	Static
Calcaneal Segment	2 Offset Rear of Heel	Dynamic
	Medial and Lateral projections from Malleoli	Dynamic
Metatarsal Segment	Medial and Lateral projections from Malleoli	Static
	3 Dorsal Tracking Markers	Dynamic
	Medial First and Lateral Fifth metatarsal Heads	Dynamic
Phalangeal Segment	Medial First and Lateral Fifth metatarsal Heads	Dynamic
	Dorsum Hallux	Dynamic
	Dorsum Fourth Toe	Dynamic

4.5.2. Moment referencing

Knee moments were expressed using the shank as a reference, ankle moments were expressed using the calcaneal segment as a reference (distal reference frames). Joint moments were calculated using a standard inverse dynamics approach, before being normalised by dividing by the subject's body mass (Schache and Baker, 2007). It is conventional to use a global reference (laboratory reference) to calculate joint moments Liu and Lockhart, (2006) found the global reference tended to underestimate the coronal and transverse moments during normal gait. With the present study looking at step descent it was felt at the extremes of flexion a local reference system would be more appropriate.

4.6. Procedures

Each subject started in either the shoe or sandal condition. Initially, a five second static trial was recorded. The subjects were asked to stand in the centre of the calibrated area looking down the laboratory; they adopted a relaxed position with their arms folded while this was completed. The anatomical markers over the greater trochanters femoral epicondyles and the medial and lateral malleoli were removed and subjects were randomly assigned to complete either the walking or step trials. The orthoses to be worn for the first trial were simply randomised by selecting a wedge from an opaque bag.

For the walking trials the participants were asked to walk the length of the laboratory through the calibrated area focussing their gaze on the join of double doors at the end of the laboratory this was found to centre most subjects over the force plates, and they were asked not to look at the floor. Each participant started the walk with the same limb from a start point which was adjusted so they were able to land on the force plate regularly without altering their stride pattern. Each subject completed at least five trials for each footwear condition a trial was accepted if they hit the central area of the force plates with no obvious targeting.

Each step trial consisted of the subject ascending a 20cm step onto the 40cm step, they were then asked to count to three before descending onto the second 20cm step and then onto the laboratory floor. All participants were asked to descend with the same leg each time, though it was up to them as to which leg they should use. Again five trials were recorded for each footwear condition, trials were rejected where the participant descended on the wrong limb, or there was hesitation for any reason.

4.7. Data Collection

Data collection was performed in a purpose built gait laboratory that was equipped with ten infrared Oqus cameras (Qualisys medical AB, Gothenburg Sweden) positioned around its periphery in an umbrella type configuration (Richards 2008) Fig. 4.10. these were focussed on a central area above the force plates.

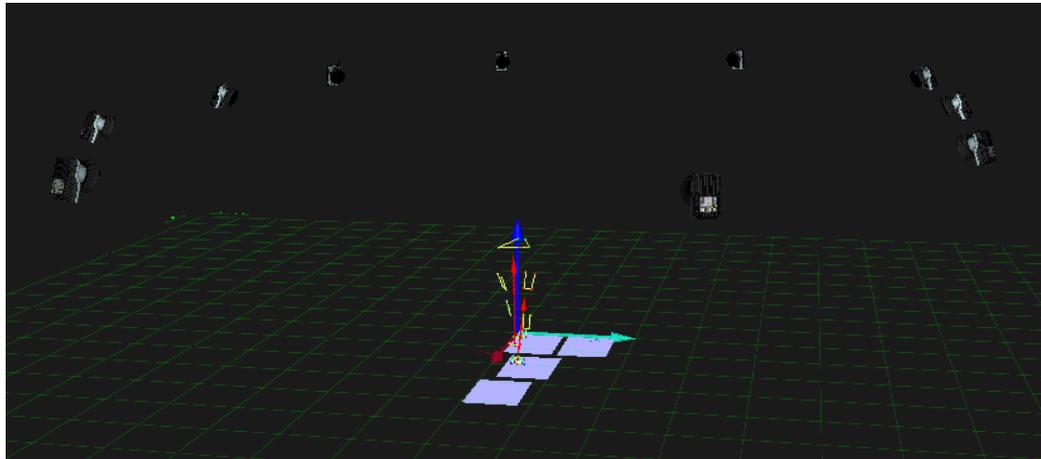


Fig.4.10. Pictorial representation of “umbrella” camera positions in Qualisys Track Manager™

For every recording session a strict protocol was followed for set up and calibration of the system; each of the ten cameras were positioned by using three groups of markers just beyond the periphery of the force plates Fig.4.11. illustrates the area covered by the cameras.

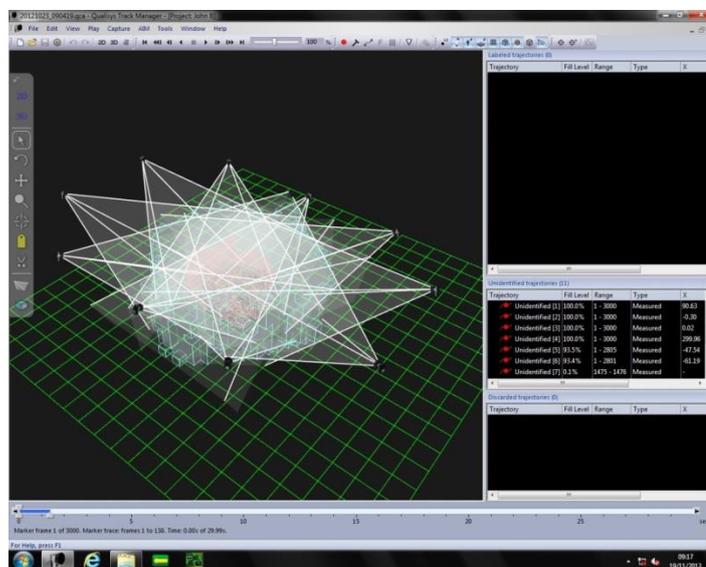


Fig.4.11. Screen shot of the coverage of the data collection area of each of the cameras and the calibrated area (pink central area).

The cameras were set to record at 100 Hz before the aperture of each was adjusted to aid focus and clarity of each of marker in each group. To calibrate the system a static “L” shaped reference structure with four reflective markers of known length (750mm) was placed on the corner of force plate one; its long side ran parallel with the length of the laboratory, this set the global reference as XYZ. A “T” shaped wand with two markers a known distance apart (298.1mm) was moved through the data collection area in all planes taking into account the height the subjects pelvis would reach on the top step (Fig.4.12.). The cameras residual values were checked to be less than 0.75mm prior to accepting the calibration (Fig.4.13.), if 0.75mm was exceeded by any camera the dynamic wand calibration was repeated.

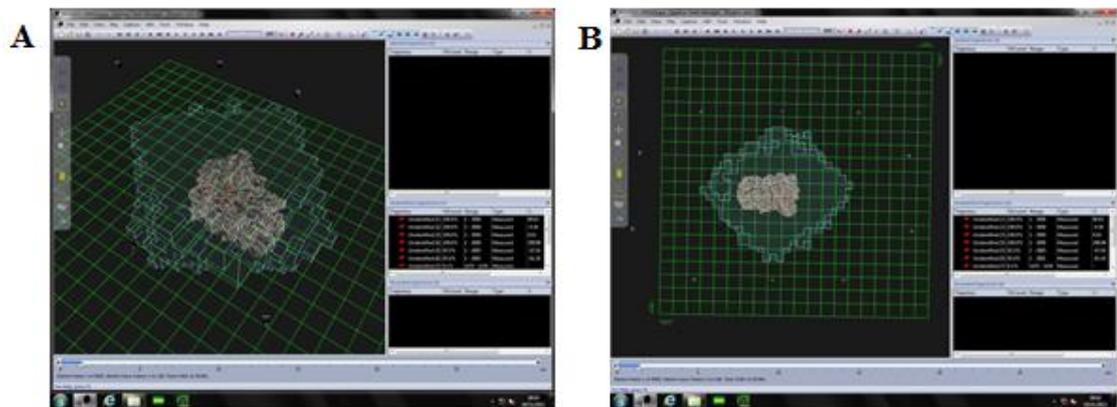


Fig.4.12. A and B show screen shots of the calibration area and the volume covered by the cameras.

A single marker was then placed at the corner of each force plate to define their positions. Kinetic data were recorded at 200 Hz using four AMTI force platforms (Advanced Mechanical Technology Inc. BP-400600, Watertown, MA. USA). The raw kinematic and kinetic data was initially synchronised and checked in Qualisys Track Manager Software (Sweden). The results from this were exported to Visual 3D Software (Version 4.87.0. C-Motion Inc., USA) where the data were filtered using fourth order Butterworth filters set at 6 Hz for the kinematic data and 25Hz for the kinetic data. This produced a dynamic visual representation and carried out all the calculations to formulate a report template.

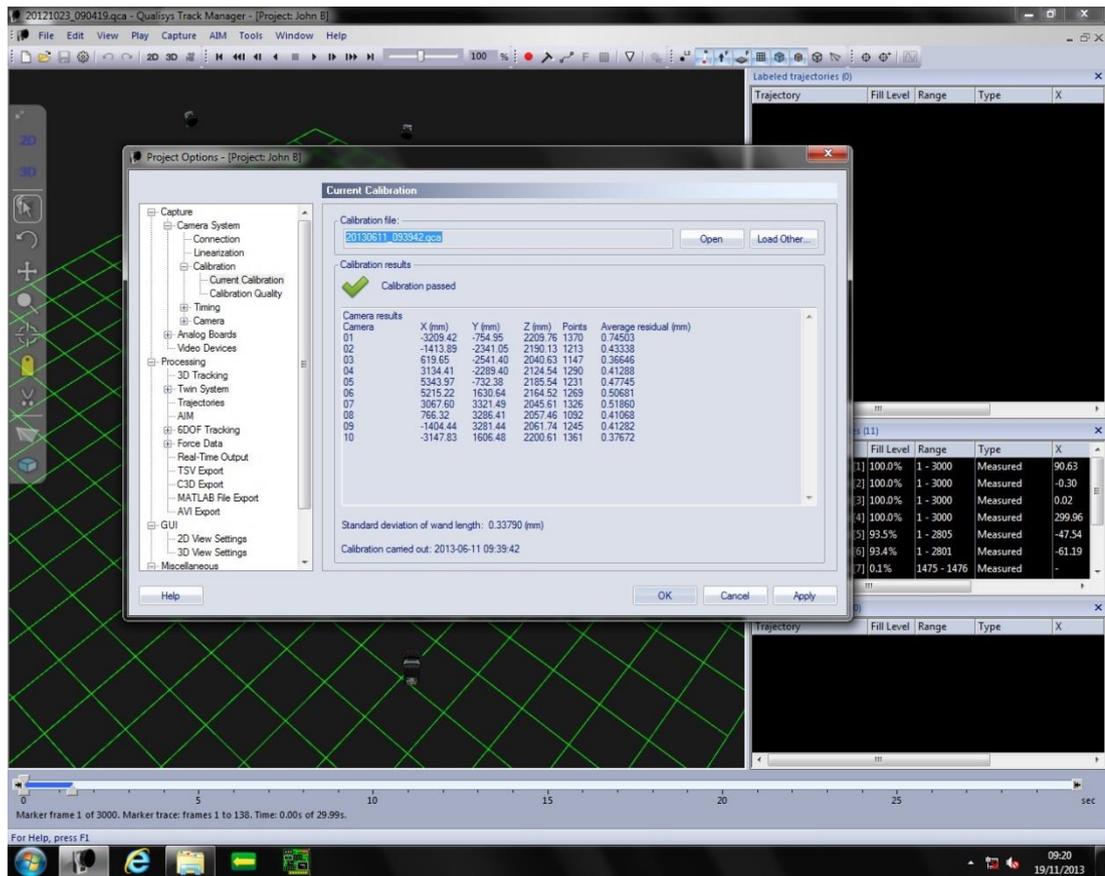


Fig.4.13. Screen shot of typical residual values from the cameras at set up.

4.7.1. Outcomes

It was not known whether it was possible to pick up small alterations in the kinematics and kinetics of multi-segment foot motion within footwear using the method previously described. Due to the fact there had to be two marker sets: one attached to the foot in the sandals and one directly on the shoes; this limited the comparisons to range of motion, as differences in the marker positions would affect maximum positions. As the proximal markers for the toe segment were also the distal markers of the metatarsal segment, only the sagittal plane motion was to be recorded. It is surmised by clinicians that medially wedged orthoses have their greatest effect in the coronal plane; hence the metatarsal and calcaneal segments were split into contact and propulsive phases. The whole of stance phase was to be reported for the sagittal and transverse planes. Similarities and differences between the footwear conditions were reported to give a balanced view of what was recorded.

4.7.2. Data Filtering

Woltring, (1985) stated there are a number of problems associated with filtering data, not least that the filters operate under the supposition that the signals being processed are stationary, which is certainly not the case when looking at walking. Erer, (2007) noted that it is inevitable that motion capture systems introduce errors into the trajectories of the markers being tracked, this means these raw trajectories have to be filtered before any meaningful data can be presented. Richards, (2008) explained that filtering is applied to the coordinates of each marker position removing modest digitizing errors, it was suggested that low pass filters are commonly used such as a second or fourth order Butterworth filter with a cut off frequency of 6 or 7 Hz; this leaves the low-frequency data alone but removes the high-frequency data.

4.7.3. Setting Events

When considering the stance phase of walking using computer gait analysis it usual to set the events of gait by using a small impact force of 8N to define heel strike. The end of propulsion is set at the point which the force diminishes to a similar value; this was how the walking trials were defined for this experiment. During the step down trials it was possible to repeat this protocol for when the “impact limb” contacted the lower step, but as the subjects were asked to stand still momentarily on the top step an impact force was unavailable. The initial step descent event was set in Visual 3D not by using force but by taking the highest hip position as the subjects were starting on the top step. This was then fine-tuned manually for each step trial by looking at the ground reaction force which moves towards the stance limb and tended to pass through the medial malleolus just before the swing foot lifted off the step. The initial event was set at one frame before the “swing limb” heel lifted.

4.8. Statistical Methods

Once the data had been processed in Visual 3D the mean and standard deviation data were exported as an ASCII file and imported into Microsoft ExcelTM. Maximum and minimum values were extracted for each parameter, subject and footwear condition; joint moments were also calculated before being brought into SPSS (version 20.0) to carry out all statistical calculations.

The data from the normal subjects was subjected to linear mixed model analysis by giving numeric labels to the subjects and the conditions, enabling the comparisons to be “targeted” (Fig.4.14.). This allowed the mean values obtained from the shoe only trials to be compared with the mean sandal only data. To determine if the orthoses were introducing any effects they were compared to the combined footwear conditions i.e. mean combined shoe and sandal only data (footwear) to ¾ orthoses in both shoes and sandals, footwear alone to FL orthoses in both types of footwear. Separate analyses were run on the walking and step down trials.

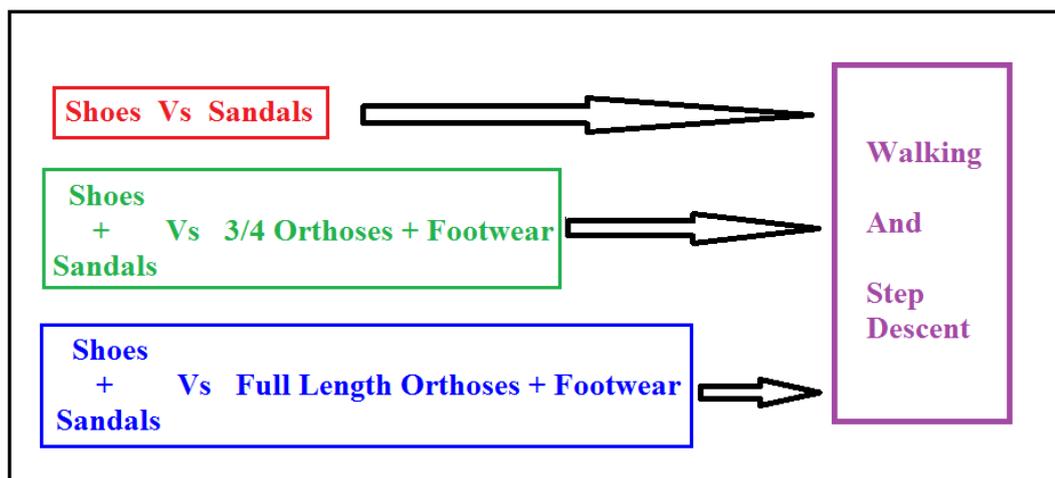


Fig.4.14. Diagram representing statistical comparisons and how they were “targeted”.

4.9. Sample Size

It was not possible to perform a sample size calculation as no data for multiple segment foot mechanics with and without orthotic management existed. However, Nester et al (2003) found a mean difference of 3 degrees in rearfoot movement with and without foot orthoses with a standard deviation of 3 degrees; this yields a sample size of 16 at a significance level of 5% with a power of 80%. It is very likely the parameters used in this current study will be more sensitive to the application of foot orthoses, therefore a sample size of 15 was used to explore the between condition effects.

Chapter 5. Results Non-symptomatic Subjects

5.1. Initial Results

All fifteen healthy subjects (7 males and 8 females, mean age 30.06 SD 9.99, mean weight 72.1 Kg SD 17.5Kg) with no lower limb pain undertook five walking and step down trials. Trials were required for each footwear and orthoses condition and accepted as long as they hit centrally on the force plates, or descended with the same limb.

5.1.1. Calcaneal segment with respect to the tibial segment - walking

In the sagittal plane the traces produced during walking were seen to be similar between subjects therefore it was decided to just record the total range of motion of the calcaneal segment on the tibial segment. The coronal plane exhibited greater variation between subjects; it was felt that it was important to capture the range of eversion in the contact phase of stance and then the range of motion into or towards inversion during propulsion. The transverse plane demonstrated the greatest variability between subjects, mostly the calcaneal segment was abducted on the shank; the overall pattern tended to be abduction reducing towards at toe off (Fig. 5.1.) it was decided to just record the overall range of motion.

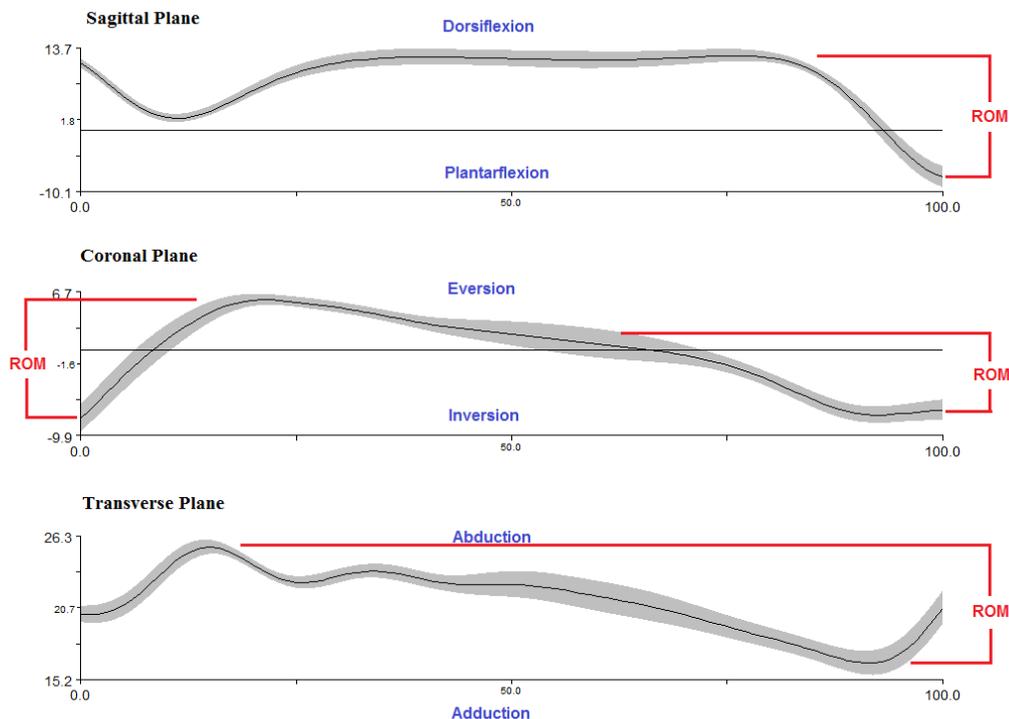


Fig. 5.1. Typical traces of rearfoot motion with respect to the shank seen in all three planes; each trace being an ensemble of 5 trials. The measurements recorded are highlighted.

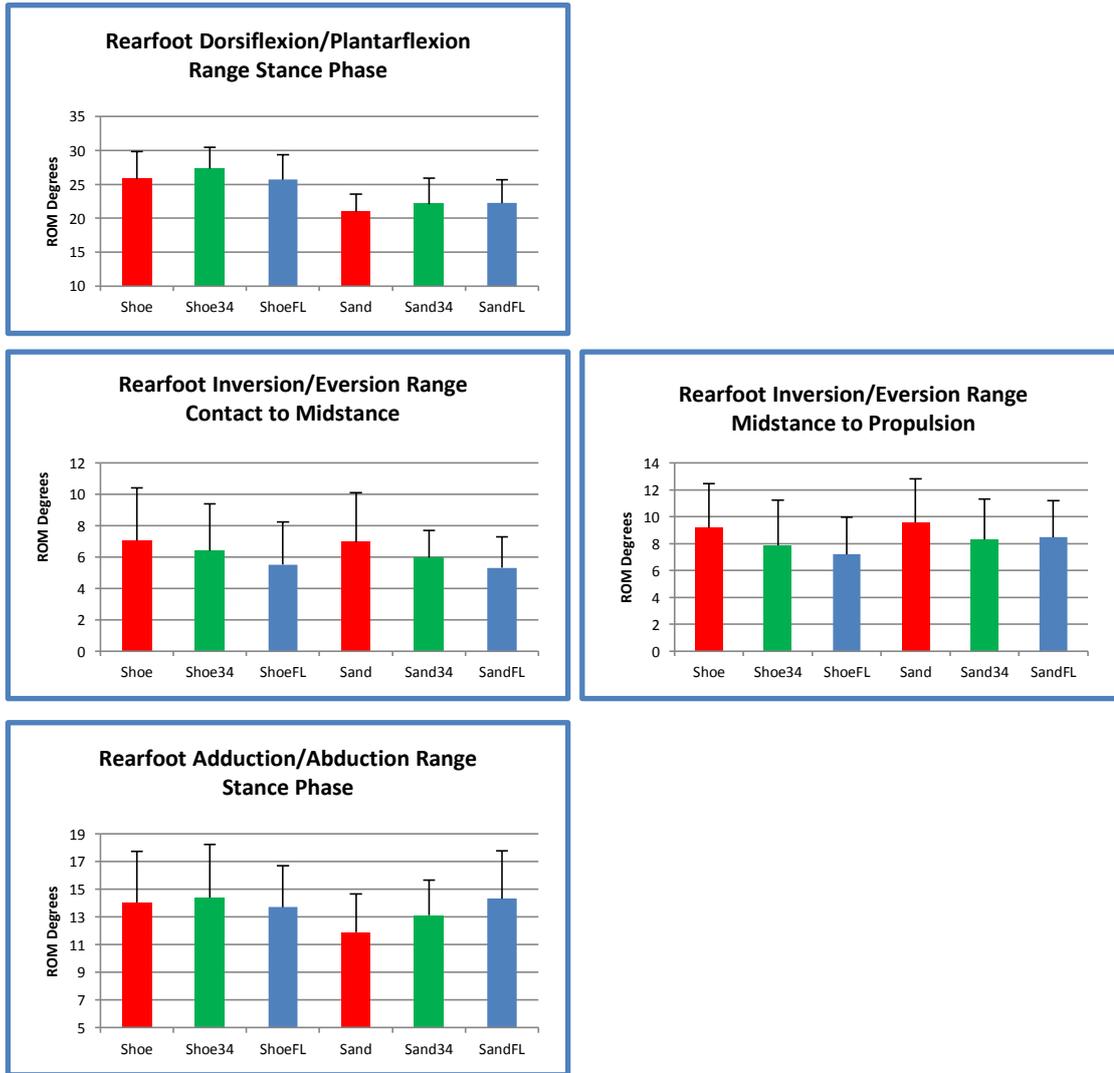


Fig.5.2. Bar charts illustrating mean ranges of motion of the calcaneal segment with respect to the shank between footwear conditions with and without orthoses.

In the sagittal plane the shoes did increase the range of dorsiflexion/plantarflexion significantly over the sandals ($p=0.000$ Table 5.1.) the mean difference was over 4° between the conditions. During early stance in the coronal plane there was no significant difference seen between the shoes and the sandal conditions (Table 5.1.), the same non-significant result was also seen from midstance to propulsion where there was less than 1° between the footwear types. The shoe condition did increase the range of mean rearfoot rotation over the sandal condition, the mean difference being just over 2° ($p=0.022$ Table 5.1.).

In the sagittal plane the $\frac{3}{4}$ orthoses also produced a further significant increase in the range of dorsiflexion/plantarflexion over the combined footwear only condition; this difference was less than that seen between the footwear conditions. The full length orthoses did not induce a significant result.

Only the full length orthoses produced a significant reduction in coronal plane range during early stance phase ($p=0.002$); the difference in mean range was 1.4° which is about 20% of the total range. The $\frac{3}{4}$ orthoses was not significant ($p=0.100$) however the shift in the confidence intervals (CI) suggested there was also a trend towards reduction (Table 1 Appendix 2).

During midstance to propulsion both the $\frac{3}{4}$ and full length orthoses significantly reduced the mean coronal range by 1.3° and 1.4° respectively ($p=0.019$ and $p=0.006$ Table 5.1.). When looking at the bar charts in Fig.5.2. the overall pattern is repeated between the sandal and shoe conditions in both early and late stance suggesting there were similar reactions to the orthoses with both footwear conditions.

When considering the transverse plane range neither orthoses produced a significant result when compared to the combined footwear only condition, however the full length orthoses result ($p=0.064$ Table 5.1.) would suggest there was a definite trend towards increasing the range of motion. It was apparent from Fig. 5.2. that this result was primarily due to the orthoses increasing the range of motion over the sandal only condition.

Summary:

- The shoes increased the mean range of motion in both the sagittal and transverse planes compared to the sandals. No differences occurred in the coronal plane.
- The $\frac{3}{4}$ orthoses significantly increased the sagittal plane range of motion over the footwear only condition; however this was less than the difference seen between the 2 footwear types.
- In the coronal plane the full length orthoses reduced the range of motion significantly from contact to midstance. Both orthoses produced a significant reduction in the later phase of stance.

- There were no significant reductions in the transverse plane though there was a trend noted with the full length orthoses; this was more apparent when the full length orthoses was combined with the sandals.

Table 5.1. Mean range of motion of the calcaneal segment with respect to the shank; highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Calcaneal ROM with respect to the shank	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM 3/4 orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	3/4	FL
dors/plant	25.8(4.0)	21.0(2.6)	P=0.000	27.3(3.2)	22.1(3.9)	25.7(3.7)	22.3(3.5)	P=0.042	P=0.359
inv/ev early stance	7.0(3.4)	7.0(3.2)	P=0.557	6.4(2.8)	6.0(1.8)	5.5(2.7)	5.3(2.0)	P=0.100	P=0.002
inv/ev late stance	9.2(3.0)	9.6(3.3)	P=0.121	7.9(3.1)	8.3(3.1)	7.2(2.7)	8.5(2.8)	P=0.019	P=0.006
ab/add	14.1(3.3)	11.9(2.7)	P=0.022	14.1(3.3)	13.1(2.5)	13.7(3.1)	14.3(3.5)	P=0.163	P=0.064

5.1.2. Metatarsal segment with respect to the calcaneal segment - walking

The traces for the metatarsal segment in the sagittal plane did demonstrate a little variance between subjects compared to the calcaneal segment results Fig. 5.3. shows a more representative trace from a single subject. However some subjects did have a small movement towards dorsiflexion during the first quarter of the stance phase; as the subjects with this early movement into dorsiflexion were in the minority it was decided just to record the overall range of motion.

The traces of the coronal plane were similar for the majority of subjects, most started close to neutral then demonstrated movement towards or into further eversion during the first half of stance. The metatarsal segment tended to move towards inversion from just before 50% of the trace peaking late in stance phase. There was then a rapid movement into or towards eversion just before toe off. As there seemed to be two definite phases it was felt important to capture the range of both of these and hence the coronal plane results were split into halves (see Fig. 5.3.).

There was more variation seen between the subjects in the traces of the transverse plane. All subjects started in varying amounts of adduction, this then reduced as the forefoot loaded, after remaining fairly constant through midstance there was a rapid

increase in adduction in the last 25% of stance Fig. 5.3. The final motion was considered to be the most important and this was captured by recording the overall range.

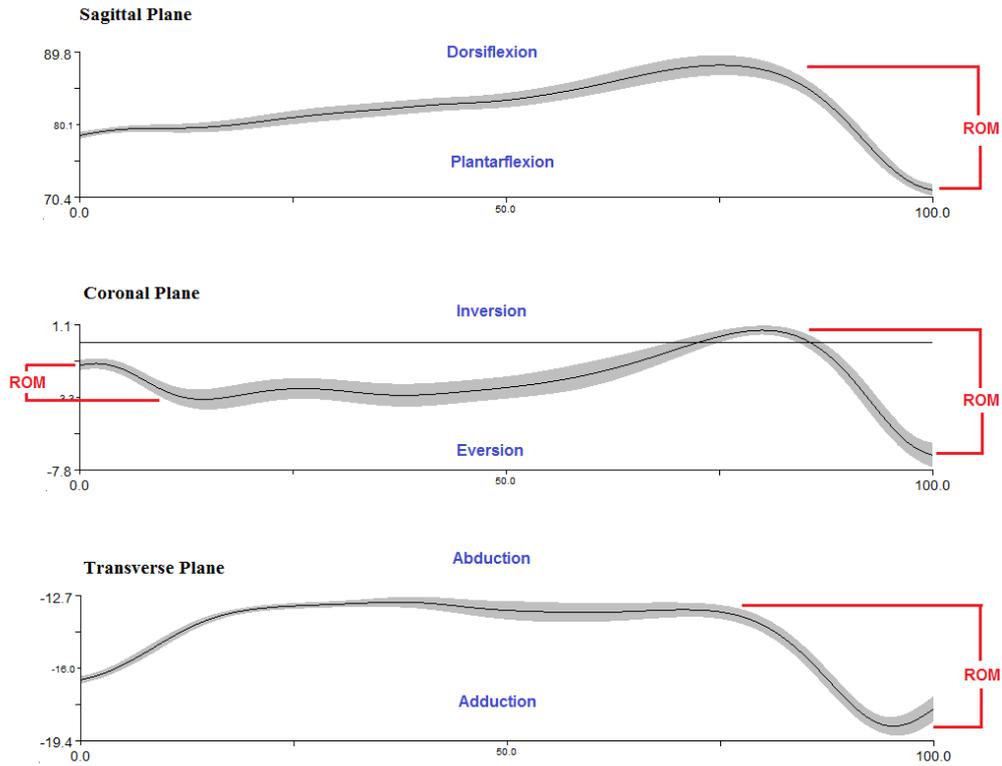


Fig.5.3. Typical traces of metatarsal segment motion with respect to the calcaneal segment seen in all three planes; each trace being an ensemble of 5 trials. The measurements recorded are highlighted.

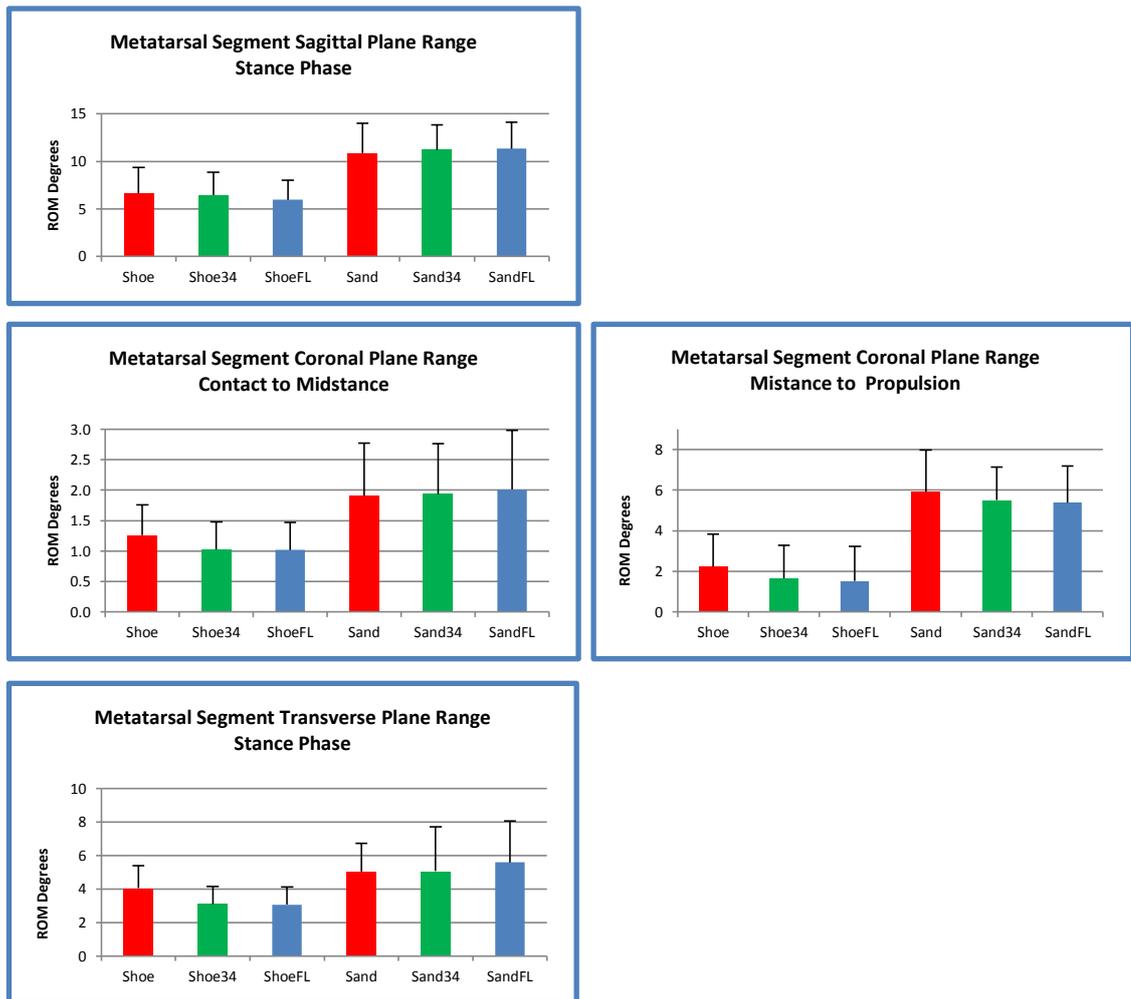


Fig. 5.4. Bar charts illustrating mean ranges of motion of the metatarsal segment with respect to the calcaneal segment between footwear conditions with and without orthoses.

When considering the range of motion of the metatarsal segment on the calcaneal segment the shoe condition did significantly reduce the range of motion in the sagittal plane over the sandal condition by about 4° ($p=0.000$, Table 5.2.). During early stance the shoes again significantly reduced the range of motion in the coronal plane by about half a degree ($p=0.000$). Through late stance the effect of the shoes was much more apparent (see Fig. 5.4.) reducing the mean range of motion over the sandals by over 3° producing a significant result ($p=0.000$). The range of abduction/adduction in the transverse plane was also significantly reduced compared to the sandals; the mean difference was 1° which equated to a 20% reduction.

Neither of the orthoses had any effect over the footwear only condition in the sagittal plane and this is illustrated in Fig. 5.4. The same non-significant result was repeated during early stance phase in the coronal plane; both orthoses reduced the range minimally when combined with the shoes; but tended to produce minimal increases with the sandals (see Table 5.2.).

During midstance to propulsion both the orthoses reduced the range of inversion/eversion significantly over the footwear only conditions. The reductions seen in the mean data were small 0.4° - 0.7° suggesting that the reductions were seen in all subjects. The orthoses did not demonstrate any significant changes over the footwear only conditions in the transverse plane, however the $\frac{3}{4}$ orthoses did demonstrate trend in the confidence intervals (Table2 Appendix 2). The bar chart in Fig.5.4. shows the orthoses in the sandals increased the range minimally, but when combined with the shoes they both demonstrated a small mean reduction.

Summary:

- The metatarsal segment demonstrated it was the shoes that had the dominant effect over the sandal condition; reducing the measured ranges significantly in all planes.
- Only during late stance phase did the orthoses have a consistent effect in both types of footwear and hence produced a further significant if small reduction.
- In the sagittal, early stage coronal and the transverse planes the orthoses tended to produce very minimal increases when combined with the sandals and minimal decreases with the shoes leading to non-significant results.

Table 5.2. Mean range of motion of the metatarsal segment with respect to the calcaneal segment highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear only conditions combined.

Metatarsal ROM with respect to the calcaneal segment	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM 34 orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	34	FL
dors/plant	6.6(2.4)	10.8(2.9)	P=0.000	6.4(1.8)	11.2(2.4)	6.0(1.6)	11.3(2.8)	P=0.817	P=0.818
inv/ev early stance	1.3(0.5)	1.9(0.8)	P=0.000	1.0(0.5)	1.9(0.8)	1.0(0.4)	2.0(1.0)	P=0.423	P=0.688
inv/ev late stance	2.2(0.8)	5.9(1.8)	P=0.000	1.7(0.7)	5.5(1.4)	1.5(0.6)	5.4(1.8)	P=0.032	P=0.007
Abd/add	4.0(1.3)	5.0(1.7)	P=0.000	3.1(1.0)	5.1(2.6)	3.1(1.0)	5.6(2.4)	P=0.163	P=0.515

5.1.3. Phalangeal segment with respect to metatarsal segment - walking

The traces produced from five combined trials for each subject were very consistent though there were some differences seen between subjects mainly in magnitude rather than shape. It was decided not to report on the coronal and transverse planes in the phalangeal segment as it was felt by using the markers over the first and fifth metatarsophalangeal joint to define the distal end of the metatarsal segment they may not have demonstrated the true range.

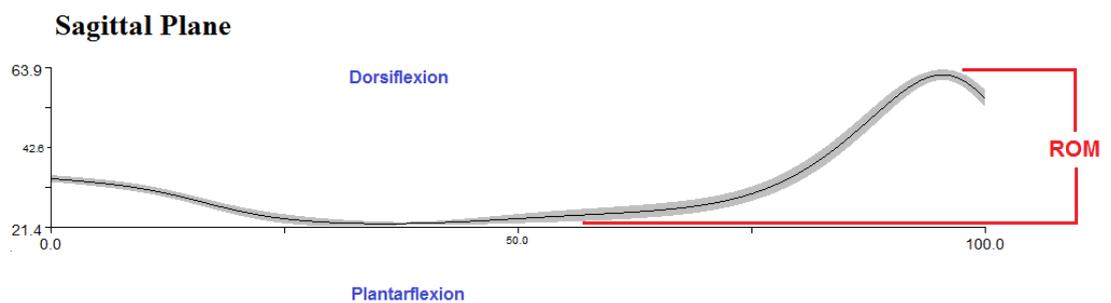


Fig. 5.5. A typical trace of phalangeal segment motion with respect to the metatarsal segment in the sagittal plane; the trace is an ensemble of 5 trials. The measurement that was recorded is highlighted.

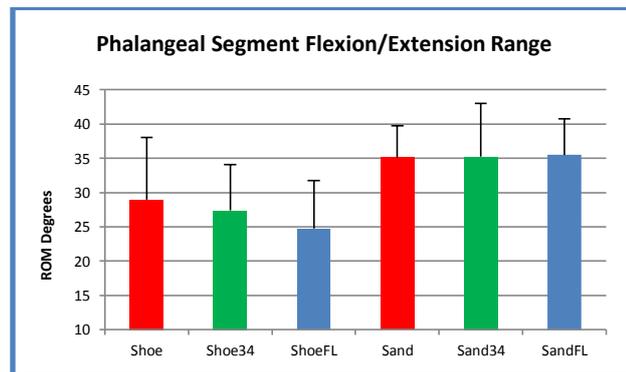


Fig. 5.6. Bar chart illustrating mean ranges of motion of the phalangeal segment with respect to the metatarsal segment between footwear conditions with and without orthoses.

The shoes did reduce the mean range of sagittal plane motion of the phalangeal segment with respect to the metatarsal segment when compared to the sandals by a little over 6° (p=0.000, see Table 5.3.) this was illustrated by the bar chart in Fig. 5.5.

When combined with the shoes both orthoses reduced the range of motion over the shoe only condition, while with the sandals the orthoses both tended to leave the mean range unaltered. When the footwear conditions were combined the full length orthoses still induced a significant reduction in the range of motion, which will be as a result of the 4.2° reduction seen within the shoe condition, the ¾ orthoses result was not statistically significant.

Summary:

- The shoes significantly reduced the measured range of motion over the sandal only condition.
- Further reductions were seen when both orthoses were combined with the shoes, these were not repeated in the sandals.
- The reduction with the full length orthoses was substantial enough to produce a significant overall reduction.

Table 5.3. Mean range of motion of the metatarsal segment with respect to the calcaneal segment in the sagittal plane highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Phalangeal ROM with respect to the metatarsal segment	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM ¾ orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	¾	FL
dors/plant	29.0(6.3)	35.1(3.7)	P=0.000	27.4(6.1)	35.2(3.9)	24.8(5.8)	35.5(4.0)	P=0.424	P=0.041

5.1.4. Knee kinematics - walking

Sagittal plane traces for all subjects were very similar in shape and demonstrated good repeatability for each subject, to get a true picture of flexion/extension it was felt necessary to record knee position at heel strike, impact peak (first peak) and maximum extension (which in nearly all cases was still a flexed position). Maximum flexion at toe off was also recorded with the total range of motion during the stance phase (see Fig. 5.7.).

It was abduction and adduction of the shank on the thigh that demonstrated the smallest range of motion, relatively there was more variation within and between subjects. Because the markers remained constant between the sandal and shoe conditions maximum and minimum positions were able to be reported in addition to range of motion. Some subjects did not post an adduction value remaining abducted throughout the stance phase (unlike the subject in Fig. 5.7.). Maximum adduction was recorded but for some subjects they remained in a small amount of abduction. It was therefore decided that reporting the maximum abduction angle and the total range of motion would capture the variation seen within maximum “adduction” (Maximum adduction Fig.A2.1. Appendix 2).

The rotation of the shank on the femur in the transverse plane posted a mean value almost three times as great as the coronal plane. In most subjects the shank was externally rotated on the femur at heel contact, the subject’s graph in Fig. 5.7. illustrates the shank does not attain an internal position on the femur through the stance phase. This was seen in a number of subjects almost half attained an internal position; this was also variable between conditions. Due to the values for internal rotation being so inconsistent between subjects it was decided to only report maximum external position and range of motion, it was believed the variability in the maximum internal position would be reflected in the range (maximum internal rotation Fig.A2.2. Appendix 2).

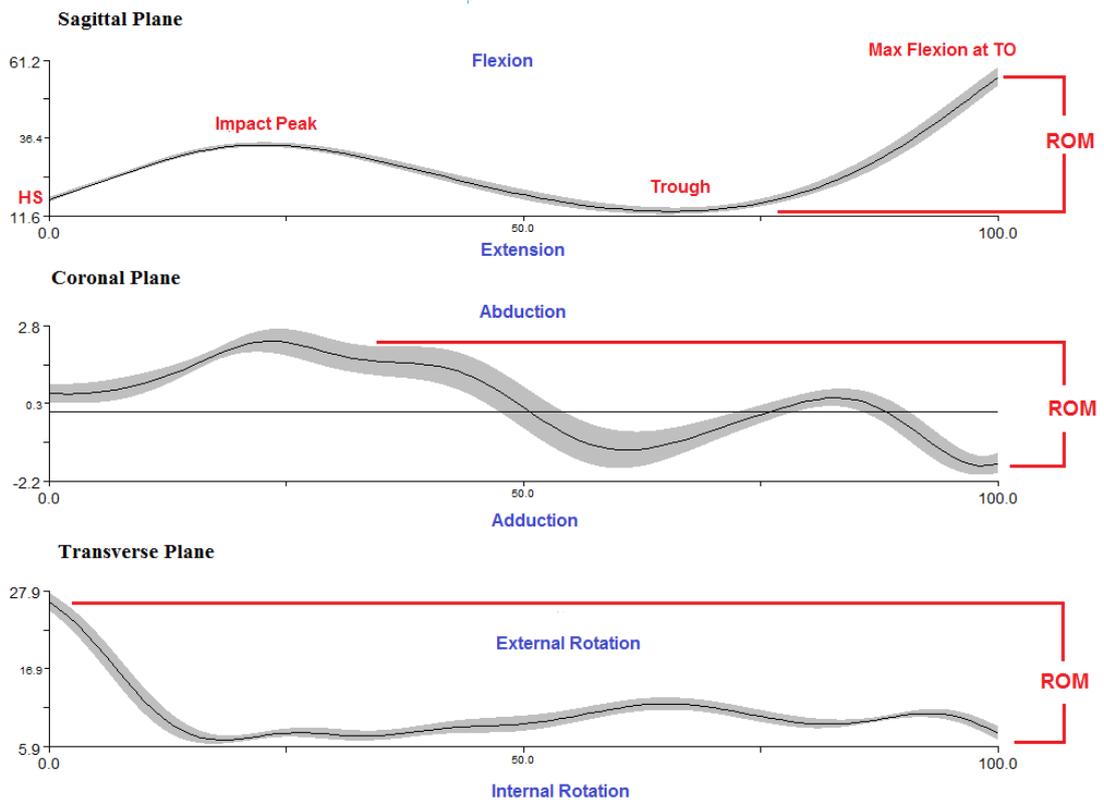


Fig.5.7. Typical traces of knee motion seen in all three planes; each trace being an ensemble of 5 trials. The measurements recorded are highlighted

The shoes only significantly reduced the maximum mean knee flexion position over the sandals at heel strike, minimum flexion angle (trough) and maximum flexion angle at toe off ($p=0.000$, $p=0.004$, $p=0.001$ respectively Table 5.4.). The type of footwear made no difference to the impulse peak ($p=0.319$) similarly the total range of flexion was not significantly altered by the shoes over the sandals.

There was no significant difference seen between the shoes and the sandals in maximum abduction but the shoes did produce a significant increase over the sandals when the range of coronal plane motion was considered ($p=0.033$ Table 5.4.). There were no significant differences between the shoe and sandal conditions when considering maximum external rotation and range of rotation at the knee (Table 5.4.).

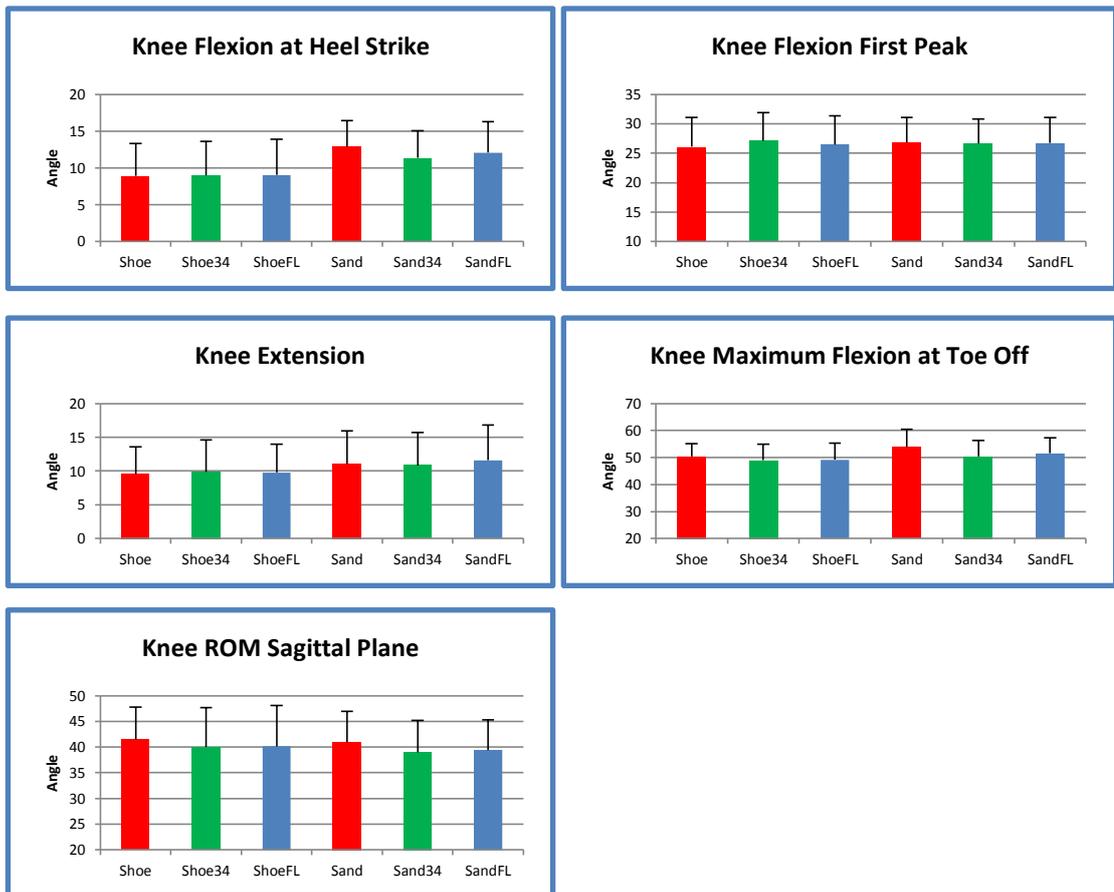


Fig.5.8. Bar charts illustrating mean ranges of motion of the knee in the sagittal plane between footwear conditions with and without orthoses.

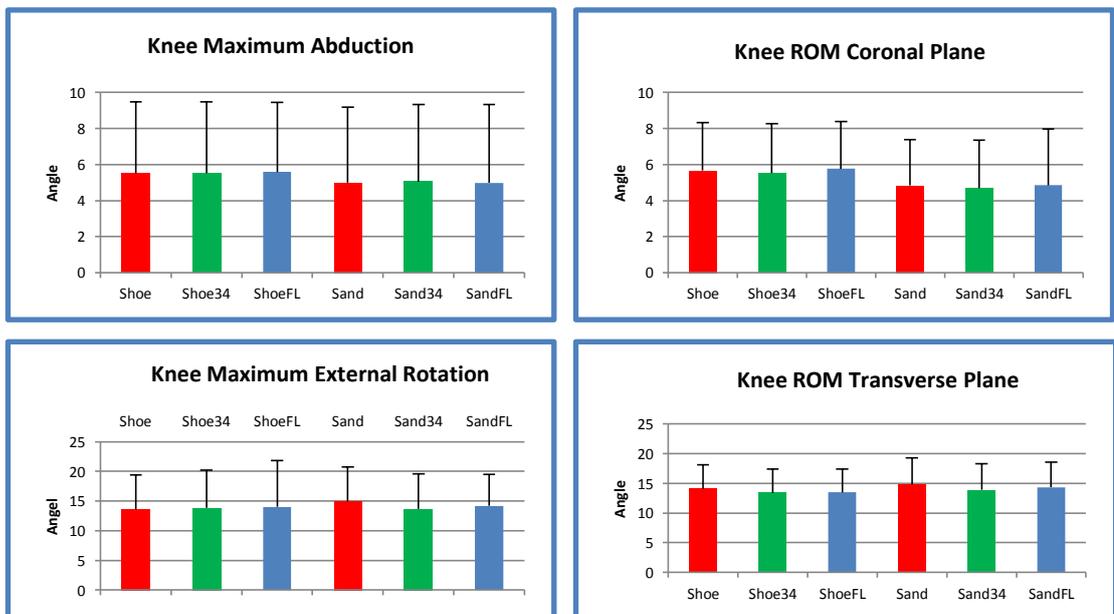


Fig.5.9. Bar charts illustrating mean ranges of motion of the knee in the coronal and transverse planes between footwear conditions with and without orthoses.

Neither orthoses had a significant effect on knee flexion at heel strike, when combined with the sandals the mean position tended to be reduced a little, however this was not repeated when they were combined with the more supportive shoes (see Fig. 5.8.). The first peak of knee flexion was generally unaltered by the orthoses there was an increase of one degree seen with the $\frac{3}{4}$ orthoses when combined with the shoes compared to the shoes alone but all other combinations produced mean results within one degree this is illustrated by the bar chart in Fig. 5.8.

The mean knee extension between subjects varied from 0.6° to 20.4° of flexion, but when the subject data was combined, it was not significantly affected by the orthoses as can be seen by the bar chart in Fig. 5.8. Maximum knee flexion at toe off was reduced by both orthoses whether combined with the shoes or the sandals, the $\frac{3}{4}$ orthoses produced a significant result ($p=0.022$ Table 5.4), the bar chart in Fig.5.8. suggested that the main effect was in the sandals. The full length orthoses were just outside the level of significance ($p=0.083$) this trend was reflected in the confidence intervals.

The orthoses reduced the range of flexion when combined with both the sandals and the shoes, but as the differences recorded were so small (2.0° max) they were not great enough to be significant. The $\frac{3}{4}$ orthoses did reveal a mean difference of 1.7° , the confidence intervals suggested that there was a trend towards reduction, as the maximum reduction was just over 4° while the minimum was an increase of 0.7° over footwear only condition (Table3 Appendix 2).

Neither orthoses exhibited any effect on the maximum abduction angle as illustrated by the bar chart (Fig. 5.9.). When looking at the total range of coronal plane motion again there were no significant results as the alterations in the mean differences were far too small. (Table 5.4.) The orthoses did not induce any significant differences in the transverse plane; the results were fairly consistent there were minimal reductions seen in the range of rotation at the knee with both. The maximum external position demonstrated minor increases when combined with the shoes but similar minor decreases with the sandals (See Fig.5.9.).

Summary:

- The shoes significantly reduced knee flexion angle over the sandals at heel strike, maximum extension and maximum flexion at toe off, the impulse peak and overall range of motion were unaffected.
- Maximum abduction was not affected by the type of footwear but the coronal plane excursion was significantly increased by the shoes over the sandals.
- Only the ¾ orthoses produced a significant reduction in flexion at maximum flexion at toe off. This was mainly due to the reduction seen in the sandals.
- There were no significant differences produced by the orthoses in either the coronal or transverse planes.

Table 5.4. Mean range of motion of the knee highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Knee angle (degrees)	Mean Position Shoe	Mean Position Sandals	Significance Shoe Vs Sandal	Mean Angle 34 orthoses		Mean Angle FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	34	FL
Flexion at HS	8.6(4.2)	13.2(3.4)	P=0.000	8.5(4.4)	11.8(3.6)	8.6(4.7)	12.3(4.0)	P=0.318	P=0.519
Flexion 1 st Peak	26.0(4.9)	26.9(4.4)	P=0.319	26.9(4.7)	26.9(4.2)	26.2(4.9)	27.3(4.7)	P=0.587	P=0.712
Max Extension (trough)	9.3(3.7)	11.3(5.0)	P=0.004	9.4(4.3)	11.3(5.1)	9.4(3.9)	11.4(5.0)	P=0.969	P=0.916
Max Flexion at TO	49.9(4.6)	54.3(6.5)	P=0.001	48.6(5.9)	50.8(5.7)	49.0(6.0)	51.5(6.2)	P=0.022	P=0.083
ROM Sagittal	41.3(6.4)	41.1(5.9)	P=0.399	40.1(7.7)	39.0(6.2)	40.5(7.8)	39.2(6.0)	P=0.163	P=0.262
Max Abduction	5.5(4.0)	5.0(4.1)	P=0.387	5.5(4.0)	5.1(4.2)	5.6(4.0)	5.0(4.3)	P=0.955	P=0.955
ROM Coronal	5.7(2.6)	4.8(2.5)	P=0.033	5.5(2.7)	4.7(2.6)	5.8(2.6)	4.9(2.8)	P=0.707	P=0.905
Max external rotation	13.6(4.6)	15.0(5.4)	P=0.584	13.8(5.1)	13.6(5.7)	14.0(6.9)	14.2(5.3)	P=0.590	P=0.853
ROM transverse	14.1(3.8)	14.8(4.8)	P=0.289	13.9(3.9)	14.1(4.6)	14.3(3.7)	14.2(4.3)	P=0.282	P=0.477

5.1.5. Knee moments -walking

As the foot contacted the floor and the knee was extended the ground reaction force (GRF) passed anterior to the knee in the sagittal plane creating the maximum extension moment. As the knee flexed to its first peak the GRF increased and began to angle backwards to create its maximum flexion moment. From the first peak the knee extends into the trough and the GRF almost passes through the knee from this point to toe off, hence any moment created is small and varied between flexion and extension and was variable between subjects (see Fig. 5.10.).

In the coronal plane the knee moment tended to start in adduction it then increased a little before going to its maximum abduction moment within the first 10% of the stance phase. After reaching maximum abduction there was a rapid shift to the maximum adduction moment, the moment remained in adduction but lessened through single limb support to another peak as the contralateral limb was almost at heel contact. In a few subjects the second peak was higher, the moment then reduced into a small amount of abduction.

The transverse moment were the smallest recorded, the traces started around the neutral position, then some achieved an external rotation moment while others wavered around neutral for the first half of stance. From midstance the internal rotation moment increased to its maximum at about 75% of stance phase this then decreased to neutral towards toe off. To capture any changes in the moments it was deemed necessary to record both the minimum and maximum moments and also the range for each plane.

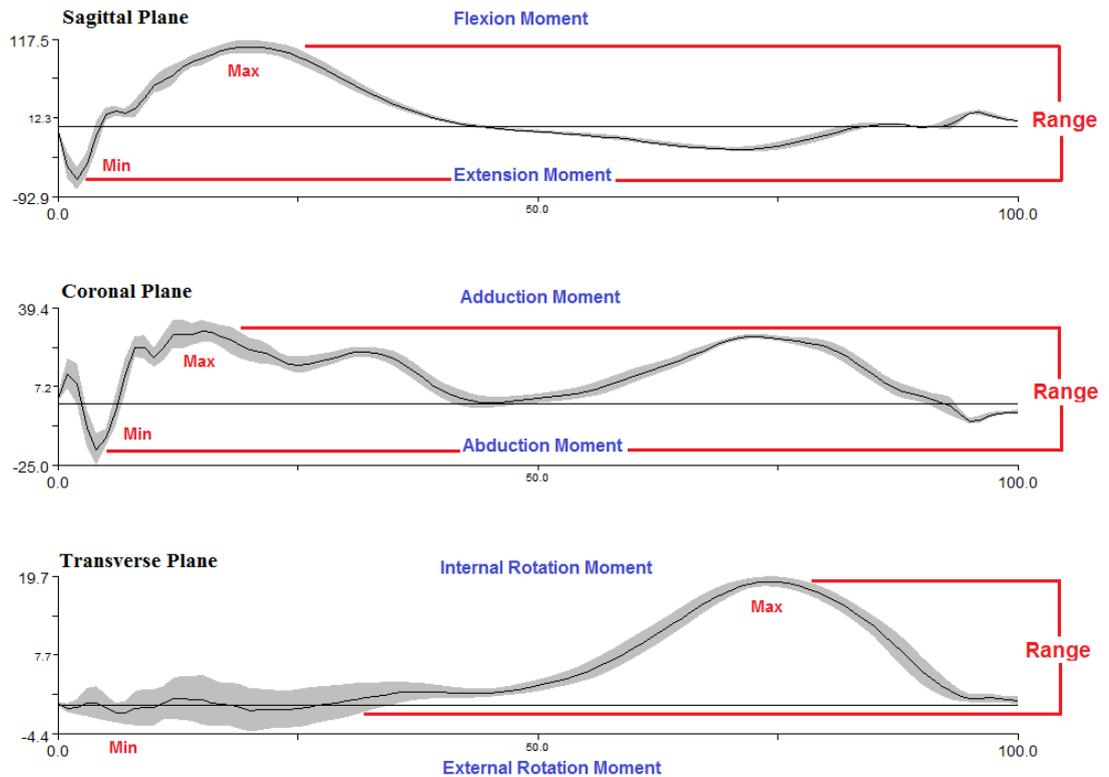


Fig.5.10. Typical traces of knee moments seen in all three planes; each trace being an ensemble of 5 trials, before normalization to body weight. The measurements recorded are highlighted.

The shoes tended to increase both the flexion and extension moments which in isolation were not significant, however when combined to give the moment range there was a significant increase seen ($p=0.047$). There were no significant changes in the coronal plane moments to report however there was a definite trend to an increase seen in the adduction moment ($p=0.078$) with the shoes over the sandals. In the transverse plane the shoes increased the external rotation moment significantly over the sandal only condition ($p=0.015$) but the larger internal moment and moment range were unaffected.

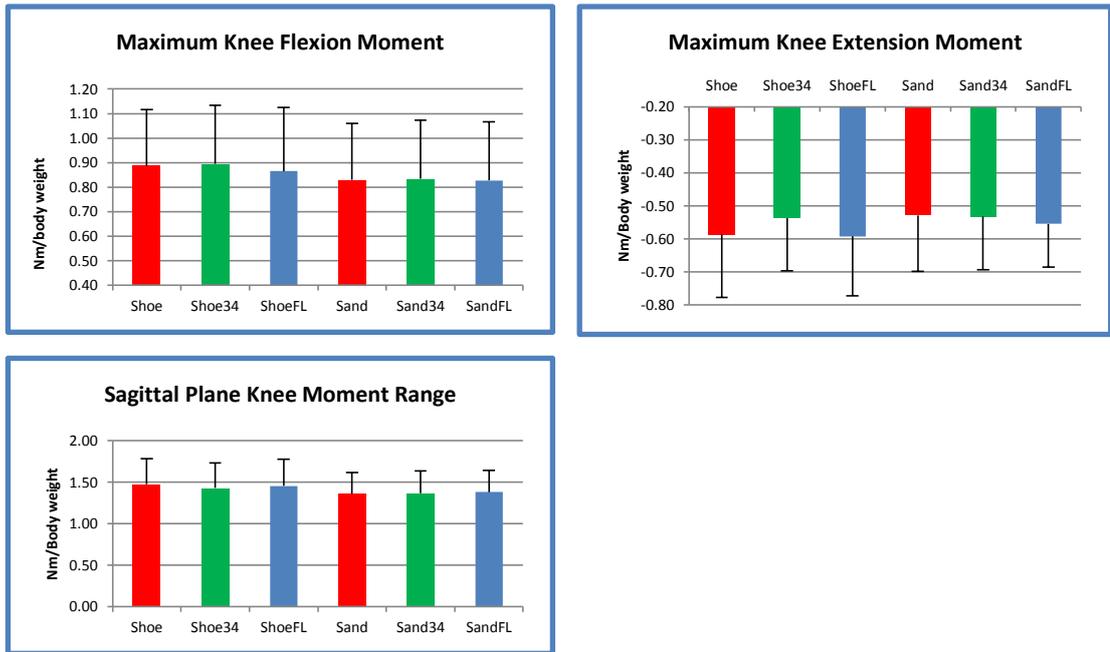


Fig.5.11. Bar charts illustrating mean ranges of motion of the knee in the sagittal plane between footwear conditions with and without orthoses.

The orthoses did not produce any significant changes to the moments of the sagittal plane (see Fig. 5.11.); the results obtained were fairly consistent and as such revealed no trends.

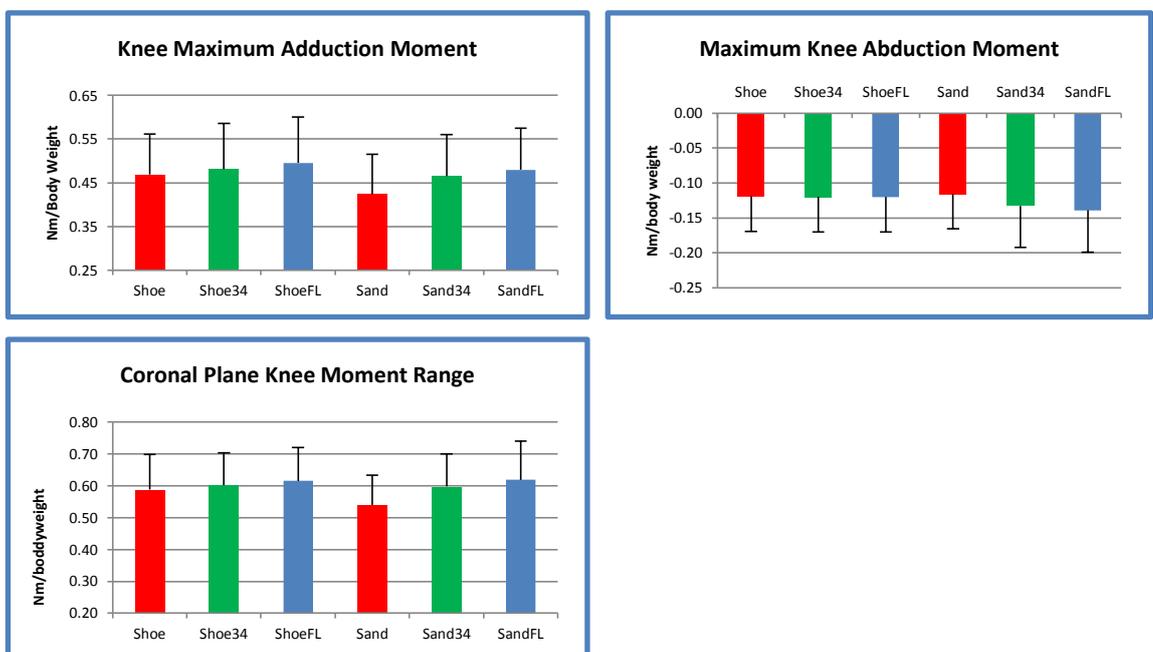


Fig.5.12. Bar charts illustrating mean ranges of motion of the knee in the coronal plane between footwear conditions with and without orthoses.

The adduction moment was significantly increased with the full length orthoses over the combined footwear condition ($p=0.022$ see Table 5.5.). The bar chart in Fig.5.12. would suggest this increase was more evident in the sandals. The $\frac{3}{4}$ orthoses were outside the level of significance; there was a mean difference of -0.027Nm/kg the confidence intervals displayed there was a shift in distribution towards increasing the moment (Table 4 Appendix 2).

The abduction moment was relatively unaffected by the orthoses in the shoes, but there was a small increase seen with both orthoses when combined with the sandals (Fig. 5.12.) however this was not enough to give a significant result. The full length orthoses effect on the adduction moment was carried over into the coronal moment range producing a significant increase ($p=0.007$), the $\frac{3}{4}$ orthoses was just outside the level of significance ($p=0.065$) but this trend towards an increase was reflected in the confidence intervals which were shifted well away from even distribution (Table 5 Appendix 2).

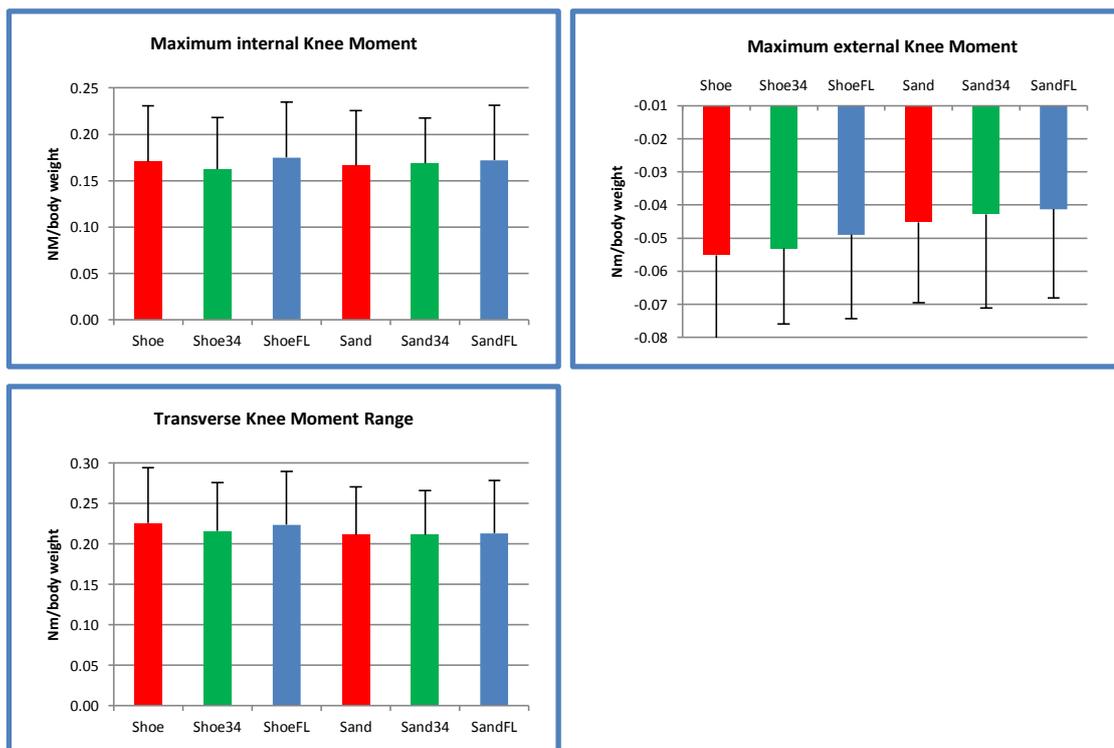


Fig.5.13. Bar charts illustrating mean transverse plane moments of the knee between footwear conditions with and without orthoses.

The bar charts in Fig.5.13. illustrates the orthoses had similar effects whether combined with the shoes or the sandals for all the transverse moment parameters, however none of the changes seen were significant.

Summary:

- The shoes significantly increased the sagittal plane moment range and the external rotation moment at the knee over the sandal only condition.
- The shoes did not produce any significant results in the coronal plane moments; however the adduction moment did demonstrate a trend towards being increased.
- The orthoses did not produce any significant results in either the sagittal or transverse planes.
- The adduction moment and coronal moment range were increased significantly by the full length orthoses, there were similar trends seen with the ¾ orthoses but they were not significant.

Table 5.5. Mean moments of the knee highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Knee Moments Nm/kg	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM ¾ orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	¾	FL
Flexion M	0.89(.23)	0.83(.23)	P=0.150	0.89(.24)	0.83(.24)	0.86(.26)	0.83(.24)	P=0.898	P=0.779
Extension M	-0.59(.19)	-0.53(.17)	P=0.176	-0.54(.16)	-0.53(.16)	-0.59(.18)	-0.55(.13)	P=0.441	P=0.610
Sagittal RO M	1.47(.31)	1.36(.26)	P=0.047	1.43(.30)	1.37(.27)	1.46(.32)	1.38(.26)	P=0.735	P=.952
Adduction M	0.47(.09)	0.42(.09)	P=0.078	0.48(.11)	0.47(.09)	0.50(.10)	0.48(.09)	P=0.129	P=0.022
Abduction M	-0.12(.05)	-0.12(.05)	P=0.255	-0.12(.05)	-0.13(.06)	-0.12(.05)	-0.14(.06)	P=0.385	P=0.237
Coronal RO M	0.59(.11)	0.54(.09)	P=0.305	0.60(.10)	0.60(.10)	0.62(.11)	0.62(.12)	P=0.065	P=0.007
Internal Rot ^a M	0.17(.06)	0.17(.06)	P=0.976	0.16(.06)	0.17(.05)	0.17(.06)	0.17(.06)	P=0.798	P=0.636
External Rot ^b M	-0.05(.03)	-0.04(.02)	P=0.015	-0.05(.02)	-0.04(.03)	-0.05(.03)	-0.04(.03)	P=0.635	P=0.278
Transverse RO M	0.23(.07)	0.21(.06)	P=0.303	0.22(.06)	0.21(.05)	0.22(.07)	0.21(.07)	P=0.668	P=0.990

5.1.6. Ankle moments - walking

In the sagittal plane, as the heel contacted the GRF passed behind the ankle giving a small plantarflexion moment as the foot plantarflexed towards foot flat and the knee flexed the GRF moved in front of the ankle joint axis creating a dorsiflexion moment. The moment remained level while the heel remained on the floor but as it lifted there was another increase of the dorsiflexion moment to its maximum. As the angle of the foot increased on the floor the GRF passed closer to the ankle joint centre and started to diminish as the weight was transferred to the leading limb.

Just after heel contact in the coronal plane, the GRF passed laterally to the ankle and the knee creating an evertor moment, again as forefoot loading was complete the GRF then passed just medial to the ankle to produce an invertor moment. The invertor moment increased as the heel lifted and the body weight was being transferred towards the other limb.

The transverse plane moments were the smallest measured at the ankle, the GRF tended to be very close to the centre of the ankle joint or just posterior and lateral at heel contact producing a small externally rotating moment. It was from midstance that the GRF passed anterior and medial to the ankle creating an internal rotation moment, this increased to its maximum as the GRF was angled more medially as weight transferred to the other limb.

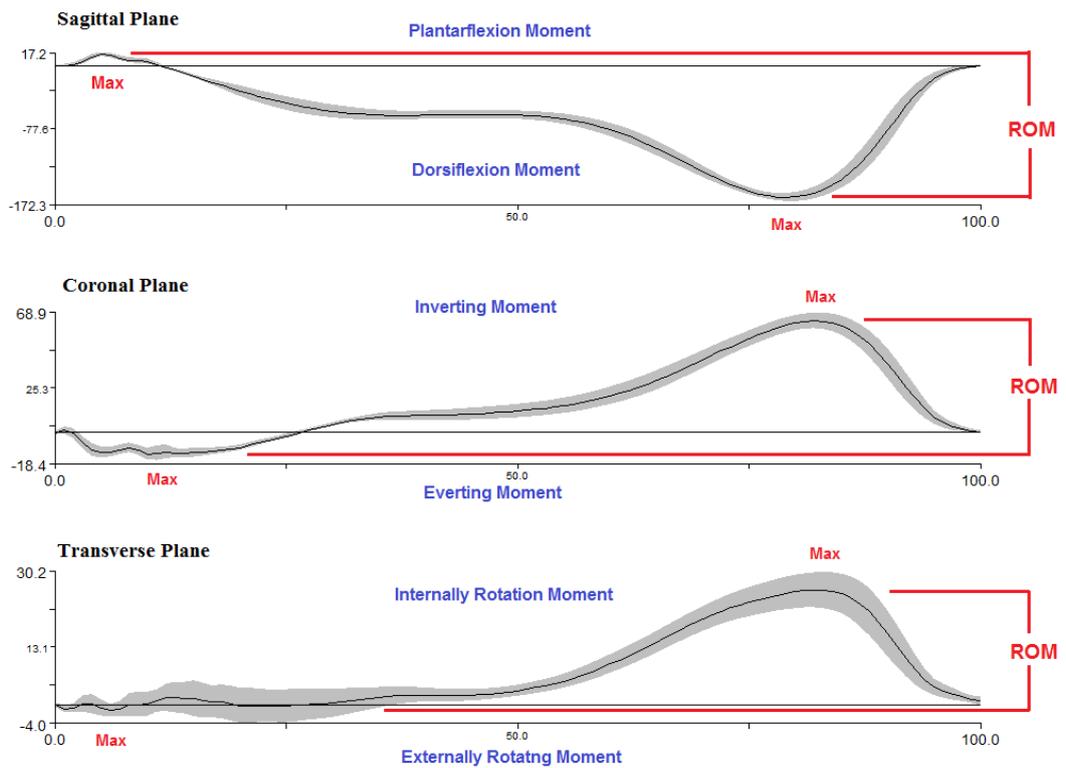


Fig. 5.14 Typical traces of ankle moments seen in all three planes; each trace being an ensemble of 5 trials. The measurements recorded are highlighted

In the sagittal plane the shoes increased both the plantarflexion and sagittal range of moment significantly over the sandal only condition ($p=0.000$ and $p=0.001$ respectively see Table 5.6.); the dorsiflexion mean was not significantly altered. The shoes significantly increased the inverting, everting and moment range in the coronal plane over the sandal condition ($p=0.040$, $p=0.000$ and $p=0.000$ respectively see Table 5.6.). In the transverse plane the shoes decreased the internal rotation moment significantly over the sandals ($p=0.000$) but both the maximum external moment and the moment range were increased significantly by the shoes ($p=0.000$ and $p=0.007$ respectively see Table 5.6.).

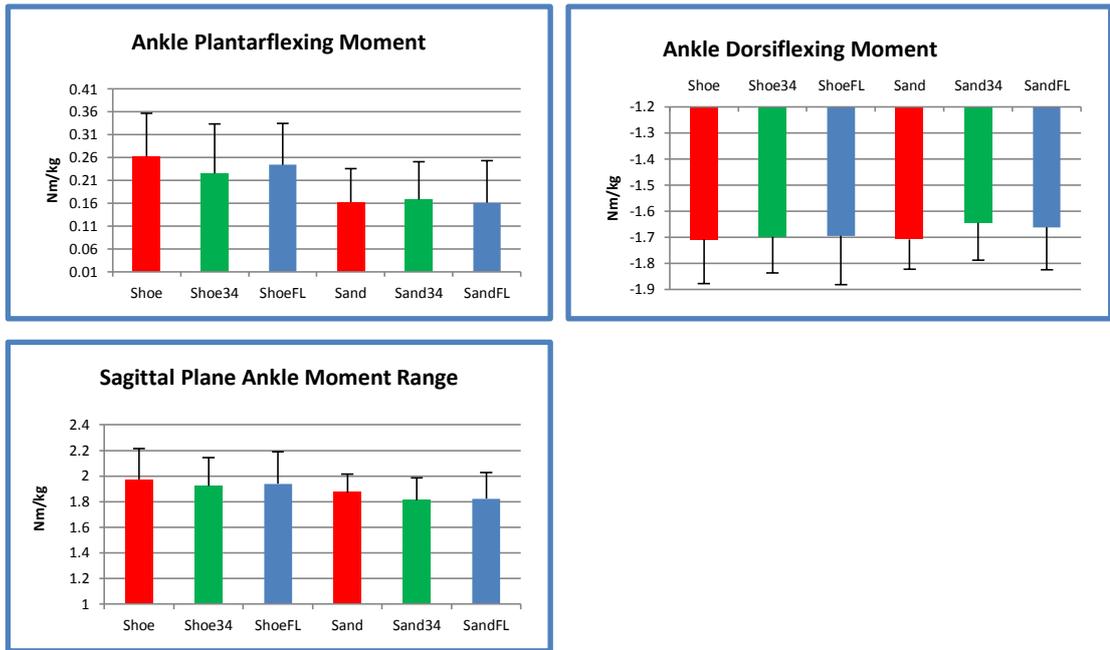


Fig. 5.15. Bar charts illustrating mean ankle moments in the sagittal plane between footwear conditions with and without orthoses.

The sagittal plane moments were generally unaffected by the orthoses, the bar charts in Fig. 5.15. illustrate that it was the footwear rather than the orthoses that was having an effect.

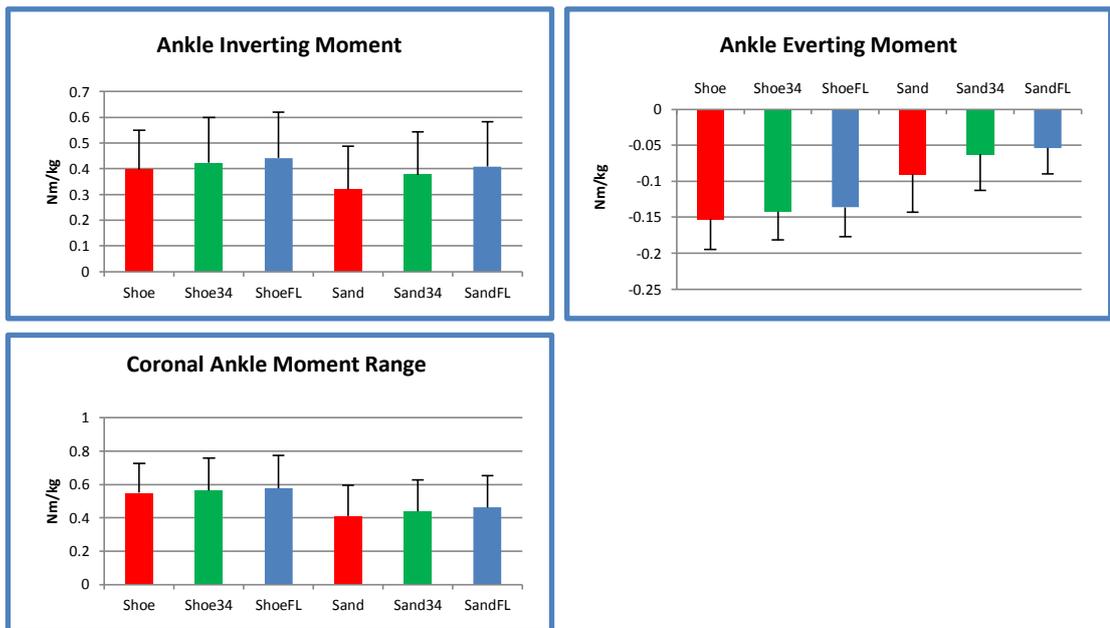


Fig.5.16. Bar charts illustrating mean ankle moments in the coronal plane between footwear conditions with and without orthoses.

The full length orthoses induced a further significant increase in the mean inverting moment over the combined footwear mean, the $\frac{3}{4}$ orthoses only produced a very small increase when combined with the shoes and only slightly more with the sandals therefore the result was not significant. However, the bar charts in Fig.5.16. would suggest that the effect of the orthoses was consistent no matter what footwear they were combined with. The everting moment was decreased significantly by both orthoses over the footwear only condition ($p=0.028$ and $p=0.000$ for the $\frac{3}{4}$ and full length orthoses respectively), the greatest reduction was seen with the sandals rather than the shoes (see Fig. 5.16.). With the inverting moment being generally increased by the orthoses and the everting moment being reduced the overall coronal moment range was not significantly affected.

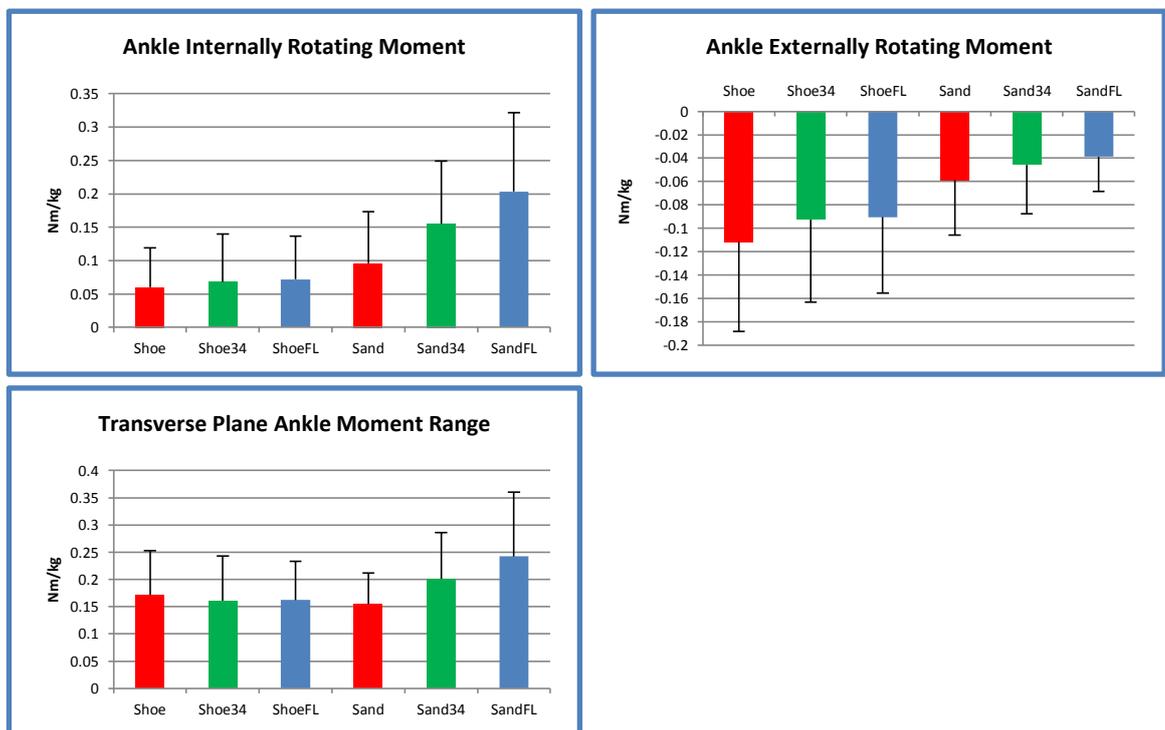


Fig.5.17. Bar charts illustrating mean ankle moments in the transverse plane between footwear conditions with and without orthoses.

Both orthoses significantly increased the internal rotation moment (Table 5.6.) though the main effect was obviously seen when combined with the sandals illustrated by the bar chart in Fig 5.17.

The maximum external rotation ankle moment was significantly decreased by the full length orthoses over the combined footwear only condition ($p=0.048$), the $\frac{3}{4}$ orthoses resulted in decreases in both the sandals and the shoes but did not reach the level of significance (see Fig. 5.17) when compared with the combined footwear condition. The transverse moment range was also increased significantly by the full length orthoses, but Fig.5.17. illustrates this was mainly due to the large increase seen in the sandal condition. The $\frac{3}{4}$ orthoses also demonstrated an increase in the sandals but this was not enough to produce a significant result when combined with the shoes which demonstrated small decreases with both orthoses.

Summary:

- The shoes increased the plantarflexion moment and the sagittal moment range over the sandals. All moment reported in the coronal plane were significantly increased by the shoes.
- Both the external rotation and transverse moment range were increased by the shoes over the sandals; however the internal rotation moment was decreased.
- The orthoses did not produce any significant changes in the sagittal plane.
- The inverting moment was only significantly increased by the full length orthoses, but both orthoses reduced the everting moment, the moment range was unaffected.
- The internal rotation moment was significantly increased by both orthoses, however only the full length orthoses significantly increased the transverse moment range. All these changes were seen when the orthoses were combined with the sandals.
- The external rotation ankle moment was decreased significantly by the full length orthoses and was mainly due to the results when combined with the sandals rather than the shoes.

Table 5.6. Mean moments of the ankle highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Ankle Moments Nm/kg	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM 34 orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	34	FL
Plantarflexion	0.26(.09)	0.16(.07)	P=0.000	0.22(.11)	0.17(.08)	0.24(.09)	0.16(.09)	P=0.332	P=0.549
Dorsiflexion	-1.71(.17)	-1.71(.11)	P=0.198	-1.70(.1)	-1.64(.14)	-1.69(.19)	-1.66(.16)	P=0.192	P=0.286
Sagittal Range	1.97(.24)	1.87(.15)	P=0.001	1.92(.22)	1.81(.17)	1.94(.25)	1.82(.20)	P=0.169	P=0.297
Inverting	0.40(.15)	0.32(.17)	P=0.040	0.42(.18)	0.38(.17)	0.44(.18)	0.41(.17)	P=0.183	P=0.034
Everting	-0.15(.04)	-0.09(.05)	P=0.000	-0.14(.04)	-0.06(.05)	-0.14(.04)	-0.05(.04)	P=0.017	P=0.001
Coronal Range	0.55(.18)	0.41(.18)	P=0.000	0.56(.19)	0.44(.19)	0.58(.20)	0.46(.19)	P=0.518	P=0.248
Internal Rot	0.06(.06)	0.10(.08)	P=0.000	0.07(.07)	0.15(.09)	0.07(.06)	0.20(.12)	P=0.028	P=0.000
External Rot	-0.11(.08)	-0.06(.05)	P=0.000	-0.09(.07)	-0.05(.04)	-0.09(.07)	-0.04(.03)	P=0.126	P=0.048
Transverse Range	0.17(.08)	0.15(.06)	P=0.007	0.16(.08)	0.20(.08)	0.16(.07)	0.24 (.12)	P=0.259	P=0.013

5.2. Normative Data for Stance Limb during Step Descent

When collating the results it was noticed that the step trials for subject 8 using the full length orthoses in their own shoes were not as expected. This was the last group of trials for that subject and all five produced outlying and inconsistent ranges. It was thought that this was due to a slipped tibial cluster which was used to reference calcaneal segment and knee motion; as such the step data for the full length shoe step trials was withdrawn. The walking data was scrutinised but seemed unaffected and was therefore included.

5.2.1. Calcaneal segment with respect to tibial segment – step descent

During the step down task set data was only reported for the eccentric limb. In the sagittal plane the calcaneal segment dorsiflexed as the tibial segment moved forwards to lower the swing limb onto the next step. For some subjects this was a steady progression where others showed an increase in the rate of flexion around 50% of the step phase, heel lift occurred around 60-65% of step down, some subjects also demonstrated a plateau in dorsiflexion from around 75% of step down. In the coronal plane most subjects started in an everted position and this increased a little until heel lift when eversion decreased. It was a similar story in the transverse plane the calcaneal segment was abducted on the tibial segment. In some subjects this increased a little as the foot prepared for heel lift and dorsiflexion of the ankle

increased but after heel lift it tended to move towards adduction. The mean range tended to be greater in the transverse plane than the coronal plane (see Fig.5.18).

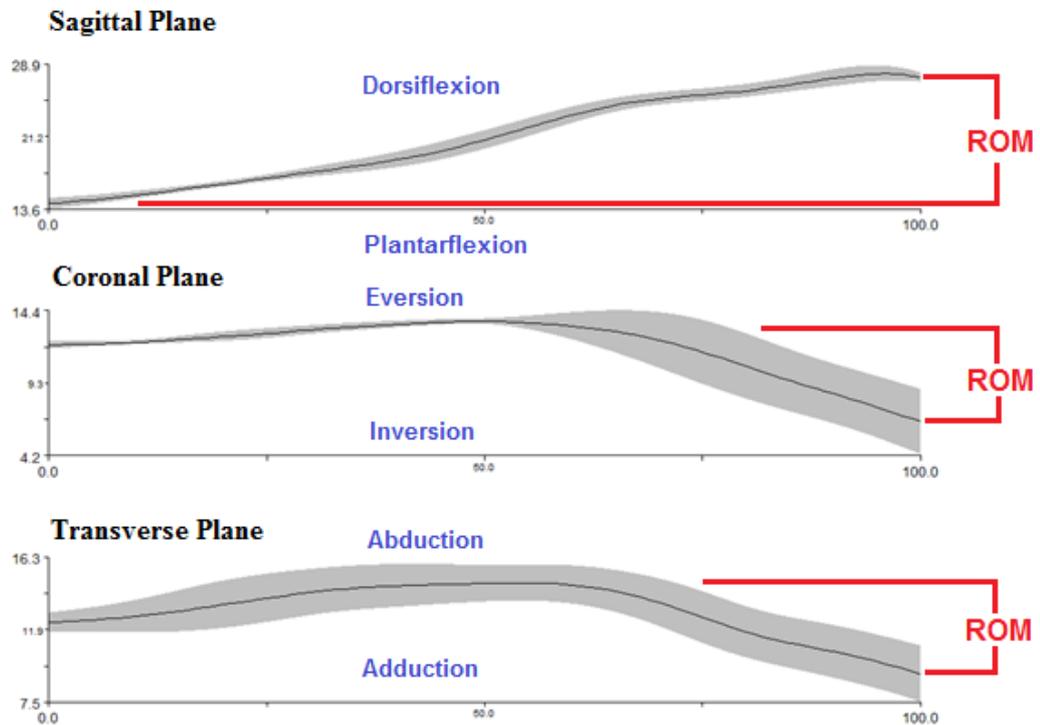


Fig. 5.18. Traces of calcaneal segment motion with respect to the tibial segment, each trace is an ensemble of five trials showing mean and standard deviation.

Similarly to the walking trials, the shoes produced a significant increase in the sagittal plane range of motion over the sandals ($p=0.000$, Table 5.7.) and this was seen as an increase in mean range of almost 5° . In the coronal plane the shoes appear to increase the mean range of motion over the sandal condition, but this was not significant. The shoes in some individuals increased the mean range of motion in the transverse plane over the sandals, but the mean difference was only half a degree and therefore did not reach a level of significance.

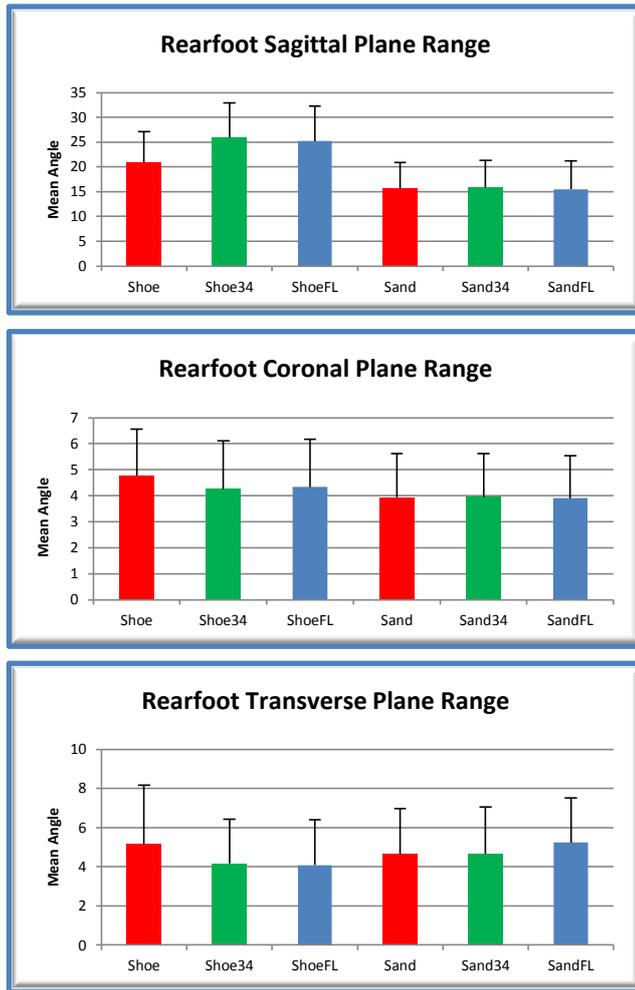


Fig. 5.19. Bar charts illustrating mean ranges of motion between footwear conditions with and without orthoses.

Neither of the orthoses generated a significant effect; however there was another 5° increase seen when they were combined with the shoes which may be clinically relevant. As there was no difference seen with the sandals this lead to a non-significant result (illustrated by Fig. 5.19.) however the increase seen with the shoes is reflected in the shift confidence intervals (Table 6 Appendix 2).

In the coronal plane both the orthoses reduced the total range of motion marginally when combined with the shoes, but there was no difference seen when they were combined with the sandals (see Fig. 5.19) hence their effect was not statistically significant. The results in the transverse plane were comparably to the coronal plane, the orthoses tended to reduce the range when combined with the shoes however there were small increases seen with the sandals leading to non-significant result.

Summary:

- The shoes significantly increased the sagittal plane range of motion over the sandal condition. There were small increases seen in the other plane but these were not significant.
- The orthoses did not produce any significant results over the footwear only condition. There were small increases in the sagittal plane and small reductions in the coronal and transvers planes noted when combined with the shoes but this was not apparent when combined with the sandals.

Table 5.7. Mean range of motion of the calcaneal segment with respect to the tibial segment during step down, highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Calcaneal segment range of motion with respect to the shank during step down	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM 34 orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	34	FL
Dorsiflexion/plantarflexion	20.9(6.1)	15.6(5.4)	P=0.000	25.9(6.1)	15.8(5.5)	25.2(7.1)	15.5(5.8)	P=0.102	P=0.185
Inversion/eversion	4.9(1.7)	3.9(1.7)	P=0.124	4.3(1.8)	3.9(1.7)	4.3(1.8)	3.9(1.6)	P=0.482	P=0.513
Abduction/adduction	5.2(3.0)	4.7(2.3)	P=0.475	4.2(2.4)	4.7(2.4)	4.1(2.3)	5.2(2.3)	P=0.424	P=0.678

5.2.2. Metatarsal segment with respect to calcaneal segment – step descent

When descending the metatarsal segment tended to be plantarflexed on the calcaneal segment throughout the step (Fig. 5.20. less than 90°). Some subjects demonstrated more plantarflexion as the heel lifted, though this tended to reduce just before the swing limb hit the lower step. There was much more variation in the coronal plane, a number of subjects started off in inversion and became more inverted as the heel lifted, others, as in Fig. 5.20, started everted but then moved towards inversion as the heel lifted. In the transverse plane the metatarsal segment remained adducted on the calcaneal segment throughout the step and this increased as the foot plantarflexed after heel lift. It was decided to record the overall range of motion for all planes.

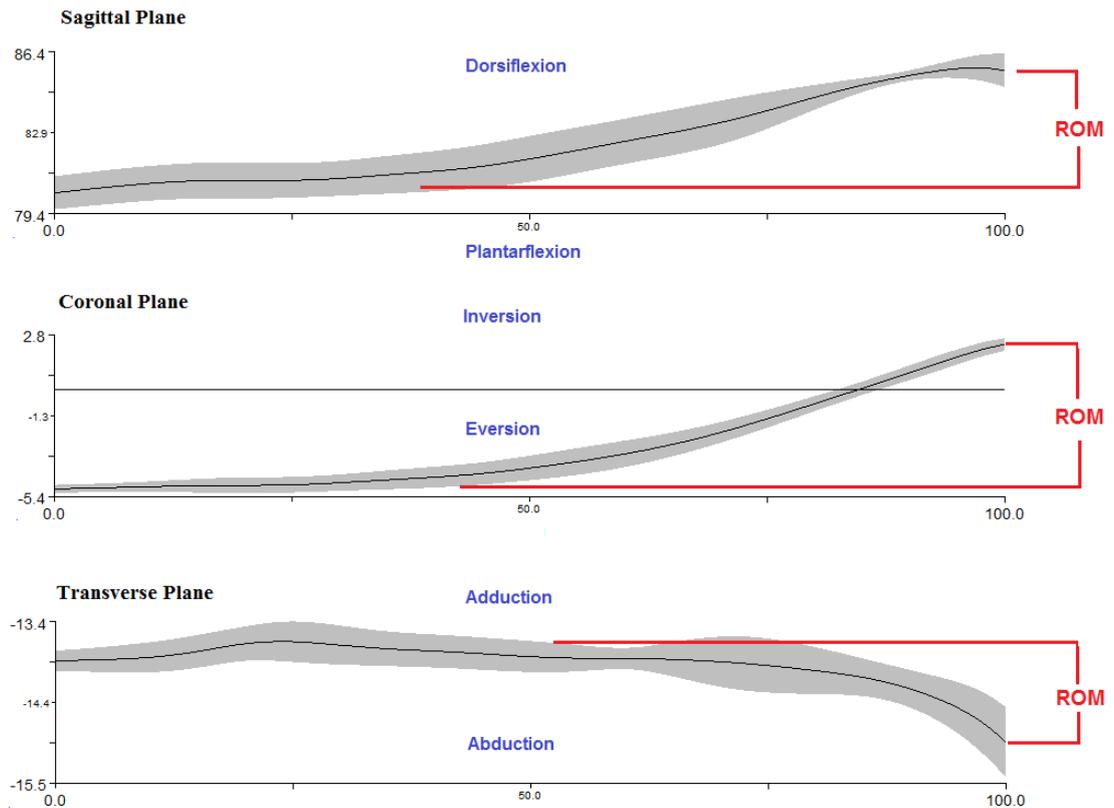


Fig.5.20. Typical traces of metatarsal segment motion with respect to the calcaneal segment, seen in all three planes; each trace being an ensemble of 5 trials. The measurements recorded are highlighted.

The shoes significantly reduced the mean range of motion over the sandals in the sagittal plane by almost 2° which although seems small equates to a 40% reduction ($p=0.000$, Table 5.8.). Similarly in the coronal plane the range was almost half the shoes (1.3°) producing a significant result ($p=0.000$ see Table 5.8.). The shoes also significantly reduced the range of abduction/adduction ($p=0.000$ see Table 5.8.) by half compared to the sandal condition (Fig.5.21).

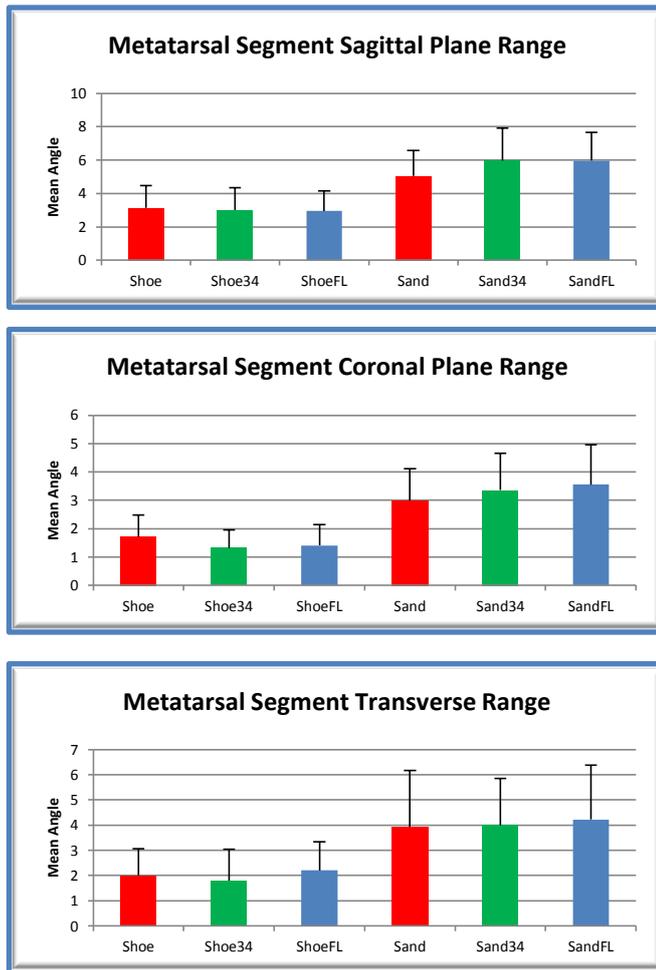


Fig.5.21. Bar charts illustrating mean ranges of motion for the metatarsal segment between footwear conditions with and without orthoses.

In the sagittal plane the orthoses made almost no difference to the mean range in the shoes but they did produce an increase of one degree over the sandal only condition; this was not enough to induce a significant difference.

The orthoses tended to reduce the coronal range of motion by a fraction (0.3°) when combined with the shoes, but increased the range by a similar amount (0.5°) when worn with the sandals. When the footwear conditions were combined to compare with the orthoses neither produced a significant result.

Similarly, in the transverse plane, both orthoses conditions tended to increase the range minimally with the sandals, however this marginal increase was also seen with

the full length orthoses in the shoes. Only the ¾ orthoses demonstrated a small decrease with the shoes, neither orthoses produced a significant result.

Summary:

- The shoes restricted the metatarsal segment range in all three planes and though the measured differences were small, they all demonstrated relatively large percentage reductions.
- The orthoses did not produce any significant differences though it was noted that the orthoses tended to increase metatarsal segment range when combined with the sandals and reduce the range with the shoes over the footwear only conditions. The only exception to this was the full length orthoses when combined with the shoes in the transverse plane where there was a minute increase.

Table 5.8. Mean range of motion of the metatarsal segment with respect to the calcaneal segment during step down, highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Metatarsal segment range of motion with respect to the calcaneal segment during step down	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM ¾ orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	¾	FL
Dorsiflexion/plantarflexion	3.1(1.4)	5.0(1.5)	P=0.000	3.0(1.3)	6.0(1.9)	2.9(1.2)	6.0(1.7)	P=0.286	P=0.339
Inversion/eversion	1.8(0.7)	3.0(1.1)	P=0.000	1.3(0.6)	3.4(1.3)	1.4(0.7)	3.6(1.4)	P=0.880	P=0.703
Abduction/adduction	2.0(1.1)	3.9(2.2)	P=0.000	1.8(1.2)	4.0(1.8)	2.2(1.1)	4.2(2.2)	P=0.848	P=0.592

5.2.3. Phalangeal segment with respect to metatarsal segment – step

Akin to the walking trials only the results for the sagittal plane were recorded for the phalangeal segment. All subjects produced a very similar trace, which only had a small standard deviation about the mean. There were variations of magnitude between subjects; overall range of motion was less than when walking (see Fig.5.22.).

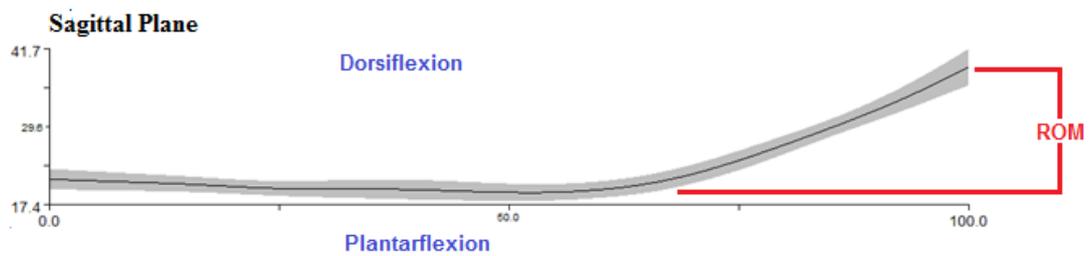


Fig. 5.22. Typical traces of phalangeal segment motion with respect to the metatarsal segment for the sagittal plane; the trace is an ensemble of 5 trials. The measurement recorded is highlighted.

The shoe condition significantly reduced the range of dorsiflexion of the phalangeal segment compared to the range seen in the sandal condition ($p=0.000$ see Table 5.9.) this was a mean reduction of almost 7° .

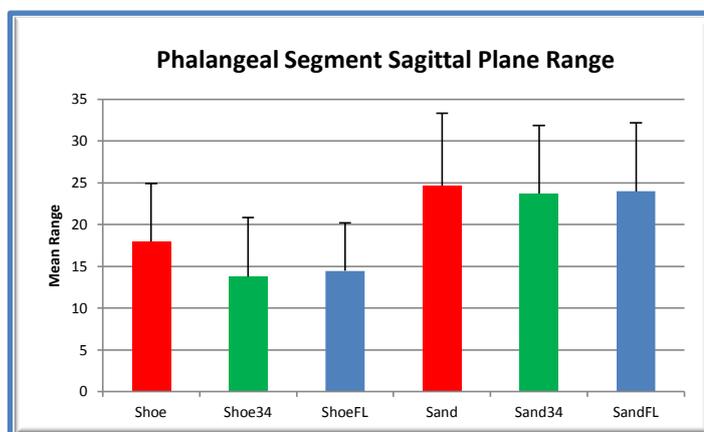


Fig.5.23. Bar chart illustrating mean ranges of motion between footwear conditions with and without orthoses.

Both orthoses restricted the total range of motion in both the shoes and the sandals; however the motion in the latter footwear condition was much less affected than when combined with the shoes. The range of dorsiflexion was not significantly altered by either orthoses.

Summary:

- The shoes significantly reduced the sagittal plane range of motion of the phalangeal segment over the sandals.
- Both orthoses reduced the range further but this was not significant, the reduction was marginally less when combined with the sandals.

Table 5.9. Mean range of motion of the phalangeal segment with respect to the metatarsal segment in the sagittal plane highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Phalangeal segment range of motion with respect to the metatarsal segment during step down	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM 34 orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	34	FL
Dorsiflexion/plantarflexion	17.9(6.1)	24.7(8.6)	P=0.000	13.8(7.1)	23.7(8.2)	14.4(5.7)	24.0(8.2)	P=0.188	P=0.288

5.2.4. Knee kinematics - step descent

Due to the step phase being recorded from just before heel lift of the swing foot the stance knee tended to be marginally flexed at the initial point, knee flexion increased gradually until the swing foot hit the lower step. The traces for most subjects did become somewhat steeper after the heel lifted (see Fig. 5.24.). Because there was a smooth increase in flexion seen in most subjects, it was decided to only record the range of motion in the sagittal plane.

During the walking trials most knees had demonstrated the smallest range of motion in the coronal plane, however when descending steps it varied; some subjects had greater coronal plane motion than transverse plane rotation for others it was the opposite way round. There was also more inter-subject variation seen during the step trials, some started in abduction and then adducted, others moved towards adduction

at the end of the step but remained abducted. Generally the traces demonstrated good repeatability for each subject, the greatest variation from the mean was seen just after heel lift (see Fig.5.24.).

Transverse plane knee rotations were the most variable within and between subjects during the step down trials (see standard deviation in Fig. 5.24.). Generally most subjects initially exhibited a small degree of external rotation and went into internal rotation in the latter 25% of the step, however some subjects did attain an internally rotation position earlier. The “recovery” into external rotation just before the swing foot hit the lower step, was also not seen in all subjects. Due to the variation in the non-symptomatic knees it was felt that only the range of motion would be useful to record.

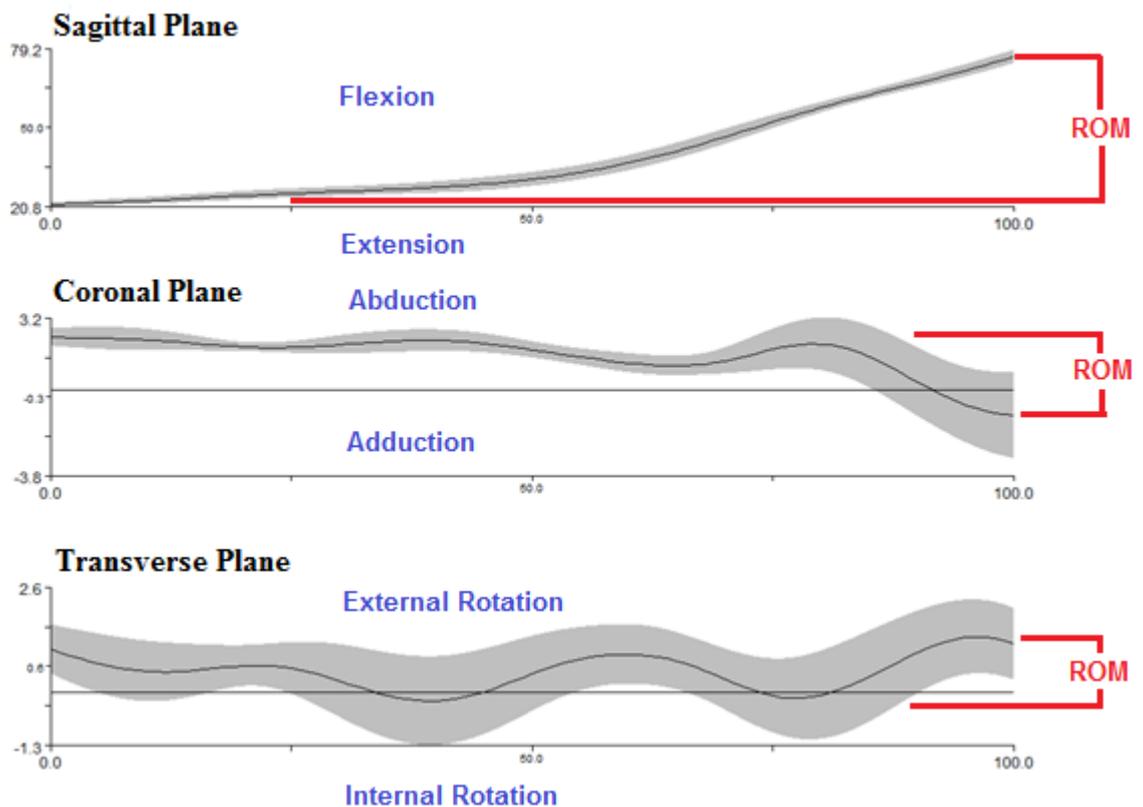


Fig. 5.24. Typical traces of knee motion seen in all three planes; each trace being an ensemble of 5 trials. The measurements recorded are highlighted.

In the sagittal plane the shoes made no difference to the range of flexion (See Table 5.10.) over the sandal condition. It was similar in the coronal and transverse planes both of which demonstrated minor increases with the shoes over the sandals, but this was nowhere near great enough to be a significant result (Fig.5.25.).

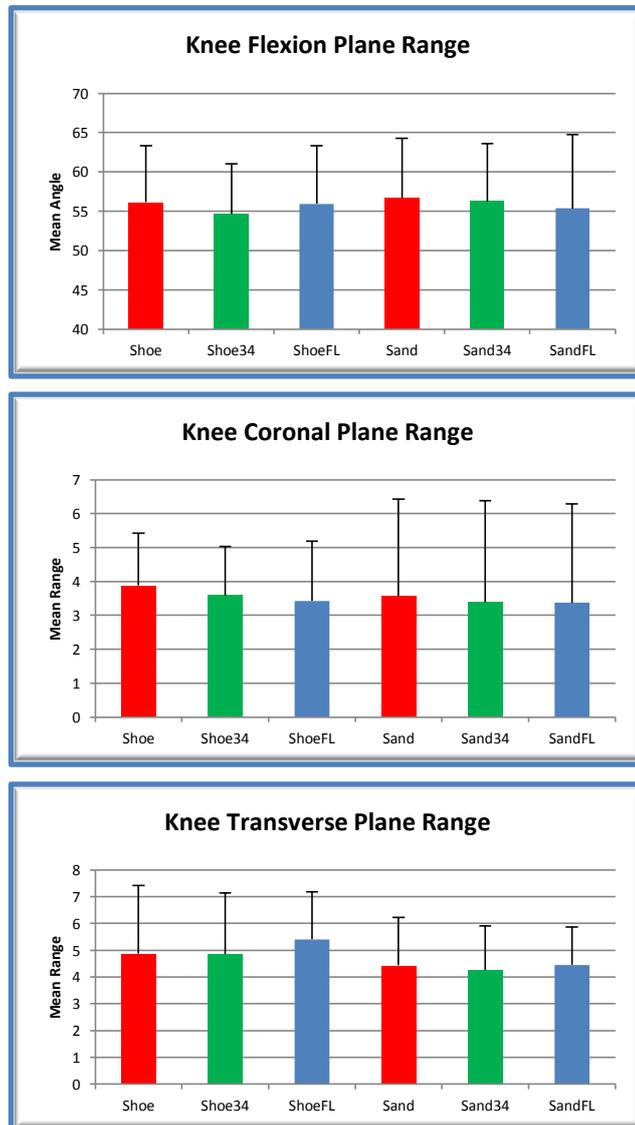


Fig. 5.25. Bar charts illustrating mean ranges of motion between footwear conditions with and without orthoses.

Fig.5.25. illustrates that both orthoses managed to reduce the range of flexion and adduction/abduction, however this was by such small amounts there were no significant differences or trends seen within the data. In the transverse plane the ¾ orthoses tended to reduce the rotational range by a non-significant amount with both footwear conditions. The full length orthoses tended to produce a non-significant increase in the mean range over the footwear only conditions though this was a little

more obvious when combined with the shoes. The bar charts (Fig. 5.25.) illustrate there were no significant changes in the data there are very similar changes seen with the orthoses with both footwear types, the sandal condition seems to increase the variability in the coronal plane (shown with the larger error bars) while the shoe condition has a little more variability in the transverse plane.

Summary:

- Knee motion was unaffected by the type of footwear worn by the subjects during step descent.
- There were no significant changes to report with either orthoses.
- The small changes seen in the mean data were consistent no matter which orthoses were combined in either footwear condition.

Table 5.10. Mean range of motion of the knee highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Knee angle (degrees)	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM 34 orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	34	FL
Flexion Range	56.1(7.3)	56.7(7.6)	P=0.755	54.6(7.0)	56.2(7.4)	55.9(7.4)	55.3(9.4)	P=0.618	P=0.697
Abd/Adduction Range	3.9(1.5)	3.6(2.8)	P=0.699	3.6(1.5)	3.4(3.0)	3.4(1.8)	3.4(2.9)	P=0.686	P=0.584
Int/external Rotation Range	4.9(2.5)	4.4(1.8)	P=0.111	4.8(2.3)	4.2(1.7)	5.4(1.8)	4.5(1.4)	P=0.836	P=0.593

5.2.5. Knee moments – step descent

Three out of the fifteen subjects exhibited a small extension moment with all footwear conditions at the start of each step, though the knee tended to be slightly flexed the ground reaction force (GRF) passed just in front of the knee (see Fig. 5.26. A). The rest started in a flexion moment (Fig 5.26. B and Fig.5.27. sagittal plane graph). As the contralateral limb started to lift off the step, the stance knee flexed increasing the distance of the GRF from the posterior margin of the joint; this tended to remain relatively static through forward continuance phase. The GRF moved forwards towards the phalangeal segment as the knee continued to flex. The moment then increased rapidly as the heel lifted and the knee flexed deeper in the lowering

phase, (See Fig. 5.26.) the flexion moment was the largest moment seen at the knee; mean was 1.36Nm/kg this varied from 0.9Nm/kg to 1.7Nm/kg between subjects.

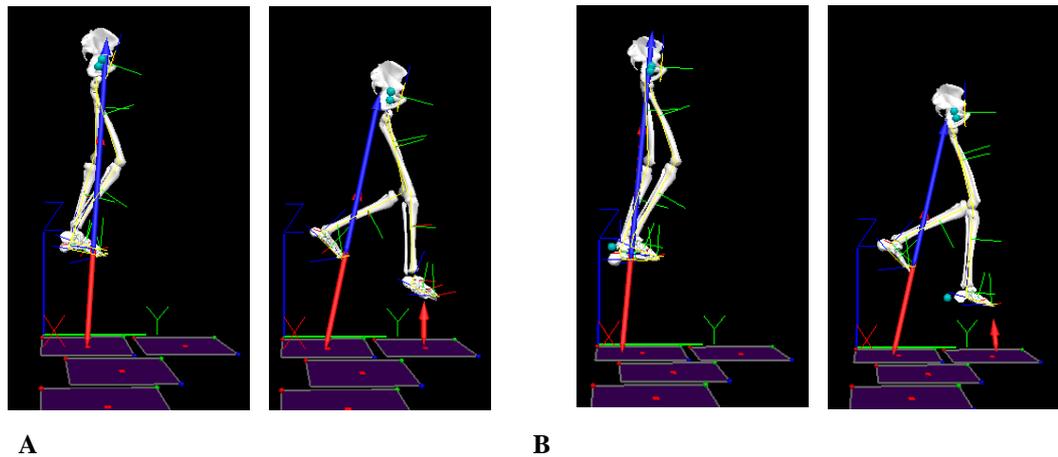


Fig. 5.26. Animations showing extension moment 12 frames into step trial (left A) and largest flexion moment at last frame of trial. Animation B shows same 12 frames into trial but an initial flexion moment.

In the coronal plane the GRF moved towards the stance limb and was angled medial to the knee creating an adduction moment, this decreased and moved towards an abduction moment. Only one subject attained an abduction moment though this was very small (-0.0003Nm/kg). Once the heel lifted the adduction moment increased again, but tended not to reach the same magnitude as was seen at the beginning of the step (see Fig. 5.27.).

Most subjects started with an internal moment at the knee in the transverse plane this quickly reduced as the swing foot progressed in the forward continuum phase. The moment shifted towards or into external rotation (four of the 15 subjects attained an external rotating moment) reaching its minimum position around 25% of the step, the moment then recovered and the internal moment increased through the lowering phase to the end of the step cycle. The transverse moment was the smallest and the most variable between subjects seen at the knee, the maximum mean internal moment was 0.12Nm/kg this ranged from 0.03Nm/kg to 0.26Nm/kg, this is illustrated by the wider standard deviations in the bar charts of Fig. 5.30.

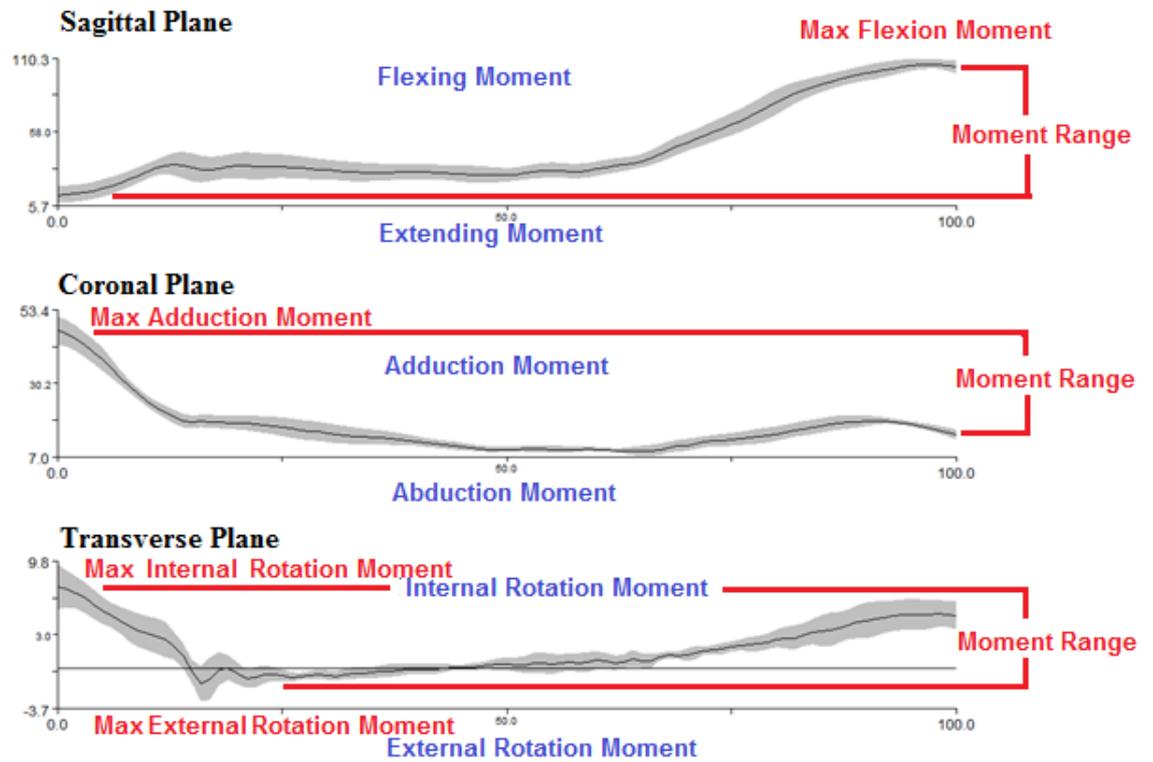


Fig.5.27. Typical traces of knee moments seen in all three planes; each trace being an ensemble of 5 trials. The measurements recorded are highlighted.

The footwear type made no significant difference to the maximum flexion moment during step down, the flexion moment range produced identical results when comparing the shoe and sandal data (Fig.5.28). The minimum adduction moment and the range of coronal plane moment did not demonstrate any significant differences between the footwear conditions (Fig. 5.29.). The rotational moments of the knee were also unaffected by the type of footwear (Fig.5.30.).

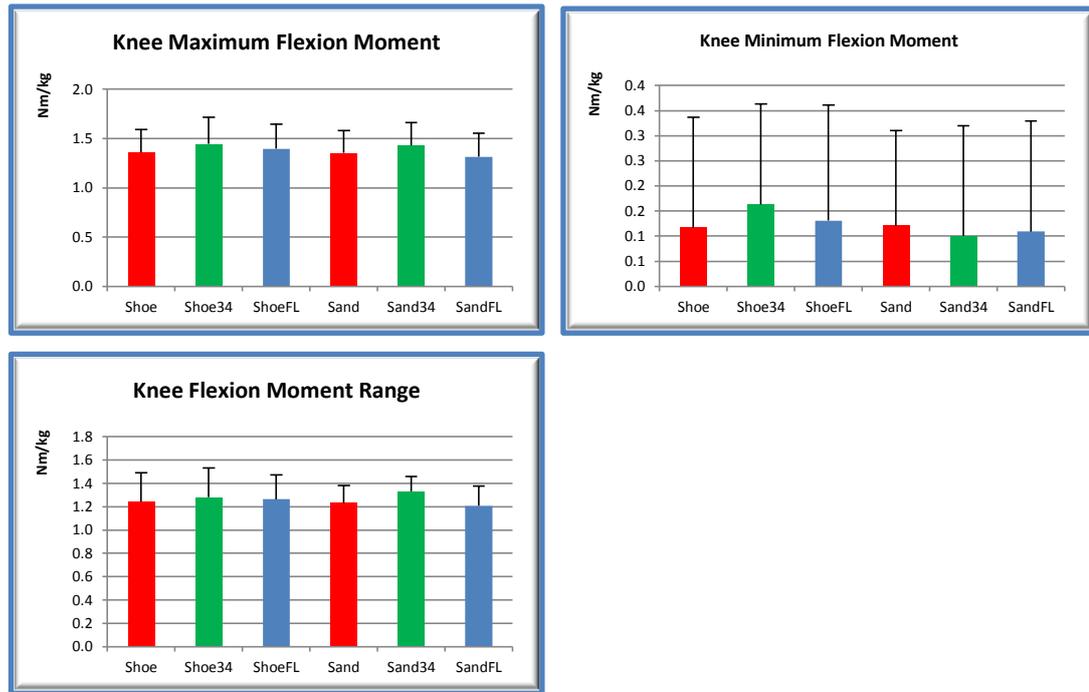


Fig. 5.28. Bar charts illustrating mean knee moments in the sagittal plane between footwear conditions with and without orthoses.

When looking at the mean results of the flexion moment the $\frac{3}{4}$ orthoses did produce an increase in both the sandals and the shoes although not great enough to produce a significant change. The full length orthoses demonstrated a minimal increase in the shoe but a small reduction in the sandal therefore was also not significant. Due to the fact twelve of the subjects did not demonstrate an extension moment the mean became a minimum flexion moment, this variation is illustrated in Fig. 5.28. The mean values being small with a large SD (illustrated by the error bars). The orthoses tended to increase the moment when combined with the shoes, but decreased it when used with the sandals, there were no significant results produced. There was a similar trend seen with the orthoses in the flexion moment range, but again there were no significant changes to report (see Table 5.11.).

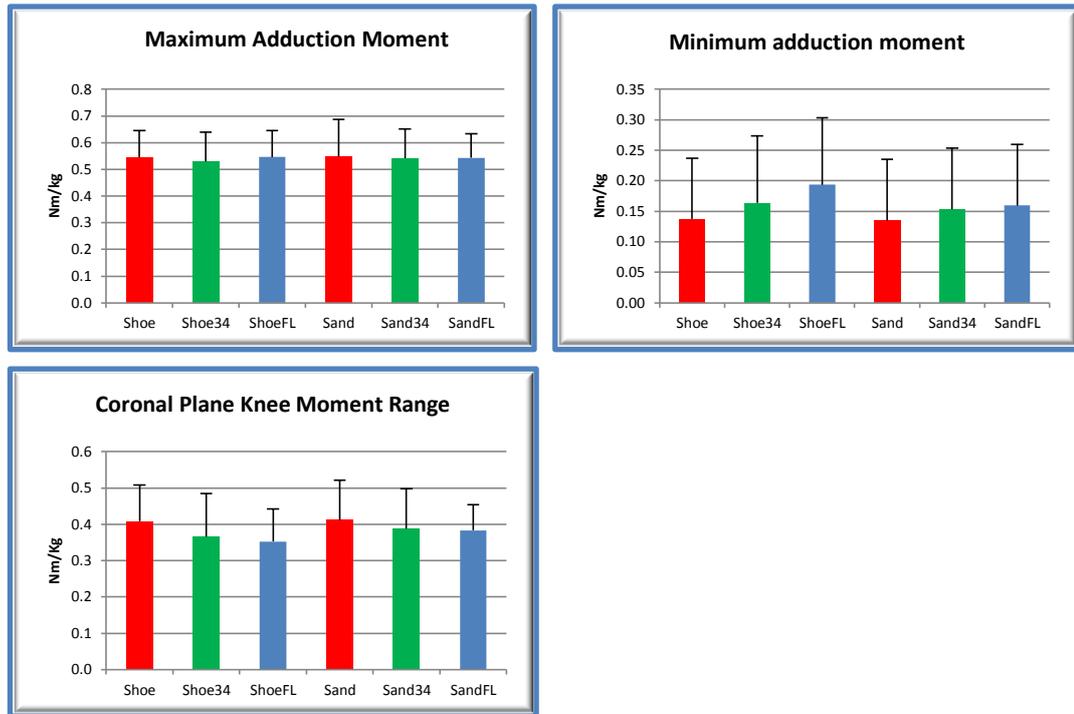


Fig. 5.29. Bar charts illustrating mean knee moments in the coronal plane between footwear conditions with and without orthoses.

The maximum adduction moment was not significantly affected by the orthoses; the error bars are small in Fig. 5.29. demonstrating there was little variation between subjects. The minimum adduction moment was increased marginally by both the orthoses, the full length orthoses produced a marginally bigger difference however this was not significant (Table 5.11.). The range of coronal plane moment was not significantly affected by the orthoses; there were small decreases in both footwear conditions (see Fig. S5.29.). The biggest reduction was seen with shoes with the full length orthoses but this was not significant.

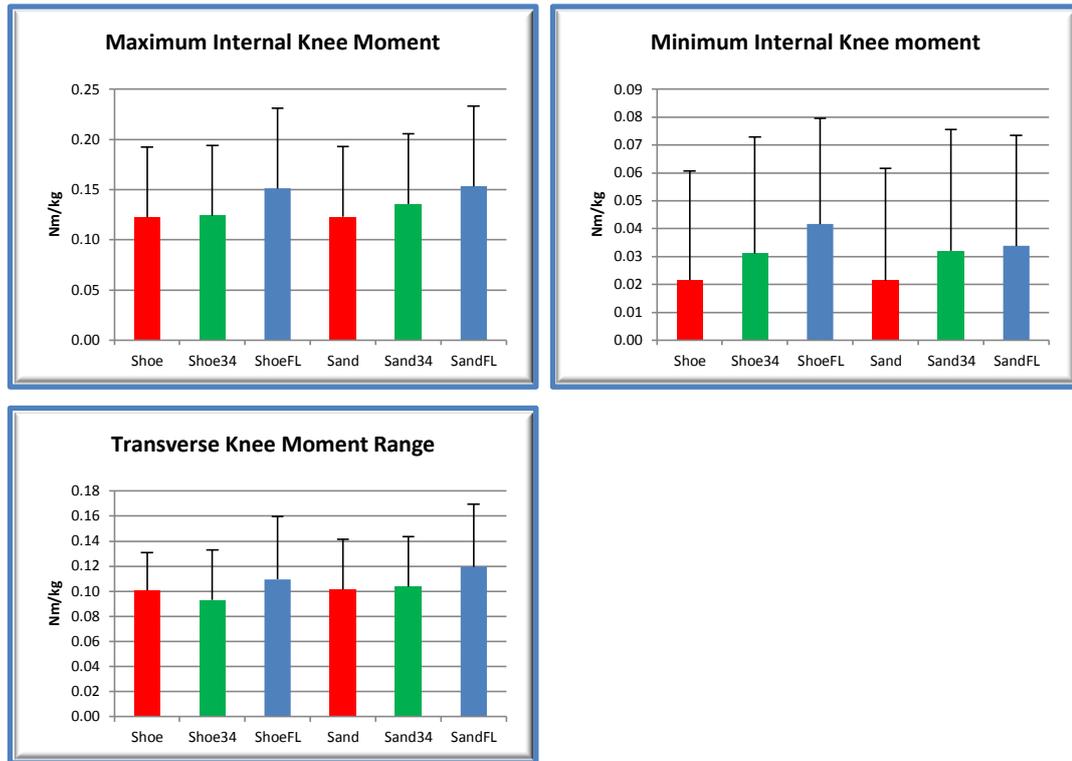


Fig. 5.30. Bar charts illustrating mean knee moments in the transverse plane between footwear conditions with and without orthoses.

When looking at the maximum internal moment of the knee the $\frac{3}{4}$ orthoses induced a minimal increase in both the shoes and the sandals though did not produce a significant result (Table 5.11.). The full length orthoses also increased the maximum moment and again this was not significant, however the increase was reflected in the confidence intervals (Table 7 Appendix 2). The full length orthoses in the shoes produced the greatest increase in the minimum internal moment and although both orthoses in the shoes and the sandals produced increases, the large SD seen between subjects (Fig. 5.30.) did not produce a significant result. The transverse moment was not affected significantly by either orthosis.

Summary:

- There were no significant differences in any of the knee moments when the shoe and sandal data was compared.
- Neither of the orthoses altered any of the knee moments significantly.

Table 5.11. Mean maximum moments and ranges of moment of the knee highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Knee Moments Nm/kg During step down	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM 34 orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	34	FL
Max Flexion	1.36(.23)	1.35(.23)	P=0.513	1.44(.27)	1.43(.23)	1.39(.25)	1.32(.24)	P=0.192	P=0.997
Min Flexion	0.12(.22)	0.12(.19)	P=0.565	0.16(.21)	0.10(.22)	0.13(.23)	0.11(.23)	P=0.816	P=0.978
Flexion Range	1.24(.24)	1.23(.15)	P=0.873	1.28(.24)	1.33(1.3)	1.26(.21)	1.21(.17)	P=0.177	P=0.979
Max Adduction	0.54(.10)	0.55(.14)	P=0.870	0.53(.11)	0.54(.11)	0.55(.10)	0.54(.09)	P=0.693	P=0.958
Min Adduction	0.14(.10)	0.13(.10)	P=0.477	0.16(.11)	0.15(.10)	0.19(.11)	0.16(.10)	P=0.399	P=0.127
Coronal Range	0.41(.10)	0.41(.11)	P=0.367	0.37(.11)	0.39(.11)	0.35(.09)	0.38(.07)	P=0.197	P=0.108
Max Internal	0.12(.07)	0.12(.07)	P=0.761	0.12(.07)	0.13(.07)	0.15(.08)	0.15(.08)	P=0.700	P=0.121
Min Internal	0.02(.04)	0.02(.04)	P=0.785	0.03(.04)	0.03(.04)	0.04(.04)	0.03(.04)	P=0.335	P=0.127
Transverse Range	0.10(.03)	0.10(.04)	P=0.354	0.09(.04)	0.10(.04)	0.11(.05)	0.12(.05)	P=0.583	P=0.380

5.2.6. Ankle moments – step descent

The GRF passed through the centre of the foot at the beginning of each step trial this tended to produce the minimum dorsiflexion moment in many of the subjects, none of the subjects demonstrated a plantarflexion moment throughout the step trials. As the forward continuum phase progressed the GRF was angled further away from the ankle joint centre increasing the dorsiflexion moment, a further increase was seen as the step progressed into the lowering phase and the heel lifted reaching its maximum as the swing limb hit the lower step or just before.

In the coronal plane the GRF passed to the medial side of the ankle joint as the swing limb left the step creating the maximum inverting moment; this reduced quickly as the stance limb took the full weight of the body. Three of the subjects, when wearing their normal shoes, produced a small everting moment (see Fig. 5.31.) the others only demonstrated a reduced inverting moment. As the GRF progressed distally along the foot it also moved medially so the inverting moment increased once more, however it tended not reach the level seen at the start of the step.

The rotating moments of the ankle were by far the smallest and the most variable between subjects. All subjects started with an externally rotating moment when wearing their own shoes this reduced as the swing limb left the step. Some subjects posted an internally rotating moment at this point while others did not (see Fig.5.31.). The moment then tended to remain fairly constant until the heel lifted. As the swing limb impacted the lower step the maximum internally rotating moment was reached; five of the subjects remained with an externally rotating moment throughout the step.

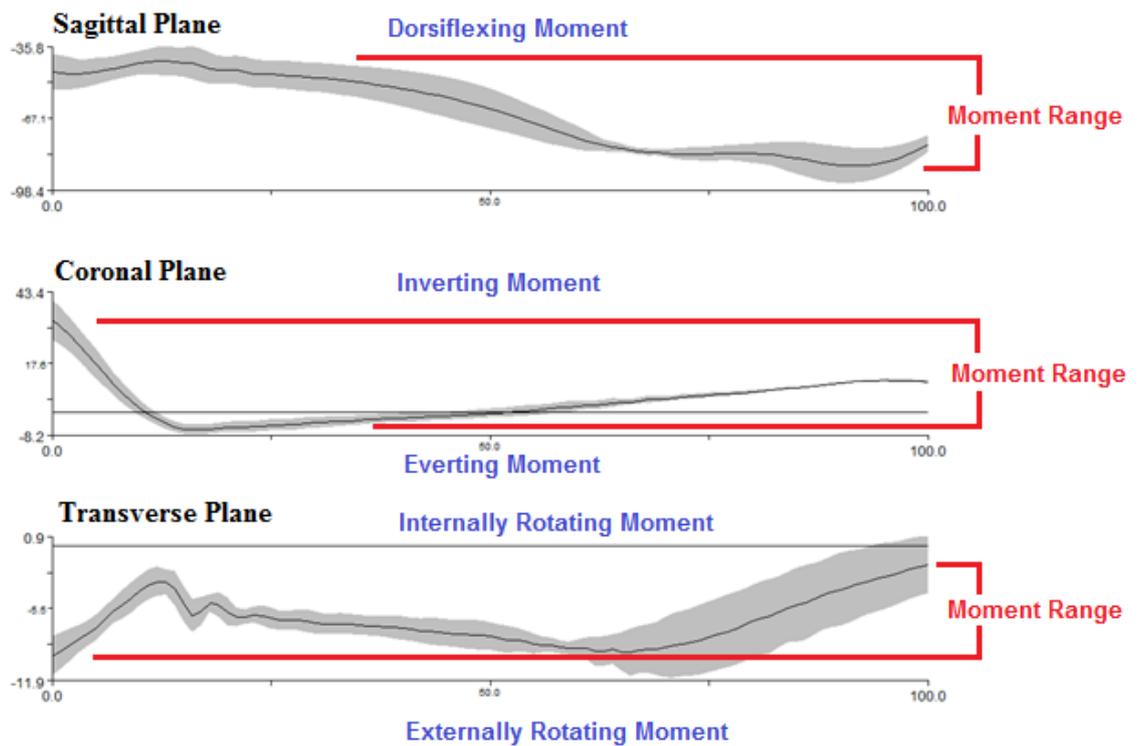


Fig. 5.31. Typical traces of ankle moments seen in all three planes; each trace being an ensemble of 5 trials. The measurements recorded are highlighted.

The maximum and minimum ankle dorsiflexion moments were not significantly affected by the shoes compared to the sandal condition. The range of sagittal plane ankle joint moment was significantly reduced by the shoes compared to the sandals ($P=0.030$) it would appear due to the relatively small mean difference, that this reduction was systematic throughout the cohort. The maximum and minimum inverting moments were not altered by the shoe compared to the sandals, and this

was also the case when looking at the coronal moment range. The rotational ankle joint moments were all affected by the footwear type, the maximum internal moment was decreased by the shoe condition over the sandal, but the converse was true when looking at the maximum external moment (both $p=0.000$) this is illustrated well by the bar charts in Fig. 5.34. The range of transverse plane moment was reduced significantly by the shoes compared to the sandals even though the difference was very small ($P=0.001$ Table 5.12.).

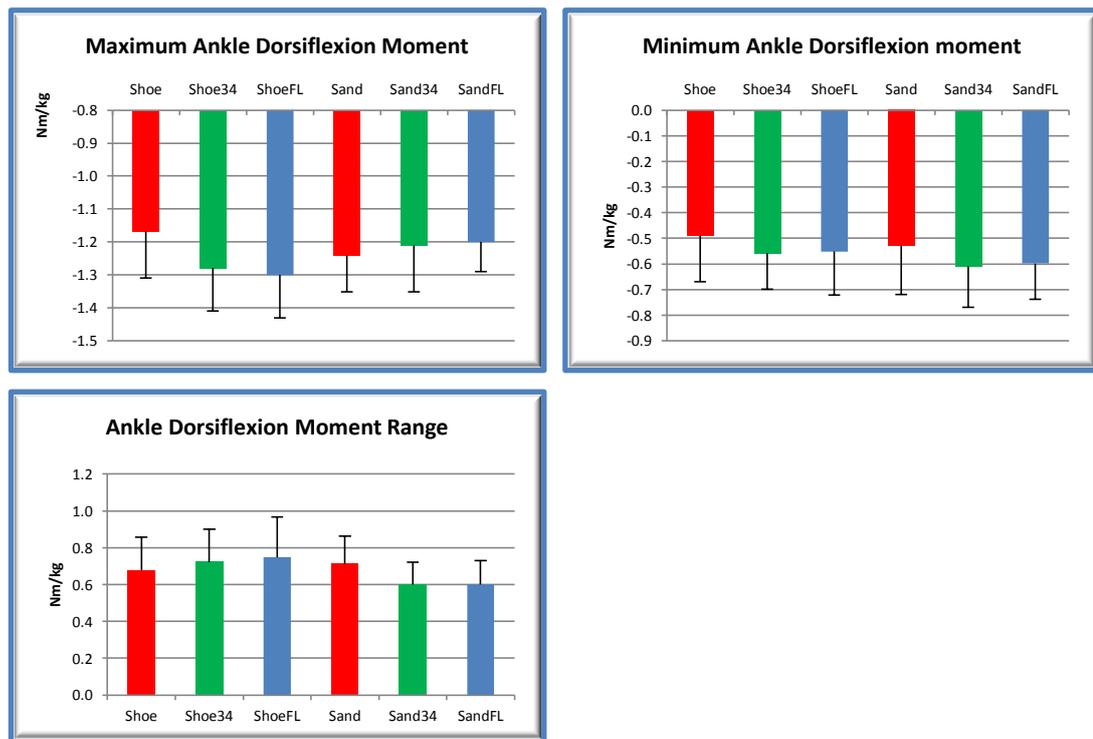


Fig. 5.32. Bar charts illustrating mean ankle moments in the sagittal plane between footwear conditions with and without orthoses.

Neither orthoses significantly altered the maximum dorsiflexion moment. When they were fitted in the shoes both orthoses posted a minimal increase in the mean moment, but in the sandals there were minimal decreases seen. When considering the results for the minimum dorsiflexion moment both orthoses increased the moment over the footwear only conditions. This was not enough to post a significant change, but $\frac{3}{4}$ orthoses did produce a trend in the data ($P=0.092$). Fig.5.32. demonstrates that the orthoses increased the range on moment when combined with the shoes but decreased it when combined with the sandals and this lead to a non-significant result.

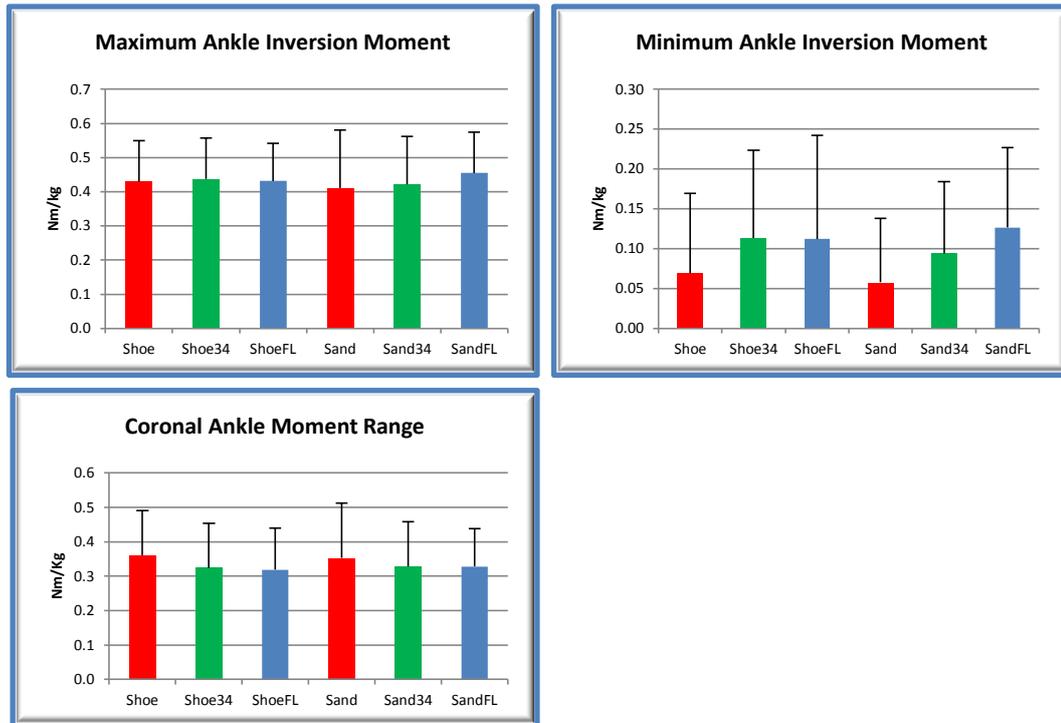


Fig. 5.33. Bar charts illustrating mean ankle moments in the coronal plane between footwear conditions with and without orthoses.

The maximum inverting moment was not altered by the orthoses (Table 5.12.). The minimum inverting moment was increased by both the orthoses, however only the full length orthoses produce a significant increase over the footwear only condition ($P=0.039$). The increases seen in the minimum inversion moments were not enough to change to the range of the moment. Generally the orthoses reduced the range in both the shoes and the sandals but this was not enough to produce a significant result (see Fig.5.33.).

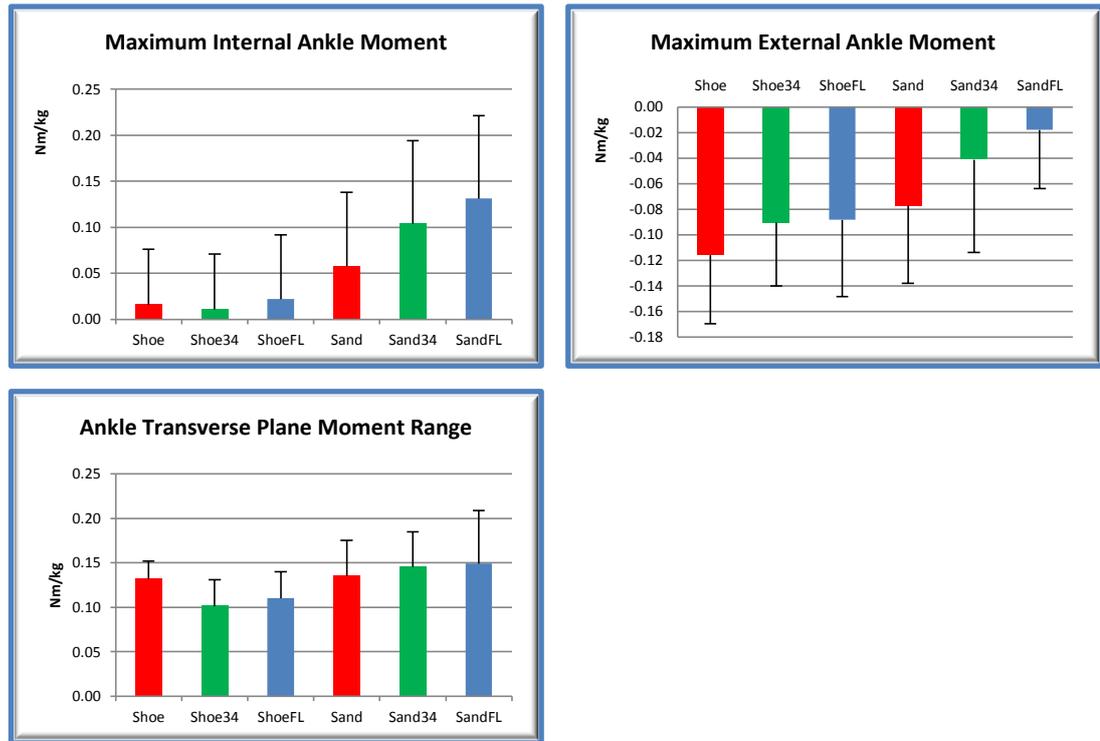


Fig. 5.34. Bar charts illustrating mean ankle moments in the transverse plane between footwear conditions with and without orthoses.

The full length orthoses significantly increased the internal motion over the combined footwear as there were increases with both the shoe and the sandal conditions ($P=0.050$). The $\frac{3}{4}$ orthoses still demonstrated an increase when combined with the sandals but there was a reduction seen with the shoes, therefore their effect was not significant. The External ankle joint moment was reduced significantly by both orthoses over the combined footwear condition ($P=0.043$ and $P=0.005$). The rotational moment range was increased by the orthoses when combined with the sandals but reduced when combined with the shoes; therefore the combined values did not produce a significant result (Table 5.12)

Summary:

- The sagittal plane moment range was significantly reduced by the shoes over the sandal condition. The coronal moments were not affected by the sandals or shoes.
- All the transverse plane ankle moments were altered by the footwear; the internally rotating moment was increased by the shoes over the sandals but

with the externally rotating moment it was the sandals that demonstrated a significant increase. The rotational moment range means were reported equal but overall the sandals returned a significant increase.

- The orthoses did not significantly alter the sagittal plane moments; however the ¾ orthoses did demonstrate an increased trend in the minimum dorsiflexion moment.
- In the coronal plane only the full length orthoses significantly increased the minimum inverting moment over the footwear only conditions. The increase was more evident in the sandals.
- Both orthoses decreased the maximum externally rotating moment significantly, but only the full length orthoses increased the maximum internally rotating moment, this increase was mainly due to the results when combined with the sandals.

Table 5.12. Mean maximum moments and ranges of moments of the ankle highlighting similarities and differences between shoe and sandal conditions and any changes the orthoses induce when compared to the footwear conditions combined.

Ankle Moments Nm/kg During step down	Mean ROM Shoe	Mean ROM Sandals	Significance Shoe Vs Sandal	Mean ROM ¾ orthoses		Mean ROM FL orthoses		Significance Footwear Vs Orthoses	
				Shoe	Sand	Shoe	Sand	¾	FL
Max Dorsiflexion	-1.17(.18)	-1.24(.19)	P=0.255	-1.28(.14)	-1.21(.16)	-1.30(.17)	-1.20(.14)	P=0.217	P=0.169
Min Dorsiflexion	-0.49(.14)	-0.53(.11)	P=0.204	-0.56(.13)	-0.61(.14)	-0.55(.13)	-0.60(.09)	P=0.092	P=0.140
Sagittal Range	0.68(.17)	0.71(.15)	P=0.030	0.72(.18)	0.6(.12)	0.75(.22)	0.6(.13)	P=0.438	P=0.650
Max Inverting	0.43(.12)	0.41(.17)	P=0.902	0.44(.12)	0.42(.14)	0.43(.11)	0.45(.12)	P=0.787	P=0.516
Min Inverting	0.07(.10)	0.06(.08)	P=0.808	0.11(.11)	0.09(.09)	0.11(.13)	0.13(.10)	P=0.131	P=0.039
Coronal Range	0.36(.13)	0.35(.16)	P=0.948	0.32(.12)	0.33(.13)	0.32(.12)	0.33(.11)	P=0.362	P=0.331
Max Internal	0.016(.06)	0.057(.08)	P=0.000	0.011(.06)	0.104(.09)	0.022(.07)	0.131(.10)	P=0.306	P=0.050
Max Externally	-0.12(.05)	-0.08(.06)	P=0.000	-0.09(.05)	-0.04(.07)	-0.09(.06)	-0.02(.05)	P=0.043	P=0.005
Transverse Range	0.13(.02)	0.13(.04)	P=0.001	0.10(.03)	0.14(.04)	0.11(.03)	0.15(.06)	P=0.329	P=0.718

5.3. Summary of Key Findings of Chapter 5

Overall Summary

Calcaneal Segment:

- The shoes significantly increased the range of motion in the sagittal plane both when walking and during the step descent trials.
- The increase in sagittal plane range was also detected with the $\frac{3}{4}$ orthoses leading to the conclusion raising the heel height was responsible for this increased range.
- There was no difference detected between the sandals and the shoes in the coronal plane however there were significant reductions with the orthoses when walking. These were not repeated during the step trials due to the variation between the sandals and the shoes.

Metatarsal segment:

- The shoes tended to have the dominant effect on the metatarsal segment reducing the range significantly in all three planes, during both the walking and step trials.
- The orthoses restricted the coronal plane range of motion during late stance phase of walking.
- It was noted there were variations in the reaction to the orthoses depending on the footwear during the step down trials.

Phalangeal segment:

- The shoes significantly reduced the sagittal plane range of motion during the walking and step down trials.
- The orthoses tended to reduce the range further the full length orthoses producing a significant reduction during walking. The restriction in range was less with the sandals than the shoes.

Knee Kinematics:

- Maximum flexion at heel strike, maximum extension and maximum flexion at toe off were all reduced by the shoes over the sandals. The shoes increased the range of coronal plane motion at the knee, during walking, the type of footwear had no effect during the step down trials.

- The $\frac{3}{4}$ orthoses did significantly reduce maximum flexion at toe off but it was noted that this was mainly due to the results with the sandals. No other changes were seen during walking or step down trials with either orthoses.

Knee Moments:

- The shoes significantly increased the sagittal plane moment range and the external rotation moment at the knee over the sandal only condition during walking. There were no differences noted between the footwear types during step down.
- The adduction moment and coronal moment range were increased significantly by the full length orthoses during the walking trials.
- The orthoses did not produce any significant results in either the sagittal or transverse planes during walking. There were no changes seen in any plane during step down.

Ankle Moments:

- The shoes increased the plantarflexion moment and the sagittal moment range over the sandals. All moment reported in the coronal plane were significantly increased by the shoes during the walking trials. However during step down the sagittal plane moment range was significantly reduced by the shoes over the sandal condition. The coronal moments were not affected by the footwear type.
- Both the external rotation and transverse moment range were increased by the shoes over the sandals; however the internal rotation moment was decreased during the walking trials. During the step down trials the internally rotating moment was increased by the shoes over the sandals but with the externally rotating moment it was the sandals that demonstrated a significant increase.
- The inverting moment was significantly increased by the full length orthoses both during the step down and walking trials
- The maximum externally rotating moment significantly decreased by the full length orthoses during both trials, the $\frac{3}{4}$ orthoses repeated this feat but only during the step trials. During the step down trials the full length orthoses also increased the maximum internally rotating moment, this increase was mainly due to the results when combined with the sandals.

Chapter 6. Initial Discussion

6.1. Discussion Non-symptomatic Subjects

This chapter discusses whether it was possible to compare the kinematics and kinetics of the foot and knee of fifteen non-symptomatic subjects. These comparisons were made when wearing specially developed sandals and everyday shoes, whilst walking and descending stairs. The recordings of the step descent trials will present original normative data. It will consider whether the technique used would be sensitive enough to pick up alterations in the range of motion and knee and ankle moments when using the two different types of foot orthoses.

6.1.1. Calcaneal segment with respect to the tibial segment -walking

6.1.1.i Comparison of sandals and shoes - rearfoot

The use of two different marker sets (one on the foot and one on the shoe) only allowed a comparison of range motion, as maximum peak motion may have been altered by the different marker positions. The results obtained for sagittal plane motion of the rearfoot in this study using the sandals (mean 21.9, CI 19.73 to 22.26) three of the ten previous barefoot studies in Table 3. 1. (Chapter 3 page 34) fell within the confidence intervals seen in these results and was within half a degree of the range reported by Cobb, et al. (2009) in their sandal study. The shoes showed an increase in the mean range of motion over the sandal only condition. This was probably due to the heel raise of most of the shoes increasing the height of the segment and hence increasing the diameter of rotation. Hong, et al. (2001) conducted a study to investigate the effect of females wearing high heeled shoes, using a single segment foot model; they found plantarflexion was increased significantly during the loading stage of stance phase with 5.1cm and 7.6cm heels, maximum dorsiflexion angle was reduced during pre-swing.

The inversion/eversion range of the calcaneal segment, with respect to the shank, was split into initial and propulsive phases of stance; making it difficult to compare the sandal data to previous studies, Cobb, et al. (2009) reported an 8° total range which was just between the two phases reported here. Neither period of stance was

significantly altered by the shoes compared to the sandals. When comparing barefoot and shod gait Wolf, et al. (2008) also found little difference in sub-talar joint rotation, however this cannot be used as a direct comparison as again this was for the entire stance phase.

The mean transverse range of 11.9° (CI 10.77 to 12.99) in the sandals did match well to Cobb, et al. (2009) who reported 10° in sandals, it also compared well to three of the previous studies in Table 1 (Chapter 3, Page 34). However some discrepancies were observed which could have accounted for by the methods used, it should also be noted that neither of these two studies reported on the transverse plane. Wolf, et al. (2008) reported foot rotation which did not isolate the rearfoot, this obviously will have included any forefoot abduction/adduction, this was almost double the results of this study (20.9°). When the shoes were fitted to their adolescent cohort there was a minimal decrease seen.

In this study the shoes significantly increased the range of motion of the calcaneal segment over the sandal condition which could be due to a number of mechanisms: the sole of the shoes are not likely to flex easily across their width, therefore the markers may not pick up the rotation of the calcaneus within the heel counter. Or possibly the rotation of the calcaneus is reduced due to the support of the heel counter of the shoe; however the tibia still rotates giving an overall increase in the range of motion.

6.1.1.ii Effects of orthoses - rearfoot

Although the heel heights in this study were nothing approaching the height used in Hong, et al. (2001) study there was an increase in mean range of motion with the $\frac{3}{4}$ orthoses, as only the proximal half of the foot will have been raised. With the full length orthoses the wedge will have also raised the distal half of the foot more equally and logically there was no significant increase seen. Nester, et al. (2003) reported a significant mean decrease of 2.6° in total range of motion at the rearfoot complex with a similar medial wedged condition, though they used a modified single segment model derived from the ankle joint centre and a marker on the fifth metatarsal head. Cobb, et al. (2011) also reported a significant increase in rearfoot

displacement of 1.1° in their orthoses/sandal study though this was measured as greater dorsiflexion at midstance. It must also be remembered, in this study, that the front strap of the sandals was situated a long way back on the foot to allow access to the metatarsal heads for uninterrupted marker placement. Due to this and the elastic straps some subjects reported feeling a little less stable when walking; this may have led to a reduction in step length which in turn could have reduced the range of motion seen at the ankle with the sandal conditions over the shoes.

Initially it may seem odd that the full length orthoses produced a significant reduction in coronal plane motion of the calcaneal segment during early stance rather than the $\frac{3}{4}$ orthoses, however the forefoot will have loaded within the first half of stance phase and this could be the reason that the full length wedged orthoses induced a significant result. During the second half of stance both orthoses reduced the range of motion by more than a degree; both were significant. Again it seems odd that the $\frac{3}{4}$ medially wedged orthoses had less of an effect on the first half of stance than the second, but this may indicate that it is the arch support rather than the wedge which is producing this effect in the second half of stance.

Eng and Pierrynowski, (1994) reported small (1.8°) but significant decreases in coronal plane range during contact and midstance using soft foot orthoses in their cohort. Ferber and Benson, (2011) did not report any differences when using orthoses in their modified training shoes but they only reported on peak eversion rather than range, similarly Cobb, et al. (2011) found no significant alterations in the coronal plane though they did mention that inversion displacement increased with the orthoses while eversion displacement decreased. From the results of this study and considering the “usual” clinical advice given for the necessity of wearing supportive footwear for orthoses to work most effectively it was surprising that the heel counters of the shoes, compared to the sandals, were not seen to limit the coronal range of motion significantly during either half of stance phase, however it did lead to the assumption that it was acceptable to place the markers on a shoe as similar reductions were seen with both footwear conditions.

Neither orthoses produced a significant result on transverse range rotation, when combined with the shoes the mean range remained almost unchanged. However they

increased the mean range in the sandals, which as the markers were attached to the skin, would suggest this, is as a result of supporting the calcaneal segment. This result does mean that the conclusions drawn from calcaneal segment rotation must be done so with caution. Ferber and Benson, (2011) reported peak tibial internal rotation increased but not significantly with their moulded and non-moulded orthoses, however it was not reported how this was referenced; whether to the laboratory or the rearfoot as the thigh was not modelled, it is therefore impractical to draw comparisons with the present study.

6.1.2. Metatarsal segment with respect to the calcaneal segment -walking

6.1.2.i Comparison of sandals and shoes - metatarsal

It was the shoes that had the main effect in all planes; reducing the range considerably in all planes. Only during the late stance phase in the coronal plane did the orthoses produce a statistically significant reduction over the footwear only condition. Comparing the results of this study to previous studies is difficult as other models have tried to separate the medial and lateral sides. Others have a “tarsal segment” as well as a metatarsal segment; however the sagittal mean range of motion recorded in the sandal condition was similar to Kidder, et al. (1996) and Woodburn, et al. (2006) who were both reporting barefoot motion. When looking at the shoe data there was a significant reduction in sagittal plane excursion over the sandal condition which may be expected due to the extra stiffness of the shoes as there was no shank incorporated in the sandals.

It was also difficult to compare the coronal plane results of this study to previous experiments due to differences in both modelling and the fact that the range here was split to highlight both contact range and the range during propulsion. The shoes did reduce the range significantly over the sandals during both phases of stance; the mean difference was only 0.6° (CI sandals 1.66 to 2.16, shoes 1.01 to 1.51) during loading but 3.7° (CI sandals 5.48 to 6.40, shoes 1.78 to 2.70) during propulsion, which may throw doubt on the statement by Bishop, et al. (2013) who questioned whether a foot-shoe complex model could detect small differences of less than 5° due to slippage and measurement noise. In this study the reduction detected between

shoe and sandal conditions could be either due to support of the shoes or slippage of the foot within the vamp.

A mean transverse range of 5.0° CI 4.40 to 5.68 (in the sandal condition) was almost the same result as reported by Carson, et al. (2001) and similar to Woodburn, et al. (2004) who were both looking at barefoot motion. However, this was lower than the other reported data in Table 3.1, as noted earlier; the differences in segment modelling must be considered as must any influence of the sandals. Morio, et al. (2009) compared barefoot walking to walking in 2 different sandals with soft and stiff soles, they found that there was coronal and transverse plane restriction of excursion but sagittal plane motion was unaffected. The shoes did significantly reduce the range over the sandals but the mean difference was only 1.0° (CI sandals 4.40 to 5.68, shoes 3.41 to 4.69). The reduction was expected primarily due to the extra rigidity of the shoes.

6.1.2.ii Effects of orthoses - metatarsal

Neither orthoses produced a significant result in the sagittal plane, though there tended to be a very small decrease in the shoes (max 0.6°) where the orthoses will add to the stiffness of the mid-section of the footwear and an increase in the sandals (max 0.5°). This may have been due to the elastic straps of the sandals allowing the midfoot to pivot around the arch support. The shoe data could be mapped to the medial longitudinal arch results of Ferber and Benson, (2011) but they were more concerned with looking at elongation of the foot; a significant reduction was reported with their moulded orthoses. There is no direct comparison possible due to the differences in modelling and the fact that their shoe condition had large windows cut in them, this will have held the foot more tightly on to the orthoses than the sandal condition in this study, but may have reduced the control of the shoe as stated by Schultz and Jenkyn, (2011) who suggested windows over 10mm diameter could affect the support of footwear.

In the coronal plane the orthoses did not produce a significant result during the first half of stance, when the midfoot will have only just have loaded, hence the orthoses would be expected to have less effect. However they both produce significant reductions in the propulsive phase, this would have been expected from a clinical

viewpoint. The full length orthoses also demonstrated a greater reduction over the $\frac{3}{4}$ orthoses; no other data was available to directly compare these results. Chevalier and Chockalingham, (2012) did report maximum and minimum motion of the midfoot on their single subject with eleven different orthoses. The ranges calculated from their published data demonstrated both increases and decreases, the differences between the effects of the orthoses was levelled at the prescriptions and ultimately the practitioners. Clinically there could be arguments for and against reducing metatarsal segment coronal range depending on foot type and symptoms, but the significant result suggests this technique is sensitive enough to pick up small alterations in segment range even when the markers are applied to the vamp of a shoe.

In the transverse plane both orthoses reduced the mean range in the shoes by 0.9° while in the sandal condition a small increase was seen. This may suggest that for orthoses to have an effect on the midfoot they need to be supported by a shoe, or attaching the markers to the vamp of the shoe may be masking the motion that is occurring. The only other study to report on abduction/adduction of the forefoot on the rearfoot was Chevalier and Chockalingham, (2012) they used a modified plimsoll to retain their eleven different orthoses on a single subject. By calculating the range from the peak minimum and maximum results published they found that that all the orthoses reduced the range on the left side and all but two of the eleven on the right did the same. This could be suggesting that the foot has to be held onto the orthoses for it to produce the expected reduction in the range of motion.

6.1.3. Phalangeal segment with respect to the metatarsal segment - walking

6.1.3.i Comparison of sandals and shoes - phalangeal

The mean range seen in the sandals only condition was comparable to the data presented in Table1, it must be remembered that some of the experiments tended to model the hallux separately, but as the primary concern of this section was looking at the effect of comparing “foot” function in sandals and shoes; a hallux segment would not have been advantageous.

The shoes did produce a significant reduction over the sandal condition and this could have been due to a combination of factors, obviously there is great potential for the toes to move within the toe box a factor highlighted by Bishop, et al. (2013) who stated that the inner surface of the shoe at this level does not articulate with the dorsal surface of the foot. The reduction of 6.1° (CI sandals 33.29 to 36.97, shoes 27.13 to 30.81) found here was a little less to the reduction found by Wolf, et al. (2008) with their more flexible shoe (9.3°) Schultz, et al. (2011a) noted their shoe condition tended to dorsiflex the hallux compared to the barefoot trials their subjects underwent, this lead to a difference of at least 5° . Not only must the stiffness of the shoes sole should be considered but also any toe spring (see Fig.6.1.) which may or may not have been incorporated in the sole design of the everyday shoes worn by the participants.



Fig. 6.1. Photograph of shoe showing toe spring, this was suspected to act as a rocker and was variable between subjects shoes used in the trials

6.1.3.ii Effects of orthoses - phalangeal

When both orthoses were combined with the sandals there were minimal increases seen which were not significant, Scherer, et al. (2006) did find a significant increase when using orthoses but their orthoses were constructed with the first ray in a plantarflexed position in an attempt to increase dorsiflexion; this was not considered in this experiment. In an attempt to keep the prescriptions constant there was either a full length or a $\frac{3}{4}$ length 5° wedge placed under the moulded slimflex™, certainly

the latter prescription could have dorsiflexed the heads of the metatarsals interfering with the normal dorsiflexion of the phalanges. Medial wedging under the forefoot is often avoided clinically as it is thought to interfere with the windlass mechanism, this was first described by Hicks, (1954) who explained how the plantar fascia is “wound” round the metatarsal heads and used to brace the medial arch of the foot as the toes extend. Kappel-Bargas, et al. (1998) found some individuals had a delayed “arch rise” as the first metatarsophalangeal joint was dorsiflexed, they suggested the function of the windlass mechanism could be inconsistent between subjects.

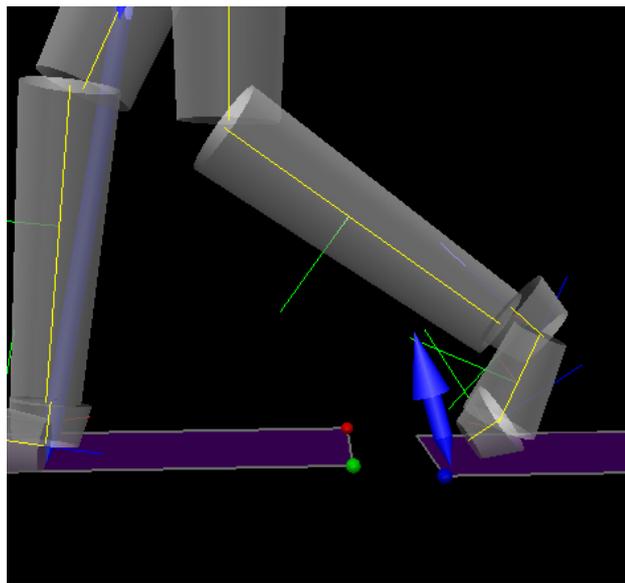


Fig. 6.2. Diagrammatic representation of maximum dorsiflexion of the phalangeal segment just before toe off from Visual 3D (segments being shown as cylinders).

The full length orthoses in the shoes compared to the shoes alone did significantly reduce the motion of the phalangeal segment on the metatarsal segment by 4.2° . This decrease in range of motion could certainly be as a result of interference with the windlass mechanism or it could simply be attributed to an increase in the stiffness of the materials under the toe segment as the wedge and the orthoses extended to the front of the shoes. Further research would be necessary to determine the exact cause of this reduction and whether or not it was clinically significant.

6.1.4. Knee angle data- walking

6.1.4.i Comparison of sandals and shoes - knee

Because the markers used to investigate the knee were not moved between the footwear conditions it was possible to report on maximum positions unlike the foot data. Knee flexion at heel strike was significantly reduced with the shoes over the sandal condition; this was attributed to the elastic straps of the sandals giving a perception of less stability. There were no differences seen between the sandal and shoe conditions at the impact peak, however maximum knee extension or the trough between the impact and propulsive peaks was significantly reduced in the sandals compared to the shoe condition. Maximum flexion at toe off was reduced significantly by the shoes versus the sandals; this again was possibly related to greater stability of the shoes, but could equally be related to the increased range of motion seen at the ankle in the sagittal plane.

Maximum abduction was increased by half a degree by wearing shoes over the sandal condition, this was not significant. When looking at the range of coronal plane motion the shoes did produce a significant increase over the sandal condition, however the mean difference was only 0.9° (CI sandals 3.89 to 5.81, shoes 4.70 to 6.63). It would be unlikely that increases of this magnitude would be documented in the clinical situation; these increases could not be coupled with any alterations in the calcaneal segment though the shoes did reduce the range of motion of the metatarsal segment in the coronal plane.

The rotation of the tibia on the femur was not affected significantly by the shoes over the sandal condition. Table 6.1. shows typical values obtained from the raw data for knee rotation. Subject 7 was one of three subjects who demonstrated an increase in rotation on both sides with the shoes 5.7° was the largest increase seen in any of the subjects, twenty out of the thirty knees demonstrated small reductions in rotation with the shoes over the sandal condition. The values also display how much variation there was between subjects, subject 3 had the smallest range while subject 6 had by far the largest.

Table 6.1. Raw data values for mean transverse plane knee rotation range highlighting variation between subjects how they reacted to the orthoses

Subject	Sand	Sand 3/4	Sand FL	Shoe	Shoe 3/4	Shoe FL
3	7.96	7.35	7.69	6.77	5.42	5.26
6	27.95	26.10	26.74	25.60	21.75	20.66
7	15.56	16.25	16.47	21.28	20.22	19.73
13	17.37	16.15	17.65	13.18	12.97	14.62

6.1.4.ii Effects of orthoses - knee

The theory that it was the “feeling” of stability that affected the amount of flexion at heel strike was corroborated by the fact the orthoses made no difference in the shoes but there was a small decrease seen in the sandals, many of the subjects reported that the orthoses did give them a “more stable” feeling when worn with the sandals, though none of them could really explain why, theoretically it may have been the arch support of the orthoses which made them feel less likely to slip forwards at contact or again it may purely have been from increasing the sandals “stiffness”. Neither orthoses made any difference to the mean flexion angle during the impact peak, or induced any difference at maximum extension when combined with the footwear.

Maximum flexion at toe off was significantly reduced by the ¾ orthoses over the footwear only condition; this was mainly due to the reduction seen in the sandal condition. The full length orthoses also demonstrated a trend to reduce the maximum flexion angle, but was not significant. This inversely mirrored the results of the calcaneal segment in the sagittal plane, only the ¾ orthoses increased the range significantly. This was in contrast to Nester, et al. (2003) who did not report any significant alterations in the sagittal plane motion of the knee with their medial wedges, however Chen, et al. (2010) did report a small reduction in peak knee flexion angle with insoles in shoes over shoes alone with a group of eleven pronated subjects; although this was not a significant result. The range of sagittal plane motion was calculated from maximum extension to maximum flexion at toe off; no significant differences were seen between the footwear or orthoses conditions, although both orthoses did tend to reduce the range in the shoes and the sandals, but not significantly.

The minimal reductions seen with the shoes over the sandals would seem of little consequence as neither parameter of maximum abduction or the range was altered at the knee with either orthoses; so the further reductions seen in the metatarsal segment were not mirrored at the knee. Lafortune, et al. (1994) used bone pins on five subjects to determine how valgus and varus wedged shoes affected the knee, they reported a 0.2° (0.4) increase with 10° varus shoes, however they reported that total range of abduction/adduction was less than one degree in their subject group. In the same year Eng and Pierrynowski, (1994) had reported significant mean reductions in coronal plane motion of the knee during contact and midstance phases of walking (0.8° and 0.6° respectively) using soft orthoses on a cohort of ten female subjects.

In the transverse plane, it was interesting that subject 7 only demonstrated a reduction of 1.21° with the full length orthoses in the sandals, but this was almost 5.0° when combined in the shoes; it is likely this would be observed and reported clinically. This may intimate in some foot types that it is extremely important to fit the orthoses into supportive footwear. Lafortune, et al. (1994) proposed that internal rotation of the knee was split into three sections an initial internal rotation of 4.5° degrees then a stationary phase followed by another internal rotation of 4.0° . It was reported that their 10° varus wedge only made 1.2° difference but all five subjects did demonstrate a reduction.

6.1.5. Knee moment data - walking

6.1.5.i Comparison of sandals and shoes – knee moments

When looking at both the maximum flexion and extension moments in isolation the shoes did not induce any significant effects, although the shoes did produce a non-significant increase in both over the sandal condition. When the mean moment range was calculated the shoes did induce a significant increase over the sandals.

In the coronal plane there was a definite trend towards the shoes increasing the maximum adduction moment over the sandal condition ($p=0.078$, CI sandals 0.30 to 0.46, shoes 0.43 to 0.50) this may be as a result of making the foot more stable. The

abduction moment was unchanged by the footwear type and this led to no change in the range of coronal plane moment being detected. Chen, et al. (2010) did report a significant increase in the peak varus knee moment between walking barefoot and shod; in their study the mean difference was 0.02Nm/kg ($p=0.031$).

It was only the smaller external rotation moment that was increased significantly by the shoes ($p=0.015$, CI sandals -0.054 to -0.036, shoes -0.064 to -0.046) the larger mean internal rotation moment was completely unaffected by the type of footwear. The moment range did reflect the increase in the external rotation but was not enough to induce a significant increase.

6.1.5.ii Effects of orthoses - knee moments

There were no significant changes seen with either orthoses for any of the sagittal plane parameters measured, this concurs with Nester, et al. (2003) who also reported no sagittal plane moment changes with either their lateral or medially posted orthoses. Schmaltz, et al. (2006) also reported that medially wedging the foot had no significant effect on the external knee flexion moment; however they did note that there was a tendency for the extension moment to be decreased with both medial and lateral wedges which were purported to be as a result of reducing coronal plane motion of the ankle and foot.

Chen, et al. (2010) reported a non-significant reduction in the peak varus moment when they combined their orthoses with the shoes over the shoe only condition. In the present study when the orthoses were combined with the sandals they tended to increase the mean adduction moment by 0.05Nm/kg and 0.06Nm/kg ($\frac{3}{4}$ and full length respectively) whereas in the shoes the effect was tempered (0.01 and 0.03Nm/kg respectively). When the footwear data was combined this was great enough for the full length orthoses to produce a significant increase. This was expected as it is well reported that medial wedging tends to angle the GRF more medially (Nester, et al. 2003). The moment range did tend to follow the adduction moment with the introduction of the orthoses, though the $\frac{3}{4}$ wedges were just outside the level of significance ($p=0.065$, CI 0.43 to 0.50, 0.44 to 0.51) but the full length ones did produce a significant increase ($p=0.007$, CI 0.45 to 0.52, 0.46 to 0.53).

Nester, et al. (2003) similarly did not find any significant changes in the coronal plane kinematics of the knee with a 10° wedge, but they did report a significant increase in the peak adduction moment. It was proposed that as their orthoses were only introduced for the experiment over time the change in moment may be eventually reflected in the kinematics, this hypothesis was repeated by Chen, et al. (2010) and could possibly map to this study.

The orthoses had no discernible effects on any of the transverse knee moments the lack of significance concurs with Nester, et al. (2003) though they did report a non-significant increase in the external rotation moment; no such increase was seen in this study.

6.1.6. Ankle moment data - walking

6.1.6.i Comparison of sandals and shoes – ankle moments

The shoe condition had the effect of increasing the smaller plantarflexion moment over the sandal condition. This was initially thought to be due to the fact all the shoes in this study tended to have a small heel compared to the completely flat sandals. Sadeghi, et al. (2001) considered the functional role of the ankle in 20 subjects. They concluded the primary task was to support body weight against gravity; its second major role was to contribute towards forward propulsion. In the present study the larger dorsiflexion moment was unaffected by the footwear only conditions, but the range reflected the increase seen with the shoes in the plantarflexion moment ($p=0.001$, CI sandals 1.79 to 1.95, shoes 1.90 to 2.05). Chen, et al. (2010) labelled the larger moment the plantarflexion moment, which was reduced but not significantly by the shoes over barefoot walking, their “dorsiflexion” moment was significantly increased by the shoes.

All the coronal moments measured were significantly increased by the shoes over the sandals; the inverting moment was the largest, and when the range results were calculated they tended to follow a similar pattern to the inverting moment.

The rotational moments at the ankle were the smallest recorded in any plane. The external and internal rotation moments were nearly equal, but the shoes induced a

significant decrease in the internal rotation moment, but an almost identical increase in the external rotation moment. Overall when considering the moment range the shoes did produce a small but significant increase over the sandal condition.

6.1.6.ii Effects of orthoses - ankle moments

The raising effect of the shoes heel causing the increase in the plantarflexion moment was counteracted with a decrease when the heel was raised a little further with the $\frac{3}{4}$ orthoses in the shoes. There was however a small non-significant increase in the sandals. Nester, et al. (2003) maintained the effect of their orthoses was reflected mainly in the rearfoot complex and that the changes they saw in the sagittal plane were unexpected, though how the sagittal plane moment was affected was not detailed. When Chen, et al. combined orthoses with the shoes the larger moment was significantly reduced over the shoe only condition (1.22Nm/kg and 1.38Nm/kg respectively). In this study both orthoses tended to reduce the dorsiflexion moment and the moment range, but not by enough to produce any significant results.

The larger inverting moment was significantly increased by the full length orthoses, when considering the maximum inverting moment happened after heel lift it was perhaps not too surprising that the $\frac{3}{4}$ orthoses did not quite reach a significant level. Schmaltz, et al. (2006) also described the largest moment as occurring during late stance phase. They described it as the “maximum frontal moment in valgus” and suggested that this was the only prominent peak for this parameter; they reported a 14° medial wedge reduced this moment significantly.

The everting moment in the present study was reduced significantly by both medially wedged orthoses which would be expected, however this was much more evident in the sandals compared to the shoes. This was more likely to be due to the shoes already supporting the heel, as when looking at coronal plane motion of the calcaneal segment with respect to the shank there was little difference in the mean range data. The moment range tended to show small non-significant increases with both orthoses again being a little more evident in the sandal condition.

Both orthoses demonstrated a significant increase in the internal rotation moment over the footwear only conditions, though once again this was more evident in the

sandal condition. The external rotation moment tended to be decreased by the orthoses, it was only the full length orthoses that produced a significant result ($p=0.048$, CI sandals -0.059 to -0.018 , shoes -0.111 to -0.070). When combined with the sandals it was the full length orthoses that demonstrated a large increase in the transverse moment range again producing a significant result over the footwear only condition. However, when combined with the shoes the same orthoses produced a minimal reduction. This was in contrast to Nester, et al. (2003) who reported a significant reduction with their 10° medial wedge and suggested this was expected. The discrepancies seen between the studies could certainly be in part due to the modelling of the foot as Nester, et al. were using a single segment model. The average traces for both the coronal and transverse planes were obviously different in this study, the joint moments for the ankle were referenced to the shank whereas Nester, et al. referenced to a global co-ordinate system see Fig. 6.3. below. This highlights the obvious difference seen in the transverse plane even when using the same data. This substantiates the work of Liu and Lockhart, (2006) which is discussed later (chapter 10). When looking at the transverse moment range the shoes with both the orthoses tended to induce a very small decrease while the orthoses in the sandals produced a marked increase. The full length orthoses in the sandals being so marked as to produce a significant result.

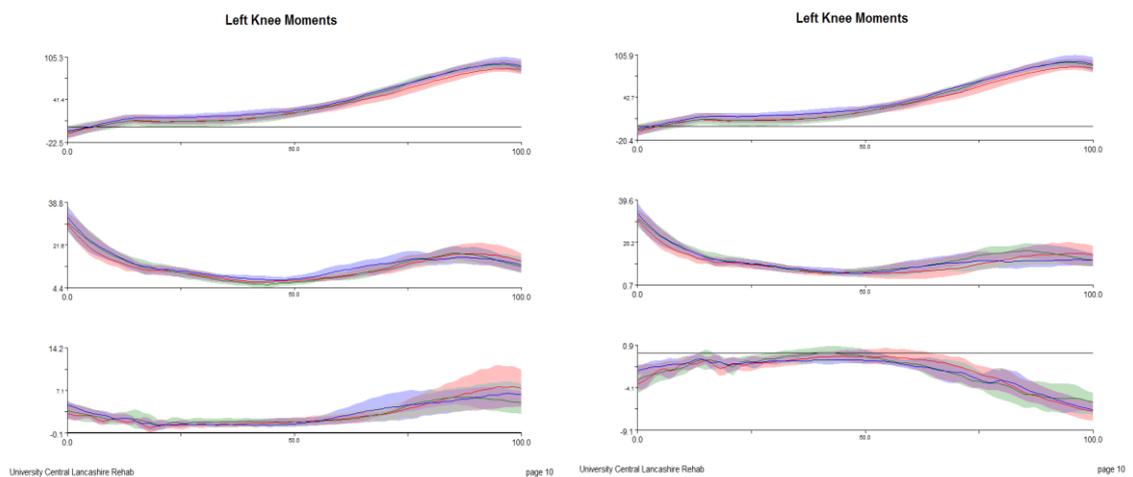


Fig. 6.3. Left knee moments for a single subject during step descent the graphs highlighting the differences the reference system has, (sagittal-top, coronal- middle, transverse-bottom) on the left are referenced to the shank while graphs on the right are referenced to the lab (global co-ordinate system).

6.1.7. Calcaneal segment - step descent

6.1.7.i Comparison of sandals and shoes - rearfoot

The shoes increased the mean range of dorsiflexion by 5° over the sandal condition which was significant ($p=0.000$, CI sandals 12.41 to 17.71, shoes 17.76 to 24.07), the difference being just a little more than the walking trials. It is difficult to compare this with previous studies as only Rao, et al. (2009) has reported on multi-segment foot motion during step descent, while no previous studies have investigated the effect of footwear or foot orthoses have during this common task.

Rao, et al. stated that step descent required greater ankle dorsiflexion, this lead to the assumption they were looking at the stance limb as this was not stated. Due to the protocol of this study comparing sandal and shoe conditions, only range was reported. This looked reduced during step descent over walking, but walking will be a combination of both dorsiflexion and plantarflexion (mean range 21° sandal condition), while step descent will be all dorsiflexion (mean range 15.6° sandal condition), this certainly could explain the difference between this study and Rao, et al. Due to the fact most subjects demonstrated a smooth transition into dorsiflexion from “hip max” to contact of the swing foot on the lower step. The increased range seen with the shoes versus the sandals could be put down to the heel raise, starting the sagittal range from a greater plantarflexed position. Due to this result it seemed logical that further investigations should possibly look at the forward continuum and lowering phases separately.

In this healthy cohort the mean range of coronal range motion was 4.9° (CI 3.90 to 5.68) and 3.9° (CI 3.04 to 4.83) in the shoes and sandals respectively which was less than reported in the walking trials. This would seem to contradict the results reported by Rao, et al. (2009) who stated step descent required greater calcaneus eversion, however they were reporting peak movement, therefore no direct comparison can be made with the present study, as due to the protocol only mean range could be reported. The 1° increase in the mean range between the shoe and sandal conditions was not significant.

The mean range in the transverse plane was within half a degree between the shoes and the sandals the shoes being greater, this was similar if a little less marked than the walking trials. It is not possible to compare this movement to any other studies at the present time.

Though there was a lack of significant results in the both the coronal and transverse planes between the shoe and sandal conditions this may indicate that it is acceptable to look at calcaneal segment motion with respect to the tibia when wearing shoes. This is important as most steps encountered outside of an individual's home will be approached whilst wearing shoes, also foot orthoses will only tend to be used with shoes and therefore it is essential that the effect of shoes is accounted for.

6.1.7.ii Effects of orthoses - rearfoot

When both orthoses were combined with the shoes it resulted in a further 5° increase in dorsiflexion range, this was not seen in the sandals which were relatively unchanged by either orthoses, which lead to a non-significant result. The most obvious explanation for the increase seen with the ¾ orthoses in the shoes could have been as a result of the increase in heel height, but this was contradicted when the same result was recorded with the full length orthoses. The lack of change when the orthoses were combined with the sandals could also be due to the relative instability- leading to compensation by the participants.

When looking at the effects of the orthoses on the coronal plane range of the calcaneal segment when combined with the sandals there was no measurable difference, while in the shoes the 5° wedge resulted in a range reduction of 0.6° which was also not significant.

In the transverse plane the orthoses did not induce any significant results though in the shoes there were minor reductions. There was no change in range seen with the ¾ orthoses in the sandals while the full length orthoses demonstrated a half degree increase. Again it is not possible to compare the results with any other previous studies as the use of orthoses on multi-segment foot function has not been previously reported.

6.1.8. Metatarsal segment - step descent

6.1.8.i Comparison of sandals and shoes - metatarsal

Similarly to the walking trials the shoes significantly reduced the total range of sagittal plane motion over the sandals. Rao, et al. (2009) found the first metatarsal exhibited significantly greater peak plantarflexion during step descent compared to walking and though there was an increase in total range of motion during step descent this was not significant. The findings of the present study demonstrated the opposite where both footwear only conditions during the step trials showed roughly half of the range seen when walking. It must be recognised that the method and foot model used were completely different, Rao, et al. used a magnetic tracking system (Flock of Birds) and a five segment model which separated the first metatarsal from the tarsal area (see Fig. 6.4.).



Fig. 6.4. Pictures from Rao, et al. (2009) showing magnetic tracking system adhered to the foot and lower leg (left) and how the foot was modelled (right).

Predictably the shoes significantly reduced the mean range of motion in the coronal plane over the sandal condition (3.1° , CI 2.45 to 3.52 and 5.0° CI 1.18 to 2.25 respectively). There is nothing in the literature which looks at coronal plane motion of the metatarsal segment during step descent barefoot or shod. From this study it would appear that as the heel lifts and the swing limb is lowering towards the smaller

step this segment tends to move towards inversion which may be as a result of ground reaction force moving towards the impact limb, or in some subjects may be used as an aid to dorsiflexion of the foot. The recommendation from this result was that further studies may consider splitting the range of motion to capture the forward continuance and controlled lowering phases.

The shoe condition reduced the range of transverse plane motion over the sandal condition ($p=0.000$, CI sandals 3.06 to 4.80, shoes 1.14 to 2.87). Considering this is the range of motion of the metatarsal segment on the calcaneal segment, resistance to movement will come from the width of the sole plus the heel counter of the shoe. Logically it could be the support of the shoe that produces this result; conversely it would be equally plausible for the foot to be moving within the shoe, hence the markers may not pick up the full range of motion. It is beyond the scope of this study to define the cause of this reduction but more research is needed in this area to establish the underlying effects recorded.

Table 6.2. Raw data for transverse midfoot mean range

Sub	Sand	Sand 34	Sand FL	Shoe	Shoe 34	Shoe FL
5	6.96	3.30	2.55	1.19	3.00	2.34
7	7.69	6.30	7.04	3.15	1.70	2.02
8	7.17	7.48	6.57	3.68	1.77	
12	1.15	2.62	4.20	1.37	1.11	1.78
13	2.64	2.46	4.91	4.26	4.31	4.57
14	0.95	1.80	1.53	1.28	1.52	1.79
15	6.37	6.63	8.54	1.03	1.16	1.48

It can be seen in Table 6.2. that subject 5 and 15 both displayed huge decreases from the sandal only to the shoe only condition, similarly subjects 7 and 8 who had the widest range in the sandals were more than halved by the shoes. However subjects 12, 13 and 14 all had small overall range in the sandals and demonstrated increased range in the shoes. This could be related to the type of foot wear as this was not controlled in this study in an attempt to relate it to clinical practice. This would be a convenient explanation; however subjects 12 and 13 wore running trainers for the trials which clinically would have been expected to “control” midfoot motion.

6.1.8.ii Effects of orthoses - metatarsal

When the orthoses were introduced in the sandals there was an increase of one degree with both orthoses while there were very minor reductions in the mean range when they were combined with the shoes. It was a similar picture in the coronal plane both orthoses in the shoes reduced the mean range by around half a degree, but when combined with the sandals the range was increased by a similar amount- resulting in no significant differences.

The transverse plane was a little different in the fact that the full length orthoses did produce a small increase in the shoes but not enough to show a trend in the data. The differing responses to the type of orthoses and when combined with the different footwear are also well demonstrated in the Table 6.2. This makes the explanation of what is happening difficult and could be due to footwear, comfort, foot type, or a combination of all three leading to the fact there may just be individual responses to orthoses which was proposed by Payne, et al. (2002).

6.1.9. Phalangeal segment - step descent

6.1.9.i Comparison of sandals and shoes - phalangeal

Just as in the walking trials the shoes significantly reduced the range of dorsiflexion of the phalangeal segment on the metatarsal segment, the suggestions given for why this may happen are just as pertinent to step descent. The range was again less than the walking trails which was converse to the results published by Rao, et al. (2009) who presented an increased range and increased peak dorsiflexion at the first metatarso-phalangeal joint in their cohort during step descent. Differences in the modelling and measurement technique may account for this discrepancy; the findings of this study could also be non-comparable with barefoot motion. Though the markers were attached to the skin in the sandal trials it is plausible that the subjects flexed their toes to aid with holding the sandals on, as any instability introduced in the more demanding step descent may introduce more radical compensations.

6.1.9.ii Effects of orthoses - phalangeal

Both orthoses in the shoes and sandals reduced the range further though this was much more marked in the shoes, there was no significant reduction. As with the walking trials further research is needed to define the cause of this non-significant reduction, and to investigate whether this would have any clinical relevance.

6.1.10. Knee kinematic data - step descent.

6.1.10.i Comparison of sandals and shoes - knee

Knee flexion range was unaffected by the type of footwear the mean range being 56.1° (CI 52.16 to 60.08) in the shoes and less than a degree more in the sandals, this was within one degree of Grenholm, et al. (2009) control group though they were only looking at women. Andriacchi, et al. (1980) recorded a mean peak flexion angle of 68.9° which looks a lot higher than this study but as most subjects in the present study started somewhat flexed at “hip max” the ranges could possibly be similar. Protopapadki, et al. (2007) reported a maximum flexion angle 90.5° which is higher than this study. There are many factors that will affect the results the technique of how the knee was modelled and recorded the height of the steps and the height of the subjects to name some, all could have an effect on the flexion angles.

The range of abduction/adduction of the knee was unaffected by the type of footwear worn for the trials the mean range being within 0.3° (3.9° CI 2.67 to 5.09 and 3.6° CI 2.36 to 4.79 for shoes and sandals respectively). It is difficult to find other references in the literature to compare coronal knee angle as most experiments concentrate on sagittal plane angles with coronal kinetics rather than kinematics Selfe, et al. (2007) did not report on the range but it would appear to be around 4.5° from a non-intervention graph they published, which is not dissimilar to this experiment.

In the transverse plane a similar non-significant increase was induced by the shoes over the sandals ($p=0.111$), the 4.9° (CI 3.87 to 5.87) and 4.4° (CI 3.43 to 5.43) ranges seen here were comparable to the graphs published by Selfe, et al. (2007). Again it is difficult to find other papers that report transverse plane rotation of the knee during step descent, Tillman, et al. (2003) did report a significant decrease in

internal rotation of the tibia but they were looking at the impact limb from a 43cm drop test.

6.1.10.ii Effects of orthoses - knee

Neither orthoses induced any significant effect on knee flexion during step descent though all the mean ranges tended to be a fraction less when the orthoses were combined with the footwear. Again in the coronal plane all the mean ranges were reduced marginally by both orthoses but this was not significant. In the transverse plane the $\frac{3}{4}$ orthoses tended to reduce the mean range marginally, while the full length orthoses tended to increase the mean range, none of these results were significant. It must be remembered that all the subjects used for this experiment had no knee problems during the trials and therefore it must be assumed that their knees were stable. Further investigations are necessary on subjects with knee pain to explore the effects of these devices on subjects who would be prescribed this form of treatment (which will be discussed later).

6.1.11. Knee moment data - step descent

6.1.11.i Comparison of sandals and shoes – knee moments

Sagittal: The mean maximum flexion moment was much higher during step descent than when walking (1.35Nm/kg to 0.83Nm/kg respectively for the sandal only condition) Protopapadaki, et al. (2007) reported their maximum external knee flexion moment as 0.46Nm/kg, but also had a much larger extension moment (0.40Nm/kg). It must be noted that this previous study was not using an initial step for their data, Protopapadaki, et al. did note that subject height step height marker placement and how the analysis was conducted may all influence the results obtained. The result produced in this study was a little higher than 1.11Nm/kg Novak and Brouwer, (2010) reported for the non-dominant limb of their younger subjects; they labelled this as their peak extensor moment. Novak and Brouwer, also highlighted speed of descent will affect the moment, this was not controlled in this study again to try and make it as clinically relevant as possible. The footwear type made no significant difference to the sagittal plane moments.

The maximum adduction moment was not significantly affected by the type of footwear; the mean maximum moment (0.55Nm/kg CI 0.49 to 0.60 for the sandal condition) in this cohort was a little larger than that reported by Selfe, et al. (2007) at 0.39Nm/kg. None of the participants demonstrated an abduction moment when descending the step and hence it was labelled as the minimum adduction moment. Like the maximum moment there were no significant differences introduced by the footwear. This was reflected in the coronal moment range where the footwear only results were essentially equal.

Similarly to the coronal plane there were no significant differences seen in the transverse plane between the footwear conditions. The smaller moment in the transverse plane was labelled as the minimum internal moment however four out of the fifteen normal subjects did demonstrate a small external moment, and as such there was more variation seen. The mean rotational moment range (0.10Nm/kg, CI 0.08 to 0.13) was similar to that recorded by Selfe, et al. (2007) 0.121Nm/kg but again it was unaffected by the type of footwear.

6.1.11.ii Effects of orthoses – knee moments

To the authors knowledge no other studies have investigated the effect of foot orthoses on knee moments during step descent, therefore no useful comparisons can be drawn from previous literature. Neither of the orthoses made any significant changes in the maximum, minimum or range of flexion moment, however the $\frac{3}{4}$ orthoses did tend to increase the maximum flexion moment and the moment range; these trends were reflected in the data. The reaction to the full length orthoses was more variable where there were both minor increases and decreases seen.

The maximum adduction moment demonstrated small non-significant increases and decreases with both orthoses over the mean footwear only conditions. The minimum adduction moment did show a 0.05Nm/kg increase in the mean data with the full length orthoses when combined with the shoes. The orthoses did tend to reduce the moment range in both the shoes and the sandals but not significantly this was more evident with the shoes.

There were no significant differences seen in the transverse plane with either of the orthoses. The bar charts in Fig. 5.30. illustrate that the full length orthoses did have more of a tendency to increase the maximum internal moment and the minimum internal moment a little more than the $\frac{3}{4}$ orthoses which was reflected in the data. Though the full length orthoses demonstrated this mild trend there was no trend seen within the transverse moment range. The non-significant results seen in all planes certainly may be due to using a “normal” cohort but also lead to decision that it may be more pertinent to look at the forward continuum and lowering phases as separate entities during normal speed step descent in future studies.

6.1.12. Ankle moment data - step descent

6.1.12.i Comparison of sandals and shoes – ankle moments

Ground reaction force (GRF) remained in front of the ankle for the whole of the step trials and as such all the values calculated for all subjects were negative denoting dorsiflexion moments. The maximum dorsiflexion moment was not significantly affected by the footwear type though there was a small reduction with the shoes over the sandal only condition. The mean maximum dorsiflexion moment calculated in the present study (-1.24Nm/kg, CI -1.30 to -1.18) was a little lower than 1.38Nm/kg reported by Protopapadaki, et al. (2007) who used an 18cm step, and higher than Novak and Brouwer, (2010) who published 1.07Nm/kg for the dominant limb of their younger group using a 15cm step, Cluff, et al. (2011) used a 20cm step the graph presented for the initial step recorded a peak plantarflexor moment of 1.0Nm/kg. When looking at the minimum dorsiflexion moment the sandals had a higher mean than the shoes, but again this was not significant. The moment range however was significantly reduced by the shoes over the sandals, though the mean difference was small. When looking at the raw data 7 of the 15 subjects had a greater range with the shoes the large reductions demonstrated by subjects 5 and 15 are possible contributing to the overall mean difference (see Table 6.3.).

Table 6.3. Raw data showing results of sagittal plane ankle moment range and differences (Nm/kg)

Sub	Sand	Shoe	Diff
1	0.79	0.75	0.03
2	0.50	0.49	0.02
3	0.77	0.78	-0.01
4	0.79	0.83	-0.04
5	0.94	0.58	0.36
6	0.86	1.00	-0.14
7	0.84	0.86	-0.01
8	0.57	0.76	-0.18
9	0.55	0.67	-0.12
10	0.57	0.47	0.11
11	0.82	0.70	0.12
12	0.45	0.56	-0.11
13	0.73	0.54	0.19
14	0.86	0.82	0.03
15	0.64	0.36	0.29

The maximum inverting moment was not altered by the type of footwear, the maximum coronal moment recorded in the present study was much higher (0.41Nm/kg CI 0.34 to 0.48) than that reported by Novak and Brouwer, (2010) who recorded 0.12Nm/kg(0.04) though as mentioned earlier their step height was only 15cm which may in part explain some of the discrepancy. The minimum inverting moment was small and the large standard deviations illustrated in the bar charts of Fig. 5.33. reflects that five of the fifteen subjects had a small everting moment with the sandals and three were similar with the shoes.

The transverse moments were the most variable between subjects; the maximum internal moment was the smallest and 6 of the subjects in the shoe only condition did not attain an internal value. The shoes did significantly reduce this moment over the sandal condition. The maximum externally rotating moment was larger and was significantly increased by the shoes over the sandals, the rotational moment range was almost imperceptibly but significantly reduced by the shoes over the sandals, however this would have to be described as a statistical result rather than a clinically relevant one, the magnitude of the result would not exceed the minimum detectable change.

6.1.12.ii Effects of orthoses – ankle moments

The maximum dorsiflexion moment demonstrated increases when the orthoses were combined with the shoes and decreases with the sandals leading to no significant results. Both orthoses demonstrated increases in the minimum dorsiflexion moment over the footwear only conditions this lead to a trend being seen in the data which was more marked with $\frac{3}{4}$ orthoses. The orthoses tended to reduce the range of sagittal plane moment in the sandals but increased it when combined with the shoes; therefore there were no significant changes seen. The changes recorded in the sagittal plane moments could not be mapped to the results seen in the angular data.

When looking at the maximum inverting moment the orthoses tended to increase the mean values but not significantly and no trend was seen within the data. The minimum inverting moment demonstrated increases with the $\frac{3}{4}$ orthoses in both the shoes and the sandals however there was no significant change recorded. The full length orthoses did however, significantly increase the minimum moment, overall the coronal moment range was unaffected by orthoses.

The maximum internal moment demonstrated large increases when the orthoses were combined with the sandals. There was also a small increase seen with the full length orthoses in the shoes hence the full length orthoses produced a significant increase. The maximum external moment was reduced by both orthoses in both footwear types and this was significant. When looking at the transverse moment range the orthoses tended to increase the range in the sandals but decreased with the shoes therefore there were no significant changes were recorded with the orthoses.

6.2. Summary of key findings Chapter 6:

Calcaneal segment:

- The rearfoot sagittal plane range was increased in both the walking and the step trials by the shoes over the sandals. In the transverse plane the shoes recorded a significant increase during walking but this was not repeated in the step down trials.
- During walking there was an increase seen with the $\frac{3}{4}$ orthoses in the sagittal plane, only the full length orthoses significantly decreased the range in early

stance. However both orthoses decreased the range of motion during the propulsive phase of stance. Neither of the orthoses produced any statistically significant changes during the step down.

Metatarsal segment:

- The shoes significantly decreased the range of motion of the metatarsal segment in all planes during walking and step down.
- Both orthoses significantly decreased coronal plane range during the propulsive phase of walking, but there were no significant results to report during the step down trials.

Phalangeal segment:

- The shoes significantly reduced the sagittal plane range of motion recorded during both the walking and step trials.
- The full length orthoses reduced the range further during the walking trials only; the range during step was not reduced significantly.

Knee kinematics:

- Maximum flexion at heel strike and toe off were reduced significantly during walking while maximum extension was increased with the shoes compared to the sandals. There were no differences in any plane during step down.
- Only maximum flexion at toe off was reduced by the $\frac{3}{4}$ orthoses during walking, there were no significant differences seen during step down in the knee.

Knee moments:

- The shoes significantly increased the sagittal moment range and the external rotating moment during the walking trials over the sandal condition. There were no differences recorded during step down between the footwear conditions.
- Only the full length orthoses increased the adduction moment and the coronal moment range during walking trials, the step down trials were unaffected by any of the orthoses.

Ankle moments:

- During walking the plantarflexion moment and sagittal plane moment were both increased by the shoes over the sandals. All the coronal moments, the

external moment and transverse moment range similarly were increased by the shoes only the internally rotating moment was decreased. The step down trials demonstrated sagittal moment range was decreased by the shoes, similarly to the walking trials the internal and external moments were decreased and increased respectively. The rotational moment range was however reduced by the shoes.

- Only the full length orthoses increased the inverting moment but both orthoses reduced the eversion moment, but increased the internal rotation moment during walking. During the step trials the full length orthoses increased the minimum inverting moment and the maximum internal rotating moment. Both orthoses reduced the maximum external rotation moment during step down.

6.3. Initial Conclusions

By modelling the foot using CAST (calibrated anatomical system technique) it would appear possible to place markers on a shoe and still record some of the changes induced by orthoses. The calcaneal segment demonstrated equal results in the coronal plane between the shoe and sandal conditions for the first and second half of stance phase. In the sagittal plane though there were discrepancies between the sandals and shoe trials. This method was able to pick up small differences induced by the heel raise effect of the $\frac{3}{4}$ orthoses, therefore it seems reasonable to assume the differences found between the footwear conditions is more effect rather than artefact. It was the transverse plane where conclusions must be drawn with caution as this study was unable to decipher if the changes were induced by slippage of the foot within the shoe or whether there was a support effect, this was beyond the scope of this experiment but further research is needed to establish what is being measured. Considering these results it was decided to record range of dorsiflexion and plantarflexion in the next part of the study, the coronal plane results were thought to record the necessary findings. However it was felt the transverse plane may be better examined by looking at early and late stance phase rather than just total excursion.

The main effect on the midfoot was induced by the shoe condition over the sandals which in isolation could have been apportioned to the foot slipping within the vamp of the shoe. However due to the full length orthoses restricting coronal plane motion, as would be expected from a clinical viewpoint, the technique used in this experiment would appear to be the first to be able to describe small restrictions in motion within normal non-modified footwear. Due to the results in both the calcaneal and metatarsal segments it was suggested that when moving on to the second study with the patellofemoral pain subjects it was acceptable only to use the shoe condition for all the trials. However it was deemed pertinent to consider all planes of the metatarsal segment during early and late stance, as it was felt important to capture the propulsive phase separately.

The knee angle parameters measured in this study were generally unaltered by the orthoses, all except maximum flexion at toe off when walking. This was a little unexpected as clinically it is the coronal and transverse planes that are purported to be altered by limiting coronal plane motion at the foot. However the cohort of subjects here had no knee pain and therefore may have relatively stable knees which could be less affected by external sources, this leads on to the next part of this study. The changes seen between the sandal and shoe data were attributed to the instability seen in the sandals, therefore it was recommended that using only the shoe data in the second study would provide more realistic knee data.

Where the ankle moments were significantly altered this tended to be due to the shoes over the sandal condition, though some of the extremes and ranges were effected significantly by the orthoses, this was not transferred to the knee moments in most cases. This would suggest that not all reductions in foot and ankle motion is transferred to the knee and may suggest it can be absorbed by the distal structures.

6.3.1. Recommendations for following study

It was highlighted at the beginning of the initial study that it was only possible to compare the range of motion between the sandals and the shoes as small discrepancies in marker placement could have affected maximum positions without the foot moving. In the second study only shoes are going to be used for all the trials

it will therefore be possible to report on maximum position as well as excursion This will allow more comparisons with previous studies and provide more objective measurements to compare the patellofemoral pain group with the control group. It was also considered crucial in the step down trials to look at the forward continuum phase and lowering phases as separate phases, as it was thought the lowering phase may be more challenging to the symptomatic group.

Chapter 7. Method Symptomatic Subjects

7.1. Aims of patellofemoral pain study/Hypothesis.

The aims of this study are to investigate the effects of three quarter and full length orthoses have on the knee and foot mechanics during walking and descending steps; on normal subjects with pronated feet compared with subjects with patellofemoral pain and over pronated feet.

Objectives:

1. Compare the kinematics of the knee and foot segments during walking and descending steps between the symptomatic and non-symptomatic groups when wearing shoes alone.
2. Compare the kinetics of the knee and ankle during walking and descending steps between the symptomatic and non-symptomatic groups when wearing shoes alone.
3. Compare the effects of the three quarter and full length orthoses have on the different segments of the foot and the knee when combined with shoes.
4. To explore the clinical importance of any differences found.

It was hypothesised that it was possible to identify differences in the kinematics and kinetics between the normal subjects and the subjects with patellofemoral pain, and to determine if foot orthoses have the same effects on the two groups.

7.2. Method Symptomatic Subjects

This experiment was conducted in order to compare the kinematics and kinetics of a pain free group and a patellofemoral pain group of subjects when walking and descending a step at a self-selected pace. While completing these tasks they would wear their normal footwear either on their own or combined with two different orthoses. When investigating the effects of the orthoses each subject acted as their own control, the trials with the shoe only condition was used as the internal control.

It was intended to recruit fifteen subjects with patellofemoral pain from the University of Central Lancashire and Wrightington, Wigan and Leigh NHS

foundation Trust. Before any subjects were recruited ethical approval was obtained from National Research Ethics Service Committee North West Greater Manchester South (Rec reference no. 11/NW/0362, Appendix 1). The committee were concerned that a CE marked product (slimflex™ insole) was being modified by moulding it to the subjects' foot though this is usual in clinical practice. They therefore requested that the MHRA (Medicines and Healthcare products Regulatory Agency) was contacted to obtain an exemption certificate. The MHRA confirmed that as the product was still being used for its intended purpose (i.e. as an insole) it was not being used outside of its CE mark and therefore there was no need for an exemption certificate, the committee accepted this communication.

Approval was also obtained from the University of Central Lancashire ethics committee (Ref. No. BuSH 082 Appendix 1). After a successful poster campaign within the university all the symptomatic subjects were recruited from this source.

7.2.1. Subjects with Patellofemoral Pain

Fifteen subjects were recruited from the University of Central Lancashire (8 males; 7 females, mean age 28.6 years SD 5.83); participants were included in the study if they regularly had a score of 3 on the numeric pain rating scale when descending stairs. All subjects had to have foot pronation with a score of at least +6 on the foot posture index (Redmond, et al. 2006; Evans, et al. 2003). Participants had to be aged between eighteen and forty years to limit the risk of them having patellofemoral arthritis. Previous surgery, an ataxic gait, back pain, or balance problems also excluded them from the study.

7.2.2. Footwear Conditions (Symptomatic subjects)

For this second phase of the study all subjects completed the walks and step descents under the following footwear conditions.

- Shoes only.
- Shoes with moulded orthoses and 5° three quarter wedge.
- Shoes with moulded orthoses and 5° full length wedge.

For the control group the shoe data that was recorded for comparison with the sandal data was reused all these subjects were pain free at the time of recording the data, though all of them had pronating feet.

7.2.3. Procedures

The walking trials were repeated to exactly mimic the initial study, five central hits on the force plate being required for each walking trial, the data from the most painful limb being used for the patient trials. For the step trials the patellofemoral pain subjects were asked to descend with their pain free limb, or where the symptoms were bilateral they lead with the least painful limb, hence the eccentrically controlling limb was the most affected side.

Chapter 8. Results Symptomatic Subjects

8.1. Results symptomatic subjects - walking

8.1.1. Calcaneal segment - walking

Movement in the sagittal plane was split into range of dorsiflexion and range of plantarflexion there was no difference seen between the pain free cohort and the patellofemoral pain subjects in the range of plantarflexion (See Fig.8.1). Reaction to the orthoses interventions was also similarly non-significant.

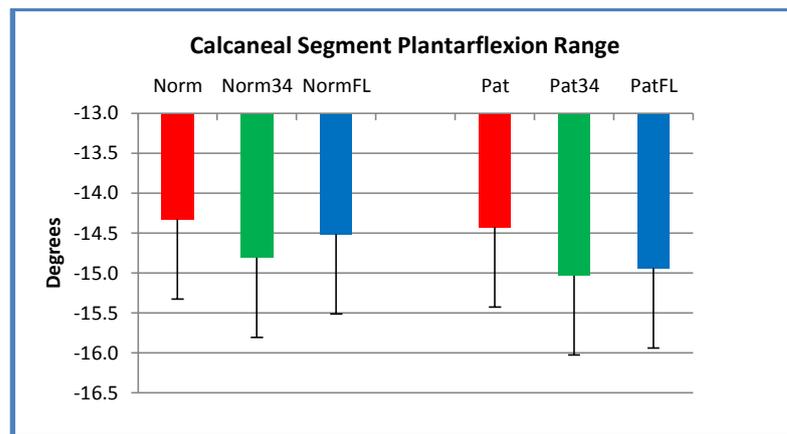


Fig.8.1. Bar chart demonstrating range of plantarflexion between the normal and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

Dorsiflexion range was less than plantarflexion, the range demonstrated by the patients was significantly reduced over the normal subjects ($p=0.016$). Neither orthoses produced any differences in the mean range when the patient and normal data were combined. The bar chart in Fig.8.2. illustrates the $\frac{3}{4}$ orthoses tended to elicit an increase in both groups, however it must be remembered the maximum mean difference was just over one degree.

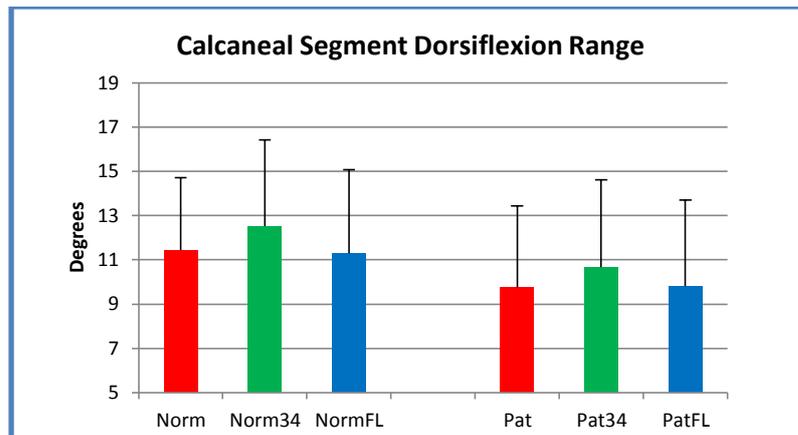


Fig.8.2. Bar chart demonstrating range of calcaneal dorsiflexion between the normal and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

In the coronal plane there was no difference seen between the patient and normal groups during early stance when looking at the mean maximum eversion position. Both orthoses reduced this maximum position. The mean difference was almost 2° with the full length orthoses which produced a significant reduction (p=0.021). The ¾ orthoses did not produce the same significant reduction but there was certainly a trend with a mean reduction of 1.5° (p=0.062, see table 8.1.).

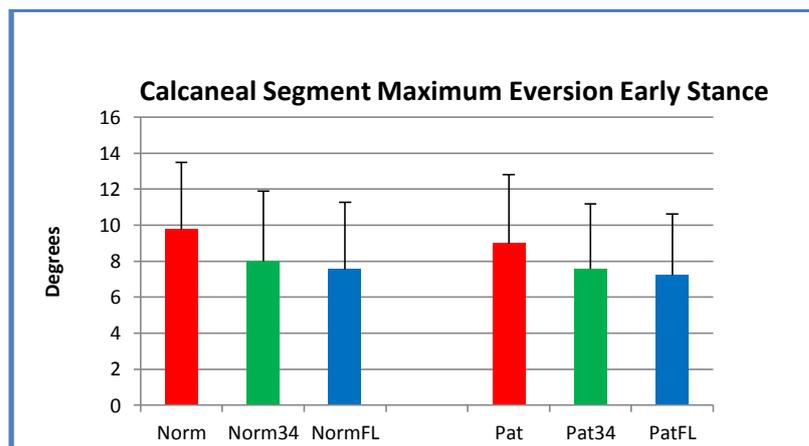


Fig.8.3. Bar Chart demonstrating maximum calcaneal eversion position with respect to the shank during early stance, comparing normal group to patient group and how they reacted to the orthoses (Error bars = standard deviation)

Similarly, when considering the coronal excursion of the calcaneal segment to the shank there was no difference seen between the groups under investigation. The full length orthoses still demonstrated a significant reduction (p=0.007) but this time

there was no trend seen with the ¾ orthoses, however the bar chart Fig.8.4. does illustrate the small reduction seen (see Table 8.1.).

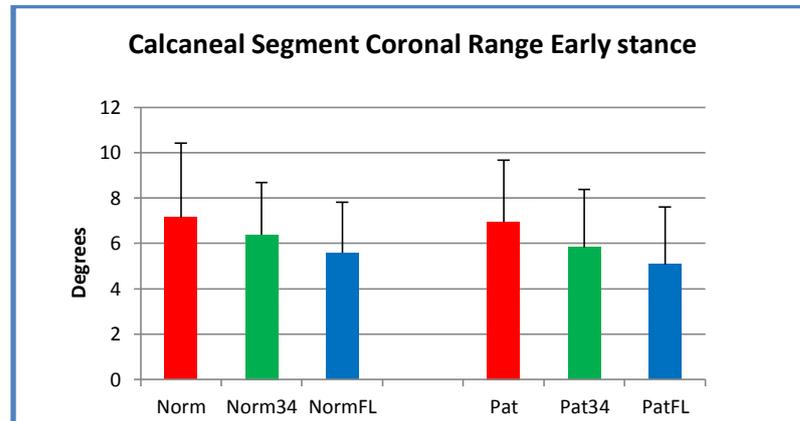


Fig.8.4. Bar Chart demonstrating maximum calcaneal eversion position with respect to the shank during early stance, comparing normal group to patient group and how they reacted to the orthoses (Error bars = standard deviation)

During late stance there were no differences seen between the normal and patient groups when considering both maximum eversion position and the coronal plane excursion, there were also no significant differences induced by either orthoses (See Table 8.1.).

Maximum calcaneal segment rotation positions and excursions were recorded with respect to the shank. There were no significant results recorded either between the subjects groups or between the shoes versus the shoe plus orthoses conditions (See Table 8.1.).

Table 8.1. Mean range of motion of the calcaneal segment with respect to the shank comparing the normal and patient group data and recording any differences seen with the orthoses.

Calcaneal range of motion with respect to the shank	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				¾	FL	¾	FL
Plantarflexion ROM	0.509	P=0.494	-0.960 to 1.978	0.929	0.348	P=0.308	P=0.703
Dorsiflexion ROM	1.688	P=0.016	0.319 to 3.018	-0.992	0.056	P=0.236	P=0.947
Max Eversion Early Stance	0.517	P=0.455	-0.849 to 1.884	1.591	1.984	P=0.062	P=0.021
Coronal Range Early Stance	0.197	P=0.701	-0.816 to 1.211	0.598	1.718	P=0.341	P=0.007
Coronal Range Late Stance	-1.585	P=0.177	-3.894 to 0.725	0.479	0.603	P=0.738	P=0.674
Transverse Range Early Stance	0.642	P=0.168	-0.274 to 1.557	-0.379	-0.148	P=0.504	P=0.794
Transverse Range Late Stance	0.823	P=0.404	-1.121 to 2.766	0.289	0.570	P=0.804	P=0.637

8.1.2. Metatarsal segment - walking

During early stance phase sagittal plane movement of the metatarsal segment did not demonstrate any differences between the patient and normal groups, or between the shoe only condition and the orthoses. However, during the late stance phase the patient group did reveal a significant increase over the control group ($P=0.003$, see Table 8.2.). Fig.8.5. highlights that though this was a significant change the mean difference was just over one degree.

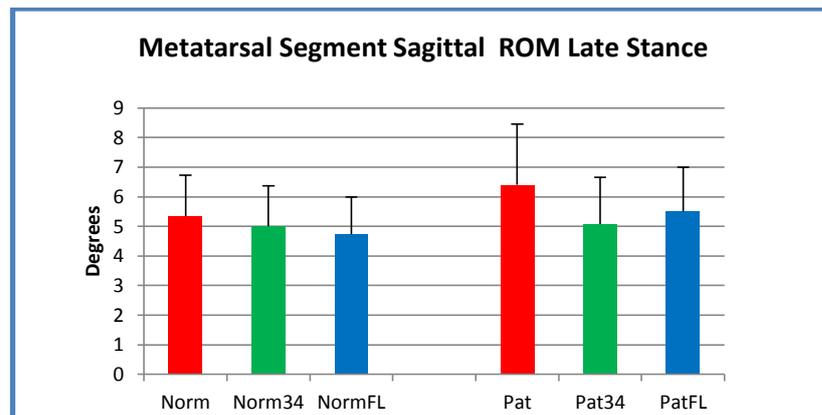


Fig.8.5. Bar chart showing the sagittal plane excursion of the metatarsal segment with respect to the calcaneal segment, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Fig.8.5. also displays that the $\frac{3}{4}$ orthoses produced the greatest mean reduction with the patient group. Although there was also a minimal decrease seen in the normals this was not enough to produce a significant result. However, there was a trend seen within the confidence intervals (Table 8 Appendix 2). In contrast the full length orthoses produced a reduction in both groups and as such demonstrated a significant result ($p=0.032$) see Table 8.2.

In the coronal plane during early stance there was a significant if small increase seen in the mean range of the patient group over the normal group. Both orthoses did reduce the range further in both groups as illustrated by Fig.8.6. but neither produced a significant result. However both did show a trend in the confidence intervals this was more evident for the full length orthoses (Table 9 Appendix2).

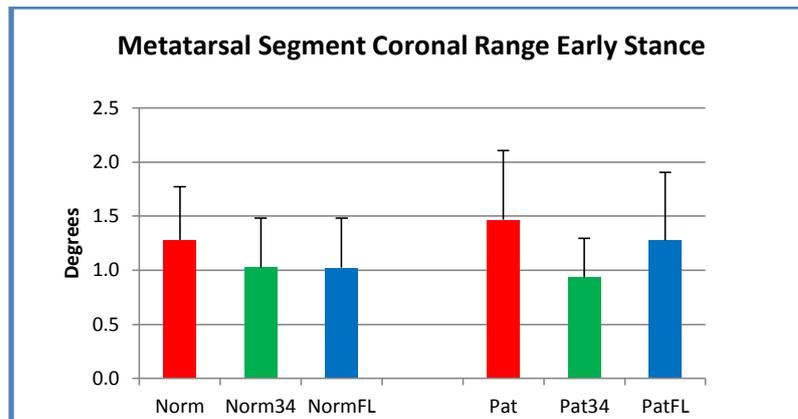


Fig.8.6. Bar chart showing the coronal plane excursion of the metatarsal segment with respect to the calcaneal segment during early stance phase, comparing the patient and normal groups, and how they reacted to the orthoses (Error bars = standard deviation)

Both groups produced very similar ranges for the late stance phase; hence there were no differences to report. However, both orthoses reduced the range of motion significantly over the shoe only condition (see Fig.8.7.). The mean difference was a little over half of one degree with both (see Table 8.2.) but the level of significance ($p=0.000$) would suggest this was seen in all subjects.

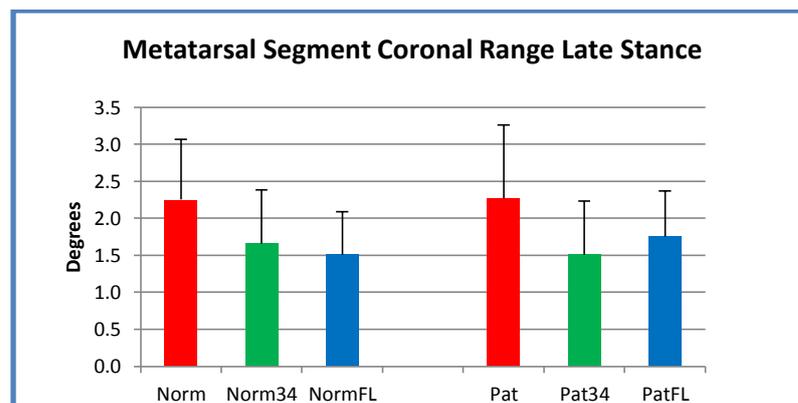


Fig.8.7. Bar chart showing the coronal plane excursion of the metatarsal segment with respect to the calcaneal segment during late stance phase, comparing the patient and normal groups, and how they reacted to the orthoses.

During both the early and late stance phase in the transverse plane there were no differences seen between the patient and the control groups. However both orthoses did produce a significant reduction in the range of adduction in early stance (see Table 8.2.). The mean differences were again small but highly significant (both $p=0.000$), this is well illustrated by Fig.8.8.

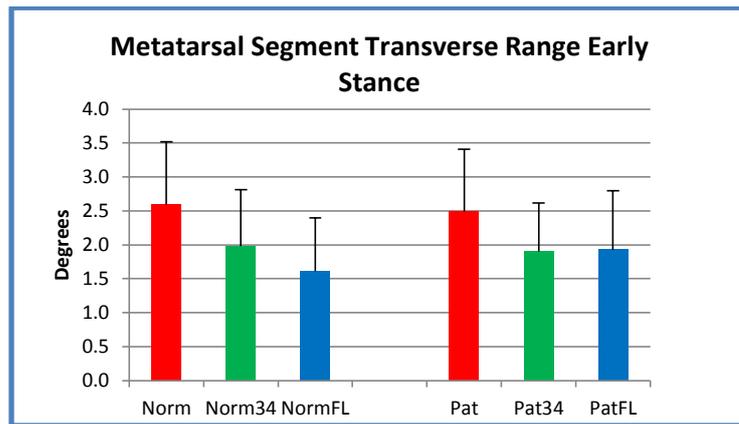


Fig.8.8. Bar chart showing the transverse plane excursion of the metatarsal segment with respect to the calcaneal segment during early stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

During late stance it was only the $\frac{3}{4}$ orthoses that demonstrated a significant reduction in the transverse range of the metatarsal segment; this was demonstrated by the bar charts in Fig.8.9. Although there was a greater range seen during this phase the mean reduction was no greater (maximum change 1° - seen in the control group).

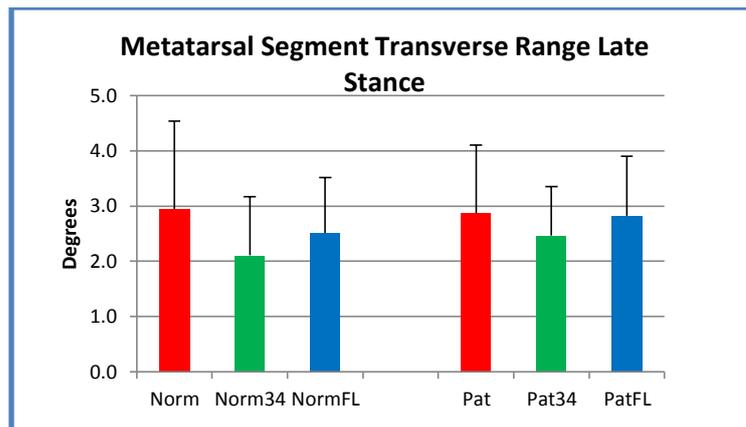


Fig.8.9. Bar chart showing the transverse plane excursion of the metatarsal segment with respect to the calcaneal segment during late stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The full length orthoses posted no difference in the mean range in the patient group. There was a small reduction in the control group, but not enough to record any trend within the data.

Table 8.2. Mean range of motion of the metatarsal segment with respect to the calcaneal segment comparing the normal and patient group data and recording any differences seen with the orthoses.

Metatarsal range of motion with respect to the Calcaneal Segment Walking	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Sagittal ROM Early Stance	0.238	P=0.280	-0.196 to 0.672	-0.299	-0.296	P=0.267	P=0.274
Sagittal ROM Late Stance	-0.842	P=0.003	-1.389 to -0.285	0.529	0.746	P=0.126	P=0.032
Coronal ROM Early Stance	-0.259	P=0.009	-0.453 to -0.065	0.175	0.223	P=0.147	P=0.066
Coronal ROM Late Stance	-0.093	P=0.493	-0.360 to 0.493	0.596	0.630	P=0.000	P=0.000
Transverse ROM Early Stance	-0.093	P=0.552	-0.403 to 0.246	0.528	0.772	P=0.007	P=0.000
Transverse ROM Late Stance	-0.037	P=0.867	-0.469 to 0.396	0.863	0.243	P=0.002	P=0.367

8.1.3. Phalangeal segment - walking

Mean maximum dorsiflexion of the phalangeal segment on the metatarsal segment was a little less in the patient cohort, but the 2.65° difference seen in Fig.8.10. was not significantly less than the control group.

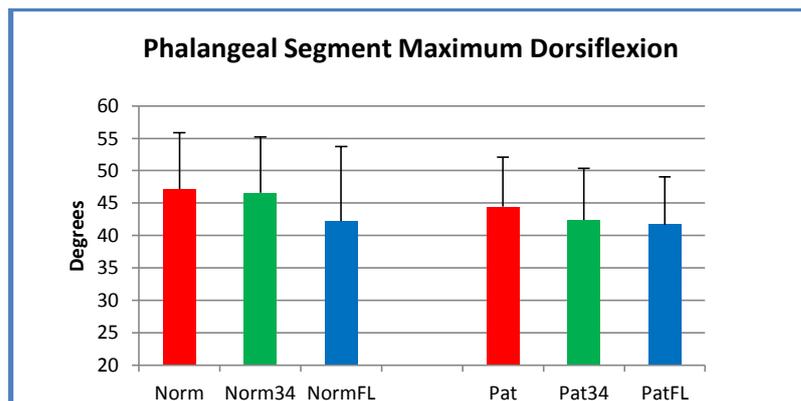


Fig.8.10. Bar chart showing the maximum phalangeal segment flexion with respect to the metatarsal segment during stance phase, comparing the patient and normal groups and how they reacted to the orthoses

Both subject groups displayed reductions in maximum dorsiflexion with each orthoses; however the reduction with the ¾ orthoses in the normal group was very small. Both orthoses reduced the maximum position in the patient group, however it was the full length orthoses in the non-symptomatic subjects which showed almost a

5° reduction. The full length orthoses did not produce a significant result, however there was an obvious trend in the data towards a reduction (see Table 8.3.).

When looking at the mean excursion of the phalangeal segment again there was no statistical difference recorded between the two groups. This time the full length orthoses produced a significant reduction versus the shoe only condition (see Table 8.3.) the greatest reduction in the maximum position was reflected in the mean range. The bar charts in Fig.8.11. also suggested that the ¾ orthoses had more of an effect on both groups, but the small reductions did not lead to a significant result, although there was somewhat of a trend seen within the data (see Table 8.3).

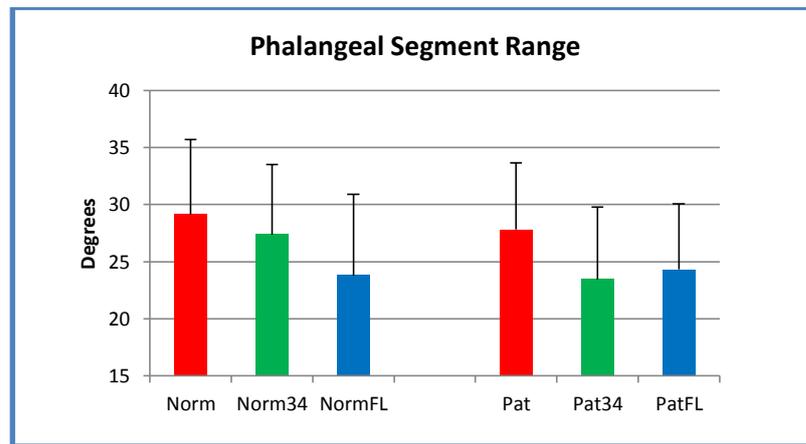


Fig.8.11. Bar chart showing the phalangeal segment excursion with respect to the metatarsal segment during stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation).

Table 8.3. Mean range of motion of the phalangeal segment with respect to the metatarsal segment comparing the normal and patient group data and recording any differences seen with the orthoses.

Phalangeal range of motion with respect to the Metatarsal Segment Walking	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Maximum Dorsiflexion	2.168	P=0.192	-1.104 to 5.440	0.827	3.846	P=0.683	P=0.060
Sagittal Plane ROM	0.848	P=0.469	-1.462 to 3.158	1.973	4.443	P=0.169	P=0.002

8.1.4. Knee kinematics - walking.

Although maximum first peak flexion did not show a significant difference between the symptomatic and pain-free groups, the confidence intervals suggested there was a mild trend towards the patients demonstrating a little less flexion (see Table 8.4.). Fig.8.12. illustrates there was only a small mean difference between the groups and that reaction to the full length orthoses was almost identical, though again not significant. The $\frac{3}{4}$ orthoses left the mean flexion unaltered in the patient group while inducing a small increase in the pain free subjects.

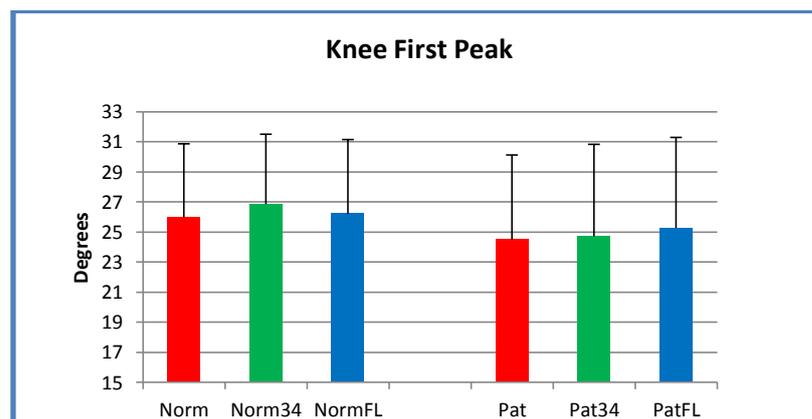


Fig.8.12. Bar chart maximum first peak flexion following initial contact during stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The only significant difference in the sagittal plane between the patient and the normal groups was seen when the knee was at its most extended position (see Table 8.4.). Fig.8.13. illustrates that most subjects remained slightly flexed though there were some subjects who attained an extended position. There were no significant differences induced by the orthoses - the confidence intervals confirming there were no trends seen either way.

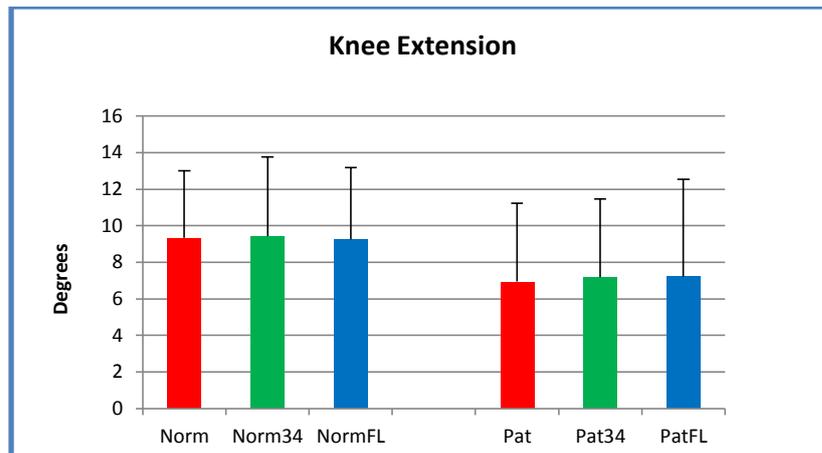


Fig.8.13. Bar chart showing maximum extension (minimum flexion) during stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Again there was no significant difference between the groups when looking at maximum knee flexion at the end of stance phase. However, the confidence intervals did suggest there was a mild trend to the patient group demonstrating less flexion, this was illustrated in Fig.8.14. The orthoses had little effect on the mean differences but Fig.8.14 would suggest that the two groups tended to react differently to the orthoses.

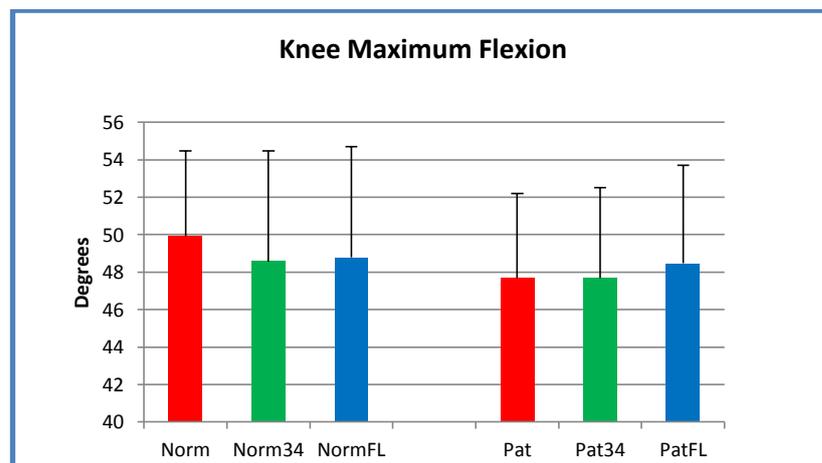


Fig.8.14. Bar chart showing maximum knee flexion during stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation).

In the coronal plane maximum adduction was significantly increased in the patient group over the normal cohort, this is illustrated by Fig.8.15. The large error bars show there was a wide variation between the individuals in both groups. The

orthoses made little difference to the position with no evident trends in the data (see Table 8.4.).

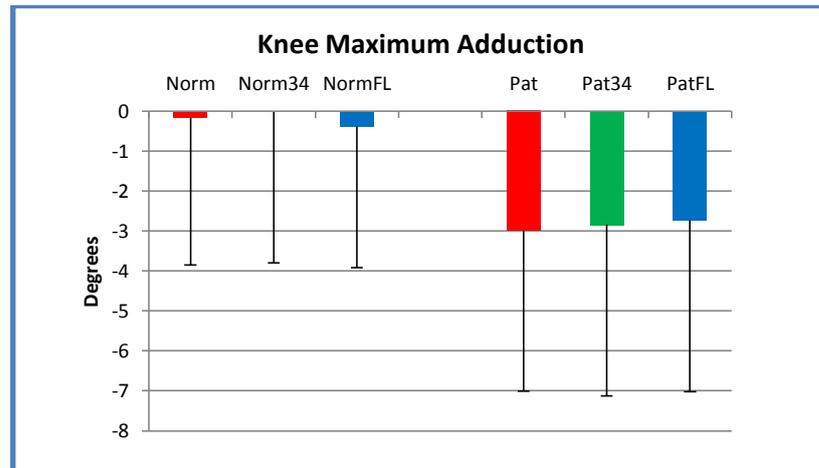


Fig.8.15. Bar chart showing maximum knee adduction during stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

It was a similar picture with maximum abduction, again there was a significant reduction in this maximum position when the symptomatic group was compared to the normal (see Fig.8.16.). Knee abduction tended to be greater than adduction and there was a little less variation between subjects, the orthoses did not introduce any changes within the data (see Table 8.4.).

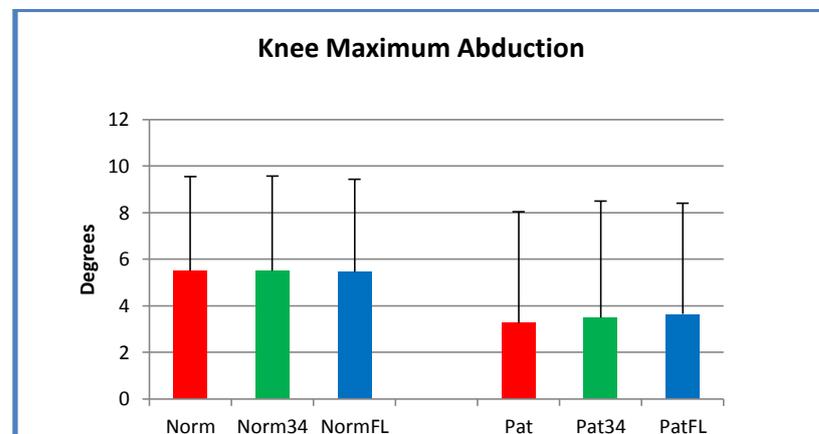


Fig.8.16. Bar chart showing maximum knee abduction during stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The normal group demonstrated greater mean abduction than the patients, but the patients exhibited greater mean adduction than the normal group; it is perhaps

unsurprising that the coronal ranges were similar (Fig. 8.17.). There was a hint in the confidence intervals that the patients tended to have a slightly larger range, but when all trials were amalgamated the mean difference was just over half of a degree (see Table 8.4.). There was no statistically significant change seen with the orthoses.

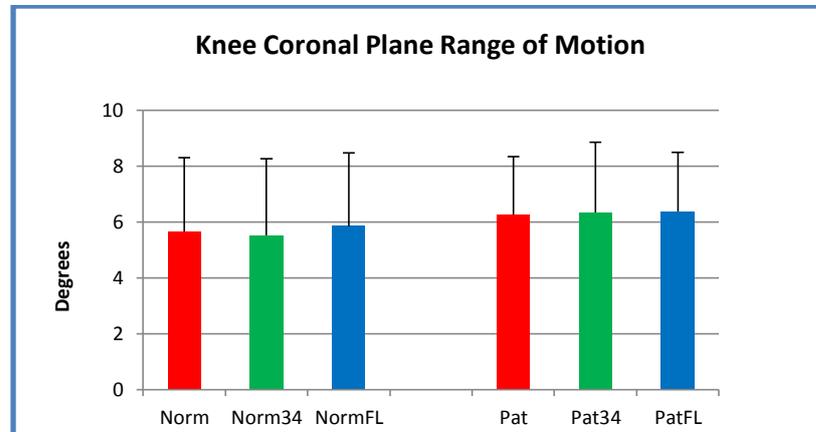


Fig.8.17. Bar chart showing coronal knee excursion during stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

During early stance, maximum external knee rotation was significantly less in the non-symptomatic group - the mean difference being over 4°. This is illustrated by Fig.8.18. which also shows the orthoses recorded no real difference when used in the footwear; this was confirmed by the P-values in Table 8.4.

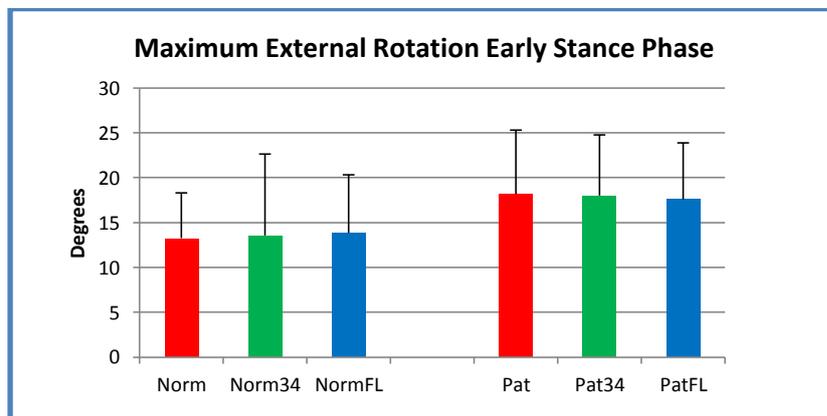


Fig.8.18. Bar chart showing maximum external knee position during early stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Knee rotation excursion was also significantly greater in the patient group; however the mean difference was reduced to 2.5° less than the maximum external position

results (see Table 8.4.). The mean differences were negligible when comparing the shoe only data with the orthoses and this lead to no significant changes being recorded with no trends seen within the data, this is illustrated in Fig.8.19.

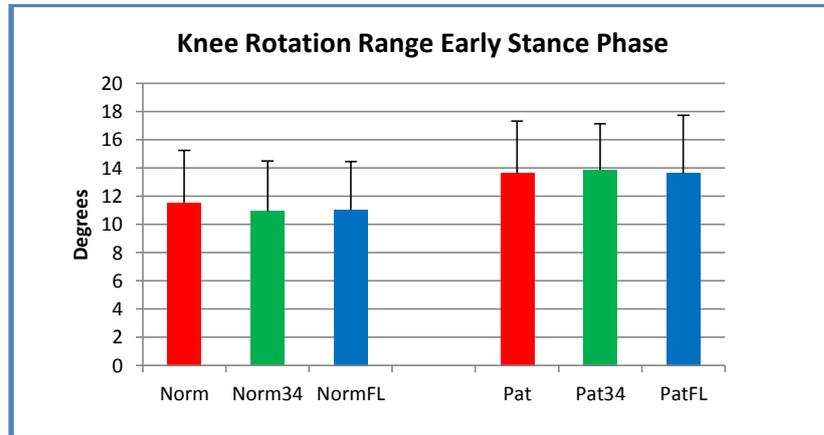


Fig.8.19. Bar chart showing transverse knee rotation during early stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Maximum external knee position was not significantly different between the groups or altered by the orthoses during late stance. The knee rotation excursion during late stance was also similar between the two groups. The orthoses did tend to reduce the range but this was only by a small amount and not enough to be significant (See Fig.8.20, and Table 8.4.)

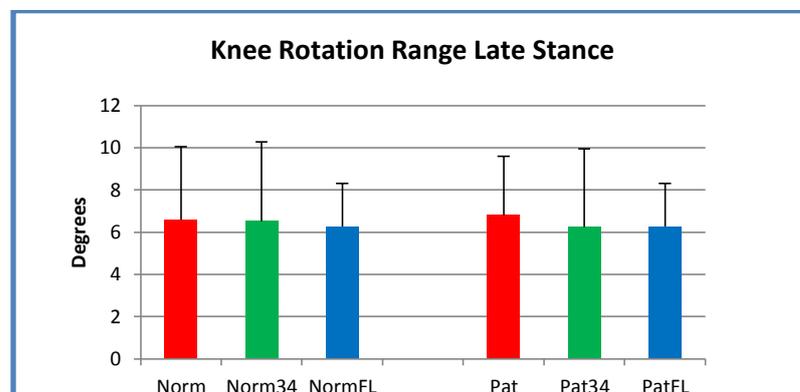


Fig.8.20. Bar chart showing transverse knee rotation during late stance phase, comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Table 8.4. Mean maximum positions and range of motion comparing the normal and patient group data and recording any differences seen with the orthoses.

Knee Range of Motion Walking	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Flexion at Heel Strike	1.082	P=0.227	-0.683 to 2.847	0.213	0.070	P=0.846	P=0.949
Flexion First Peak	1.514	P=0.115	-0.375 to 3.403	-0.543	-0.497	P=0.643	P=0.672
Trough	2.217	P=0.005	0.689 to 3.744	-0.162	-0.098	P=0.864	P=0.918
Maximum Flexion at Toe Off	1.142	P=0.241	-0.774 to 3.057	0.685	0.177	P=0.564	P=0.882
Sagittal ROM	0.059	P=0.963	-2.490 to 2.609	0.472	0.107	P=0.765	P=0.946
Maximum Adduction	2.677	P=0.000	1.281 to 4.072	-0.151	-0.019	P=0.861	P=0.983
Maximum Abduction	2.025	P=0.011	0.463 to 3.586	-0.115	-0.163	P=0.905	P=0.866
Coronal Plane ROM	-0.652	P=0.163	-1.573 to 0.268	0.036	-0.145	P=0.950	P=0.800
Max External Rotation Early Stance	-4.387	P=0.000	-6.593 to -2.163	-0.057	-0.034	P=0.967	P=0.980
Transverse ROM Early Stance	-2.543	P=0.000	-3.879 to -1.207	0.193	0.258	P=0.815	P=0.756
Max External Rotation Late Stance	1.199	P=0.239	-0.805 to 3.203	-0.313	-0.258	P=0.801	P=0.836
Transverse ROM Late Stance	0.025	P=0.962	-0.987 to 1.036	0.317	0.438	P=0.613	P=0.486

8.1.5. Knee moments - walking

The maximum flexion moment during early stance phase and the maximum extension moment during midstance demonstrated no statistically significant differences between the groups or with the orthoses compared to the shoe only condition. However the confidence intervals suggested that the patient group tended to be less than the normal group when looking at the results for maximum extension moment. The ¾ orthoses also demonstrated there was a reduction over the shoe only condition (see Table 8.5.) Fig.8.21. confirmed that both groups demonstrated a similar reaction to both orthoses.

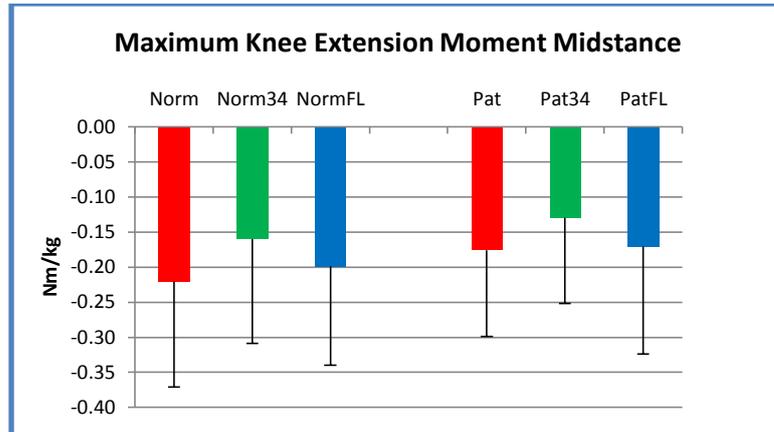


Fig.8.21. Bar chart showing maximum knee extension moment during midstance comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Maximum knee flexion moment during late stance was only small but it was significantly increased in the patient group over the normal group (see Table 8.5.). Neither orthoses produced any significant effect over the shoe only condition Fig.8.22. illustrated that there were consistent results between the shoe only and shoe plus orthoses conditions.

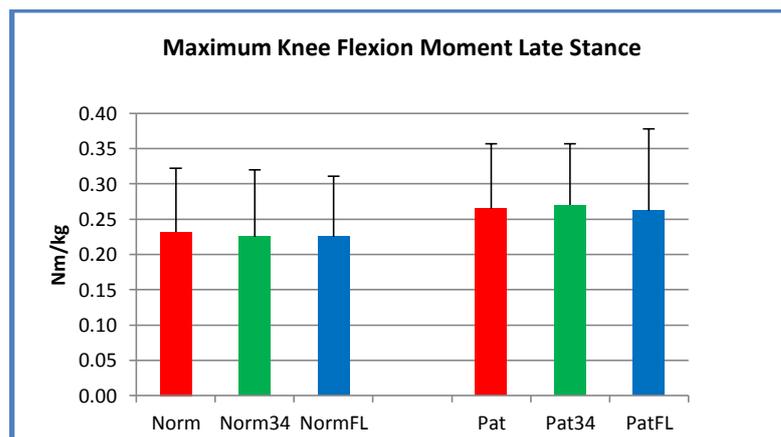


Fig.8.22. Bar chart showing maximum knee flexion moment during late stance comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Knee abduction moment during early stance was the smallest seen in the coronal plane, but it was significantly smaller in the patient group compared to the normals $p=0.035$ (see Table 8.5.). Neither orthoses reduced the mean moment significantly, but Fig.8.23. illustrated that both groups had a small reduction with the $\frac{3}{4}$ orthoses,

while the full length orthoses induced a small increase in the non-symptomatic group and a minimal reduction in the patient group.

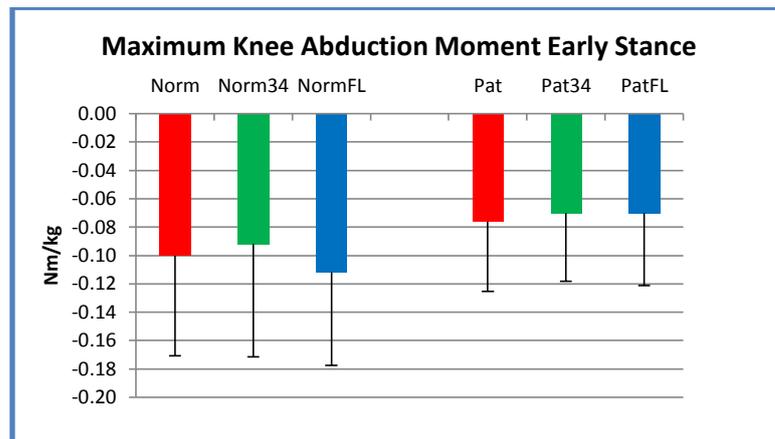


Fig.8.23. Bar chart showing maximum knee abduction moment during early stance comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The adduction moment was larger coronal plane moment in both groups; there was no statistically significant difference between the groups. However, Fig.8.24. would suggest that the patient cohort moment was a little less. It was obvious from the same bar chart that the orthoses had similar non-significant effects on both groups (see Table 8.5.).

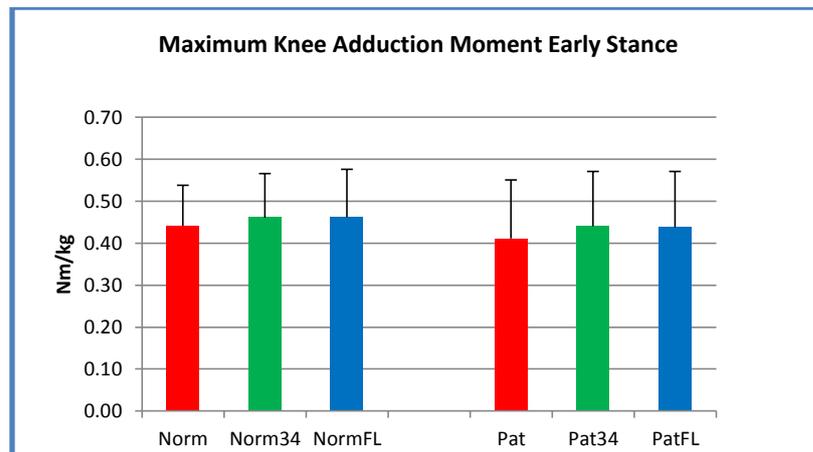


Fig.8.24. Bar chart showing maximum knee adduction moment during early stance comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The early stance coronal moment range was significantly reduced in the patient group over the normal group $P=0.012$ (see Table 8.5.). Again there were no significant differences induced by the orthoses over the shoe only condition,

Fig.8.25. shows both orthoses produced similar non-significant increase in both groups.

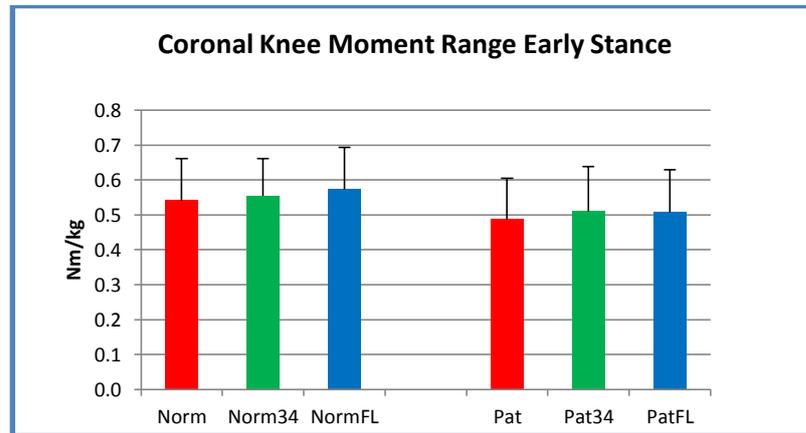


Fig.8.25. Bar chart showing coronal knee moment range during early stance comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Maximum abduction moment during late stance was again smaller than the adduction moment and about half the magnitude of early abduction moment. There was no significant difference between the two groups, but the confidence intervals did suggest that the patient group tended to have a minimally increased mean result. Fig.8.26. confirms there was no reaction by either group to the $\frac{3}{4}$ orthoses but both groups did show a reduction with the full length condition and this trend was reflected in the data (see Table 8.5).

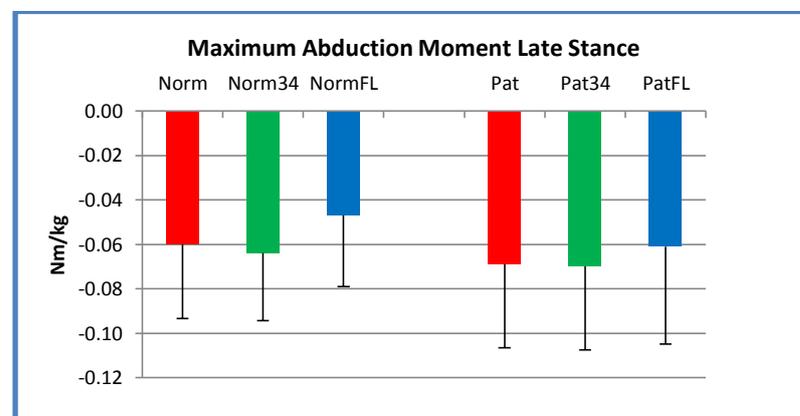


Fig.8.26. Bar chart showing maximum knee abduction moment during late stance comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

There was a larger mean difference between the pain free and symptomatic groups in the late stance phase and this trend was reflected in the confidence intervals and the

significance ($p=0.081$) see Table 8.5. Both orthoses increased this adduction moment though this was more noticeable with the full length condition. Fig.8.27. demonstrates that this increase was larger in the non-symptomatic group.

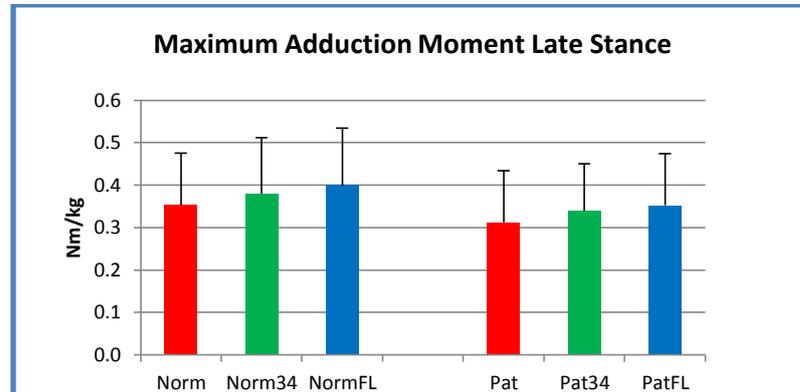


Fig.8.27. Bar chart showing maximum knee adduction moment during early stance comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

There were no significant changes in the late stance phase coronal range, it reflected the maximum adduction moment, and there were no trends to report in the data (see Table 8.5). The transverse moment remained internal all the way through the stance phase and was therefore reported as minimum and maximum internal knee moment. There were no significant differences seen between the patients and normal groups in maximum internal moment or the moment range, nor were there any trends to report between the groups or between the orthoses and shoe conditions (see Table 8.5 and Fig.8.28.).

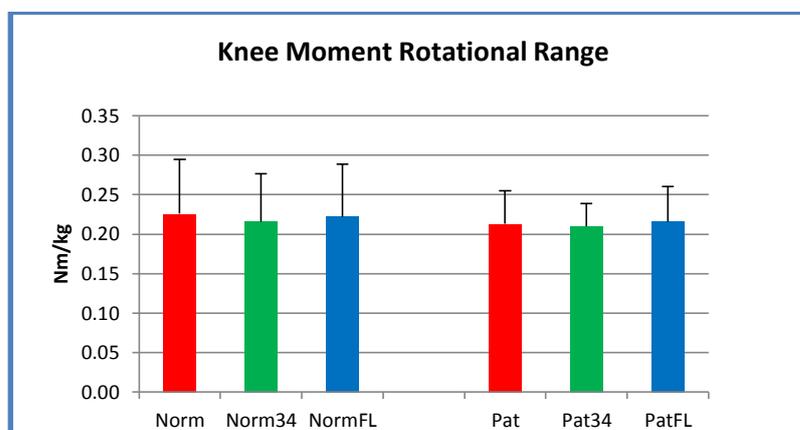


Fig.8.28. Bar chart showing rotational knee moment range for stance phase comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

Table 8.5. Mean maximum moments and range of moments of the knee comparing the normal and patient group data and recording any differences seen with the orthoses.

Knee Moments Walking	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Max Flexion M Early Stance	0.032	P=0.492	-0.060 to 0.125	0.002	0.008	P=0.975	P=0.883
Max Extension M	-0.037	P=0.158	-0.089 to 0.015	-0.053	-0.013	P=0.103	P=0.682
Max Flexion M Late Stance	-0.038	P=0.035	-0.073 to -0.003	0.002	0.005	P=0.912	P=0.822
Max Abduction M Early Stance	-0.28	P=0.022	-0.052 to -0.004	-0.006	0.003	P=0.695	P=0.839
Max Adduction M Early Stance	0.026	P=0.219	-0.016 to 0.068	-0.023	-0.024	P=0.374	P=0.367
Coronal M Range Early Stance	0.055	P=0.012	0.012 to 0.097	-0.017	-0.027	P=0.508	P=0.310
Max Abduction M Late Stance	0.009	P=0.189	-0.004 to 0.022	0.000	-0.010	P=0.994	P=0.193
Max Adduction M Late Stance	0.041	P=0.081	-0.005 to 0.087	-0.030	-0.043	P=0.296	P=0.138
Coronal M Range Late Stance	0.032	P=0.137	-0.010 to 0.075	0.030	0.032	P=0.259	P=0.227
Max External M	-0.004	P=0.714	-0.023 to 0.016	0.002	-0.004	P=0.881	P=0.730
Transverse M Range	0.007	P=0.503	-0.014 to 0.029	0.005	0.000	P=0.700	P=0.970

8.1.6. Ankle moments - walking

When walking the maximum ankle dorsiflexion moment of the patient group was significantly reduced compared to the non-symptomatic group (p=0.000) see Table 8.6. Fig.8.29. shows there was little increase with both orthoses in the patient group (maximum 0.03Nm/kg) but in the normal group there was an even small decrease observed, when combined there was no significant result seen.

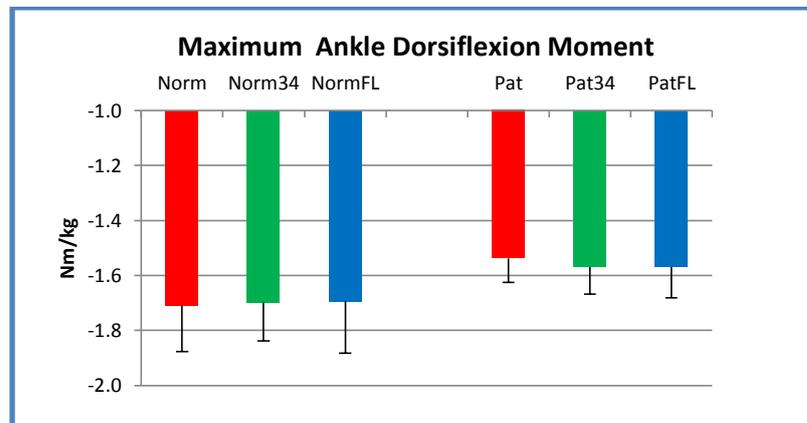


Fig.8.29. Bar chart showing maximum ankle dorsiflexion moment comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The mean maximum plantarflexion moment was less than the dorsiflexion moment and there was no difference seen between the groups (Table 8.6.). Fig.8.30. highlights that the ¾ orthoses tended to reduce the maximum moment and this was seen as a trend in the data (Table 8.6.) it did not produce a significant result.

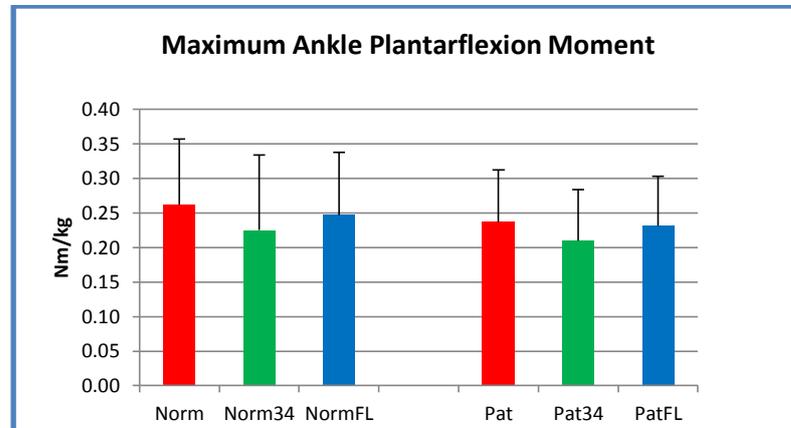


Fig. R8.30. Bar chart showing maximum ankle plantarflexion moment comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The difference between the two groups was small when considering the sagittal ankle moment range, however this was significant similar to the maximum dorsiflexion moment (Table 8.6.). Looking at the results of the orthoses compared to the shoe only condition there were no trends seen in the data. Fig.8.31. illustrates this well, but does highlight the difference between the groups.

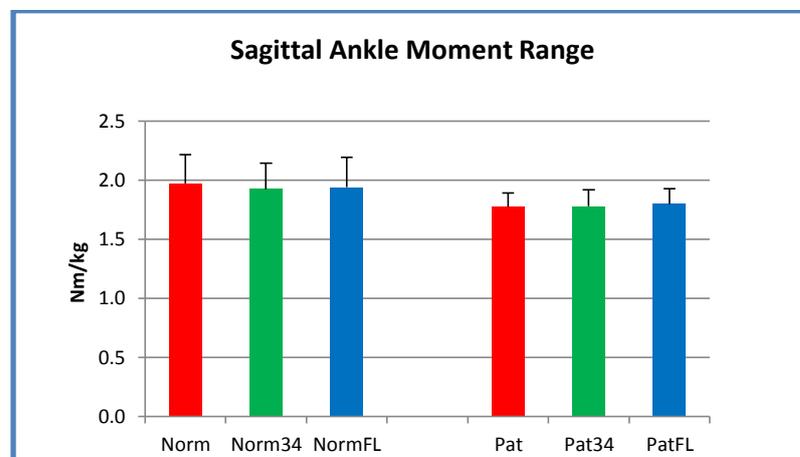


Fig.8.31. Bar chart showing sagittal ankle moment range during stance phase comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The maximum ankle inverting moment did not show any difference between the groups (see Table 8.6.) and though Fig. 8.32. would suggest both orthoses increased the moment, the differences were so small that they were not significant.

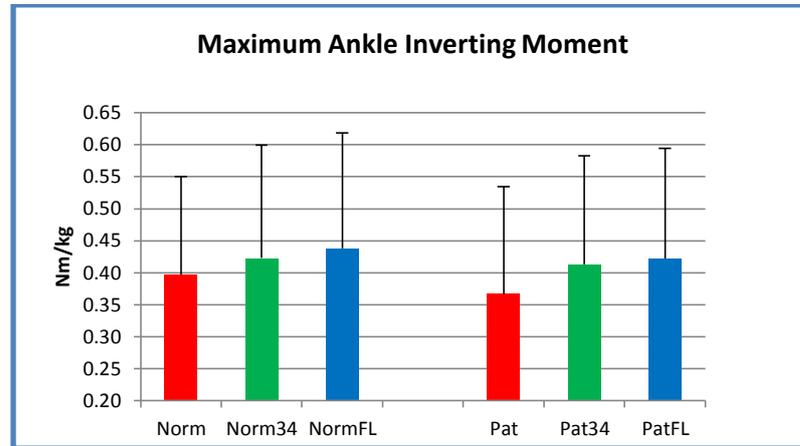


Fig.8.32. Bar chart showing maximum inverting ankle moment comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The evverting moment results displayed in Fig.8.33. were the opposite to the inverting bar chart, this time the difference between the groups was significant ($p=0.001$, Table 8.6.). There was also a trend in the data for the full length orthoses reducing the moment, the mean difference of 0.013 was not significant and Fig.8.33. suggested this was mainly from the non-symptomatic group.

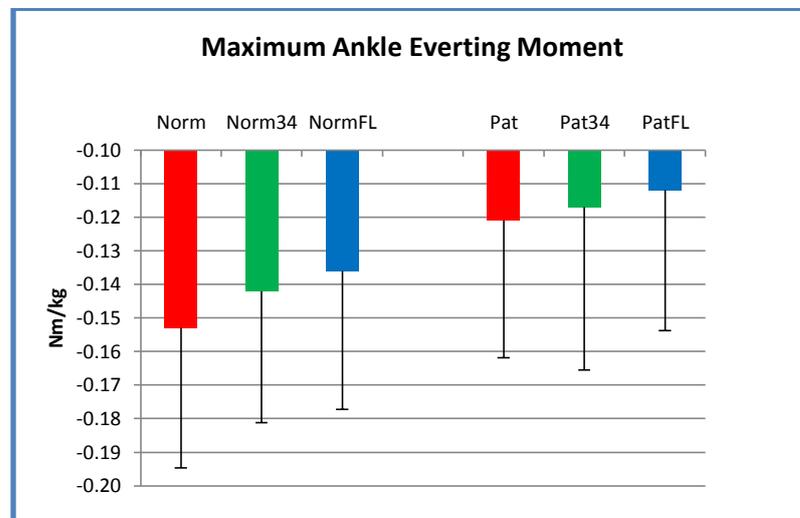


Fig.8.33. Bar chart showing maximum everting ankle moment comparing the patient and normal groups and how they reacted to the orthoses (Error bars = standard deviation)

The coronal moment range reflected the significant difference seen in the maximum everting moment and though there was a small trend in the confidence intervals the result was not significant. There were no differences recorded with either orthoses over the shoe only condition.

There were no significant results in the maximum mean moments or for the moment range in the transverse plane between the groups. The maximum internally rotating moment demonstrated a non-significant increase with the orthoses, but there was a non-significant decrease seen in the externally rotating moment; this resulted in the transverse moment range being virtually unaltered (Table 8.6.).

Table 8.6. Mean maximum moments and range of moments of the ankle comparing the normal and patient group data and recording any differences seen with the orthoses.

Ankle Moments Walking	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Max Dorsiflexion M	-0.142	P=0.000	-0.195 to -0.089	0.010	0.007	P=0.769	P=0.827
Max Plantarflexion M	0.019	P=0.260	-0.14 to 0.052	0.032	0.010	P=0.117	P=0.608
Sagittal M Range	0.161	P=0.000	0.085 to 0.237	0.022	0.003	P=0.635	P=0.946
Max Inverting M	0.018	P=0.555	-0.043 to 0.080	-0.036	-0.048	P=0.344	P=0.211
Max Everting M	-0.027	P=0.001	-0.042 to -0.012	-0.008	-0.013	P=0.412	P=0.160
Coronal M Range	0.045	P=0.178	-0.021 to 0.112	-0.028	-0.035	P=0.488	P=0.399
Max Internal M	-0.011	P=0.317	-0.034 to 0.011	-0.009	-0.016	P=0.495	P=0.241
Max External M	-0.005	P=0.729	-0.034 to 0.024	-0.020	-0.028	P=0.258	P=0.122
Transverse M Range	-0.006	P=0.683	-0.036 to 0.024	0.011	0.012	P=0.550	P=0.521

8.2. Results of stance limb symptomatic subjects - step descent

8.2.1. Calcaneal segment – step descent

There was no difference recorded between the symptomatic and non-symptomatic groups when considering the maximum dorsiflexion position. Both of the orthoses tended to increase the range over the shoe only condition (illustrated by Fig.8.34.) the mean difference was over 3.7° for each and this was not significant. The p-values were just outside the set level (p=0.060 and p=0.055 for the ¾ and full length orthoses respectively) see Table 8.7.

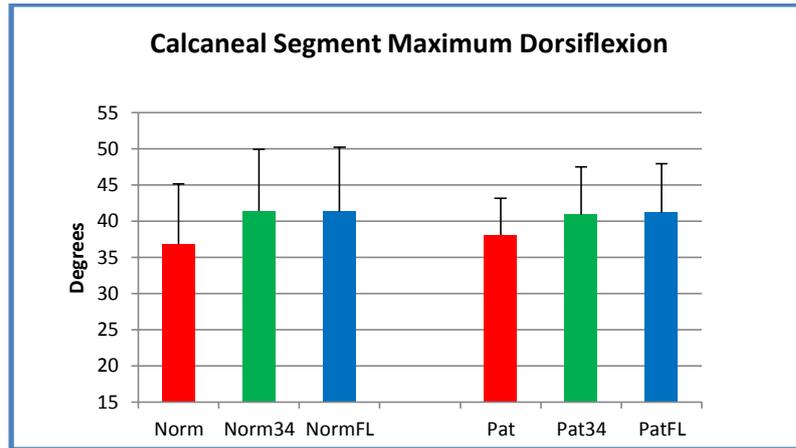


Fig.8.34. Bar chart demonstrating maximum calcaneal dorsiflexion with respect to the shank between the normal and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

When looking at the range of dorsiflexion between the two groups there was a slight trend for the patient group having a larger excursion (see Fig.8.35.) however it must be noted that the mean difference was less than 2° , see Table 8.7. Both orthoses did increase the range significantly over the shoe only condition; this was a little more marked with the $\frac{3}{4}$ orthoses (mean difference -3.741° and -3.395° for $\frac{3}{4}$ and full length orthoses respectively see Table 8.7.)

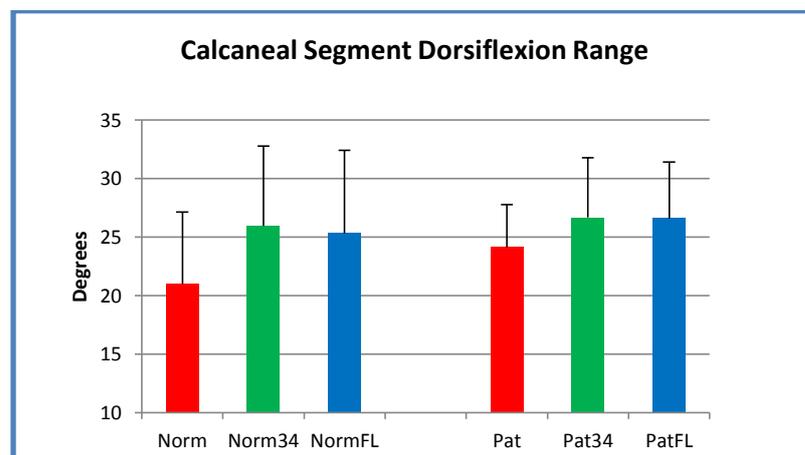


Fig.8.35. Bar chart demonstrating range of calcaneal dorsiflexion between the normal and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

During step descent coronal plane motion was similar to the sagittal plane in the fact that the segment remained in eversion and did not recover. For this study it was

decided to look at the forward continuum and lowering phases separately. The minimum everted position was significantly different between the groups; the mean difference was over 2° less in the patient group (Table 8.7.). Fig.8.36. illustrates that both orthoses reduced the degree of eversion in both groups and as such the results were significant (Table 8.7.).

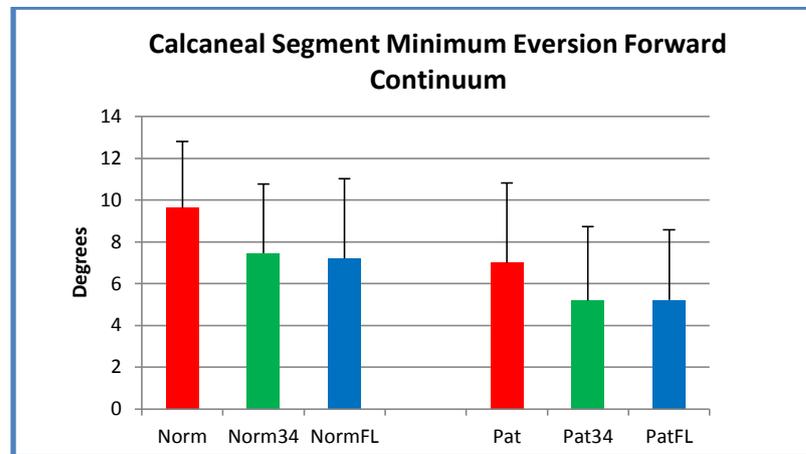


Fig.8.36. Bar chart illustrating the minimum everted position during the forward continuum phase between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

There was less mean difference between the two groups for maximum eversion (1.8°) but this was still significant (see Table 8.7.). Again both orthoses significantly reduced this position further which was highlighted by the bar chart in Fig.8.37. Similarly the mean differences were around 2° see Table 8.7.

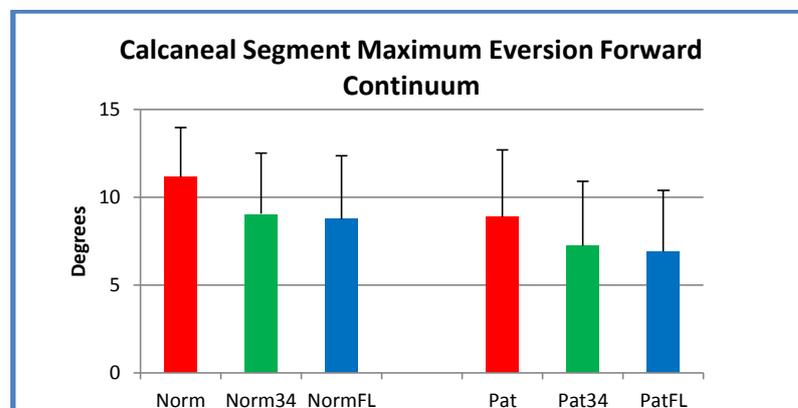


Fig.8.37. Bar chart illustrating the maximum everted position during the forward continuum phase between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

The mean coronal plane range for the forward continuum phase was not significantly different between the groups the mean difference being less than 0.4° , see Table 8.8. The orthoses did not produce any trends in the data.

Maximum eversion during the lowering phase was also significantly different between the groups, but again it must be recognised that the mean difference was only 1.72° (Table 8.7.). The bar chart in Fig.8.38. highlights the reductions made by the orthoses over the shoe only condition, however these did not reach the level of significance. The values posted in Table 8.7. would suggest there is a trend seen in both sets of data though this was more evident with the full length orthoses.

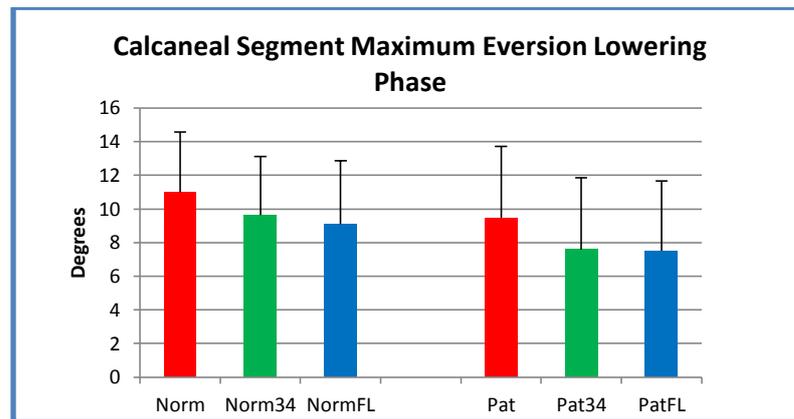


Fig.8.38. Bar chart illustrating the maximum everted position during the lowering phase between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

Though the maximum everted position was greater in the normal group during the lowering phase, it was the patient group who posted a significantly greater range, this is illustrated in Fig.8.39. It also shows there was very little difference seen with the orthoses especially in the patient group and this lead to there being no significant results or any trends seen within the data.

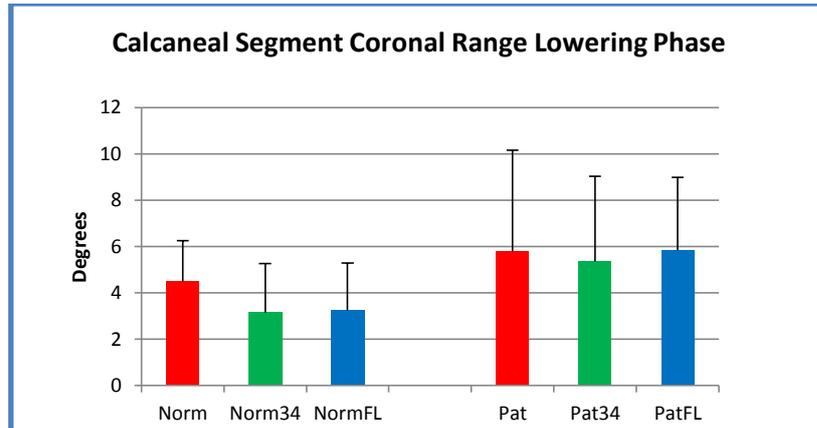


Fig.8.39. Bar chart illustrating the range of eversion during the lowering phase between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

The maximum transverse rotational positions of the calcaneal segment on the shank did not show any significant differences between groups of subjects or with any of the orthoses, therefore only the ranges for forward continuum and lowering phases were reported.

There were no significant differences found between the subject groups in the transverse plane range during the forward continuum phase (see Table 8.7.). Fig.8.40. demonstrates that the $\frac{3}{4}$ orthoses produced an increase in both groups, which was noticeable in the data, but nowhere near significant. The full length orthoses produced varying results with a small decrease in the normal group and a small increase in the patient group leading to no trend being observable.

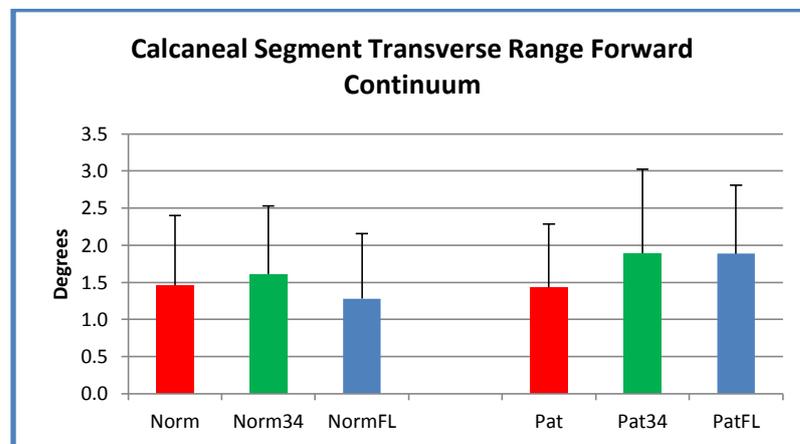


Fig.8.40. Bar chart illustrating the range of rotation of the calcaneal segment on the shank during the forward continuum phase between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

During the lowering phase the patient group did demonstrate a significantly increased rotational range compared to the normal subjects (see Table 8.7.). Both orthoses did tend to reduce the range in the lowering phase over the shoe only condition. The ¾ orthoses did reduce the range more than the full length orthoses (see Fig.8.41) but the former did not produce a significant result.

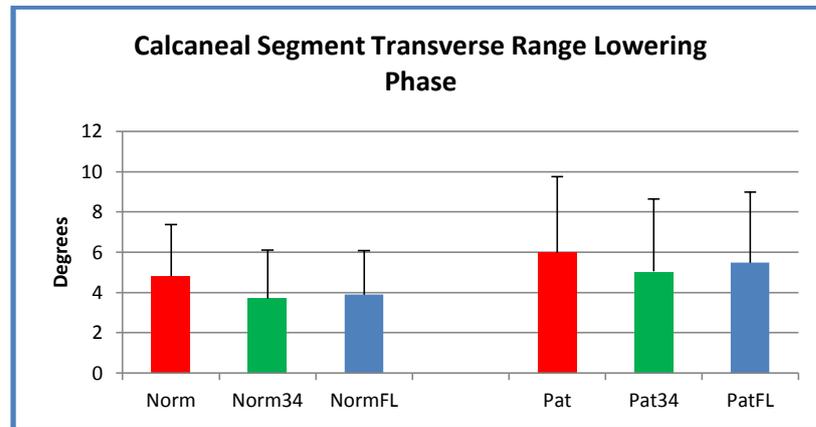


Fig.8.41. Bar chart illustrating the range rotation of the calcaneal segment on the shank during the lowering phase between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

Table 8.7. Maximum positions and mean ranges of motion of the calcaneal segment with respect to the shank comparing the normal and patient group data and recording any differences seen with the orthoses.

Calcaneal range of motion with respect to the shank Step	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				¾	FL	¾	FL
Dorsiflexion Max	-0.190	P=0.904	-3.332 to 2.951	-3.760	-3.799	P=0.060	P=0.055
Dorsiflexion ROM	-1.722	P=0.159	-4.132 to 0.688	-3.741	-3.395	P=0.013	P=0.025
Minimum Eversion F.C.	2.304	P=0.003	0.821 to 3.788	1.991	2.102	P=0.031	P=0.024
Maximum Eversion F.C.	1.796	P=0.009	0.507 to 3.446	1.874	2.183	P=0.040	P=0.018
Coronal Range F.C.	-0.382	P=0.150	-0.777 to 0.121	-0.116	0.081	P=0.674	P=0.770
Maximum Eversion L.P.	1.722	P=0.042	0.065 to 3.378	1.617	1.927	P=0.114	P=0.063
Coronal Range L.P.	-2.031	P=0.002	-3.306 to 0.757	0.874	0.593	P=0.266	P=0.454
Transverse Range F.C.	-0.285	P=0.160	-0.684 to 0.114	-0.307	-0.140	P=0.213	P=0.571
Transverse Range L.P.	-1.357	P=0.041	-2.656 to -0.058	1.024	0.711	P=0.266	P=0.378

8.2.2. Metatarsal segment – step descent

The range of motion in the sagittal plane was relatively small and tended to move from a plantarflexed position to a less plantarflexed position through step descent. Fig.8.42. shows the patients did demonstrate a little larger mean but this was not significant when compared to the normal group. The range of motion in the normal group remained almost constant with the orthoses while in the patient group there was a reduction with the $\frac{3}{4}$ orthoses though this was less than one degree, and when the results were combined no significance or trend was seen in the data.

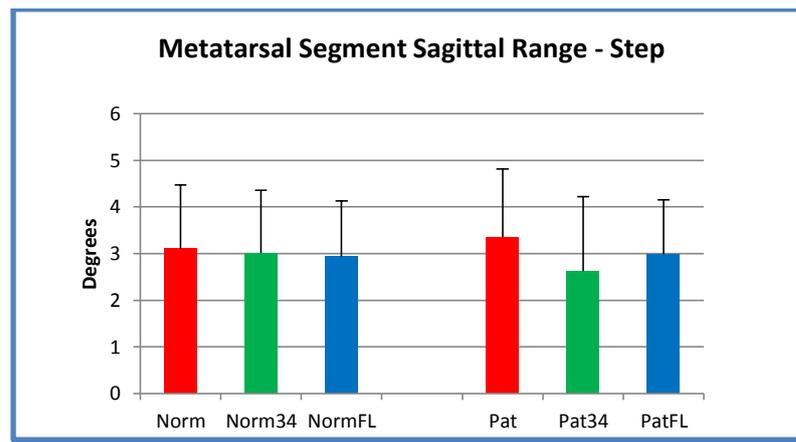


Fig.8.42. Bar chart illustrating the range of sagittal plane motion of the metatarsal segment on the calcaneal segment during step descent between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

During the forward continuum phase the coronal range of metatarsal segment motion on the calcaneal segment was less than 1° . Fig.8.43. does illustrate that the patient group demonstrated a slightly wider range but this was nowhere near significant. It was the patient group who reacted more predictably to the orthoses with the $\frac{3}{4}$ orthoses reducing the mean range by 0.2° while the full length reduced the range further. The normal group also showed a minor reduction with the $\frac{3}{4}$ orthoses, but the full length orthoses actually increased the range slightly. There were no significant results or any real trends seen within the data (Table 8.8.).

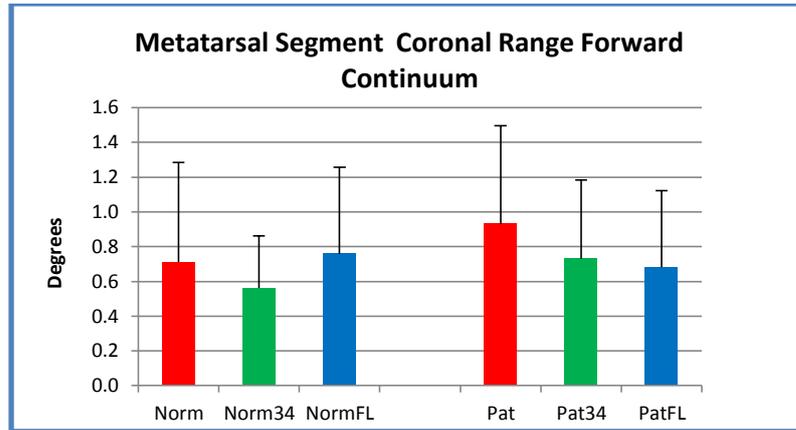


Fig.8.43. Bar chart illustrating the range eversion of the metatarsal segment on the calcaneal segment during the forward continuum phase of step descent between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

During the lowering phase Table 8.8. shows that there was no difference between the symptomatic and non-symptomatic groups, however this time both orthoses do reduce the range of motion significantly. This is illustrated in Fig. 8.44 and though the overall ranges are still small the orthoses manage to reduce the range by one third.

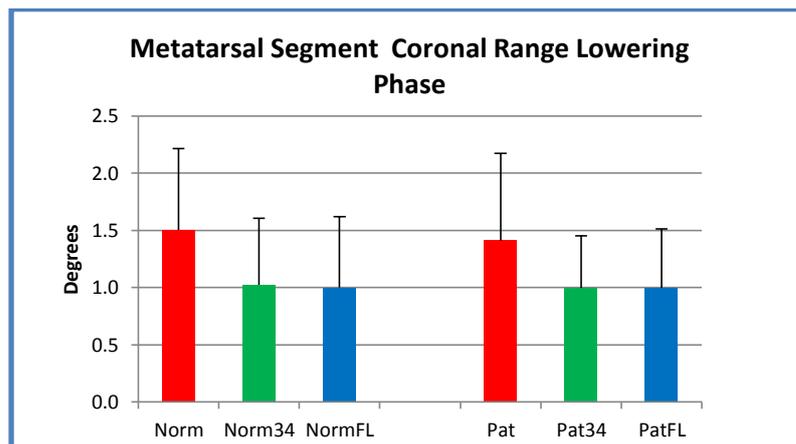


Fig.8.44. Bar chart illustrating the range eversion of the metatarsal segment on the calcaneal segment during the lowering phase of step descent between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

The range of rotation of the metatarsal segment on the calcaneal segment was small during the lowering phase and less during the forward continuum. Fig.8.45. illustrates that there was no difference between the groups and this was confirmed in Table 8.8. The orthoses did tend to reduce the range in the patient group but only by minute amounts. The non-symptomatic group did not demonstrate any differences

with the two orthoses and as such there were no significant results or any trends seen.

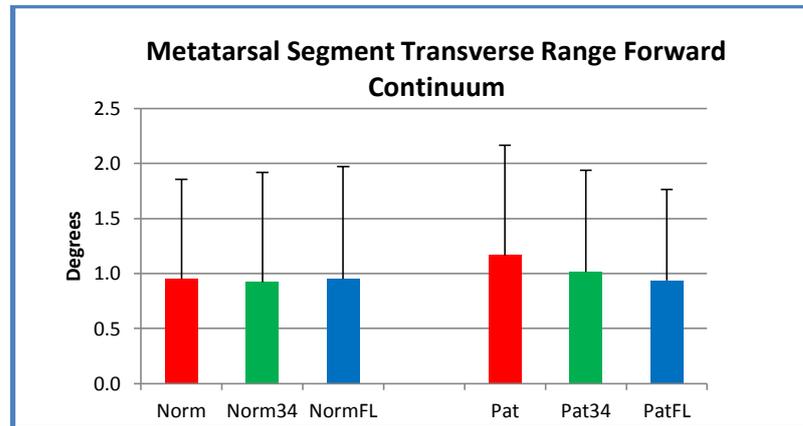


Fig.8.45. Bar chart illustrating the transverse plane range of the metatarsal segment on the calcaneal segment during the lowering phase of step descent between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

During the lowering phase the range of rotation of the metatarsal segment was a little larger but only by half a degree; there was no significant mean difference between the groups (Table 8.8.). This time the full length orthoses made very little difference to the range (Fig.8.46.). The $\frac{3}{4}$ orthoses tended to reduce the rotation by about half a degree in both groups demonstrating a trend in the data $p=0.061$ (Table 8.8.).

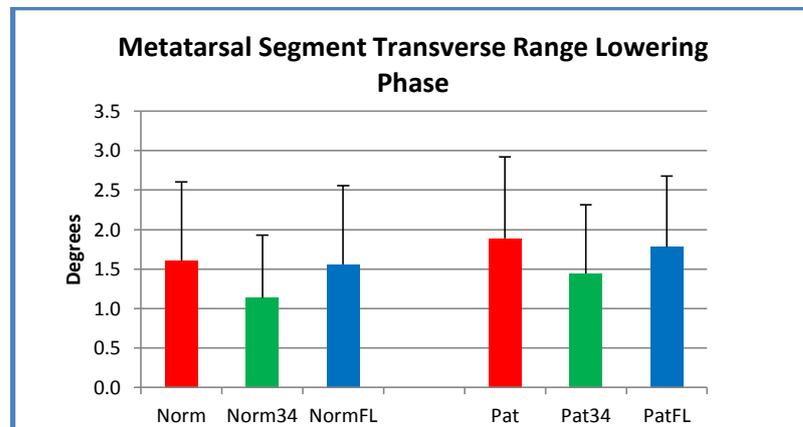


Fig. R2.46. Bar chart illustrating the transverse plane range of the metatarsal segment on the calcaneal segment during the lowering phase of step descent between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

Table 8.8. Maximum positions and mean ranges of motion of the metatarsal segment with respect to the calcaneal segment comparing the normal and patient group data and recording any differences seen with the orthoses.

Metatarsal range of motion with respect to the Calcaneal Segment Step	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Sagittal ROM	0.032	P=0.913	-0.545 to 0.608	0.414	0.267	P=0.243	P=0.454
Coronal ROM F.C.	-0.104	P=0.310	-0.307 to 0.099	0.175	0.101	P=0.163	P=0.423
Coronal ROM L.P.	0.041	P=0.757	-0.220 to 0.301	0.466	0.457	P=0.006	P=0.006
Transverse ROM F.C.	-0.94	P=0.640	-0.493 to 0.305	0.091	0.116	P=0.701	P=0.639
Transverse ROM L.P.	-0.270	P=0.177	-0.633 to 0.124	0.457	0.075	P=0.061	P=0.759

8.2.3. Phalangeal segment – step descent

When looking at maximum toe segment dorsiflexion the normal group did record an increased range compared to the symptomatic group though the mean difference was very small compared to the overall position. As such no significant difference was noted. The orthoses produced reductions in both groups though again this was not significant (Fig.8.47) the ¾ orthoses produced the greatest reduction in the normal group while it was the full length orthoses that produced the greatest reduction in the patient group.

The total range of motion of the phalangeal segment on the metatarsal segment was not significant between the groups- the mean difference being just over 1° (Table 8.9.).The orthoses however did suggest there was a definite trend with both orthoses reducing the range (p=0.054 and p=0.051 ¾ and full length respectively). The confidence intervals suggested there was a reduction of up to 7° though at the other end of the spectrum there was a minimal increase in range of less than a degree.

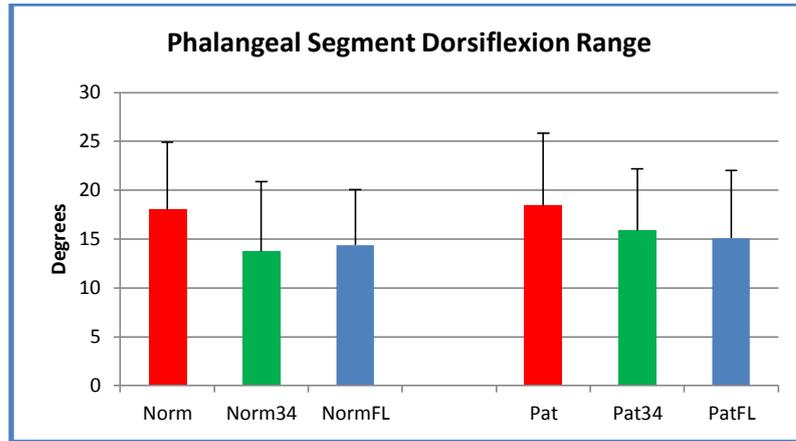


Fig.8.47. Range of phalangeal segment dorsiflexion with respect to the metatarsal segment during step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

Table 8.9. Maximum positions and mean ranges of motion of the phalangeal segment with respect to the metatarsal segment comparing the normal and patient group data and recording any differences seen with the orthoses

Phalangeal range of motion with respect to the Metatarsal Segment Step	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Maximum Dorsiflexion	0.750	P=0.668	-2.719 to 4.218	2.255	2.452	P=0.291	P=0.256
Sagittal Plane ROM	-1.079	P=0.454	-3.931 to 1.772	3.418	3.484	P=0.054	P=0.051

8.2.4. Knee kinematics – step descent

The mean difference in maximum knee flexion between the two groups was -4.8° which was significant. The confidence intervals suggested that the maximum difference was over 8° this is well illustrated by Fig.8.48. where there is an even spread of data within each group. Table 8.10. clearly demonstrated that the orthoses did not produce any significant results at the knee in any plane.

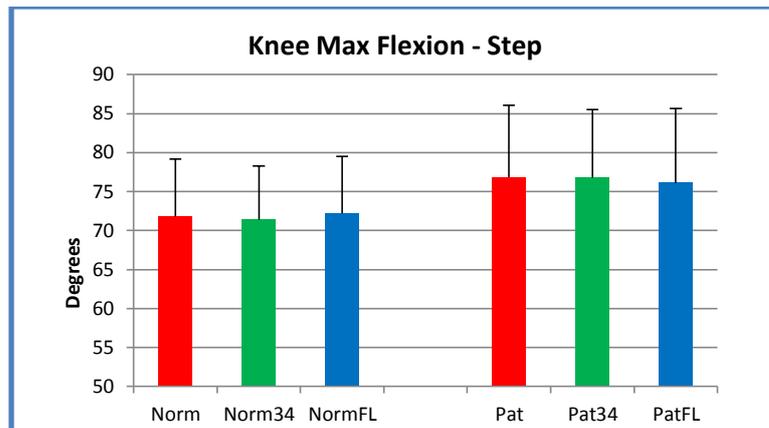


Fig.8.48. Maximum knee flexion during step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

When looking at the mean range of the sagittal motion for the two groups the difference was more marked than the maximum position. Knee excursion in the sagittal plane tended to be greater in the patient group (Fig.8.49.). The mean difference was -6.5° and significant (Table 8.10.). The confidence intervals suggesting the maximum difference seen was a little under 10° .

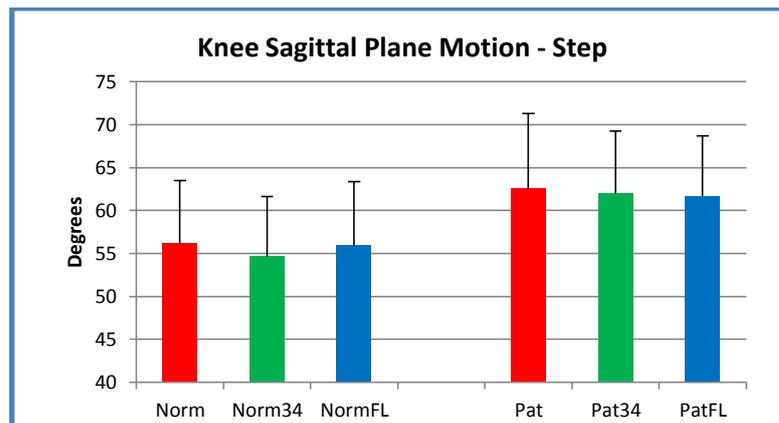


Fig.8.49. Range of knee flexion during step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

In the coronal plane maximum adduction was significantly larger in the patient group, but it was the normal group which demonstrated the greatest maximum abduction position. When the mean coronal plane range was calculated the patient group exhibited a significantly larger range (Fig.8.50.) this was mainly due to the bigger difference seen in the adduction component which is illustrated by the confidence intervals in Table 8.10. (adduction = 1.851 - 4.697, abduction = 0.320 - 3.227).

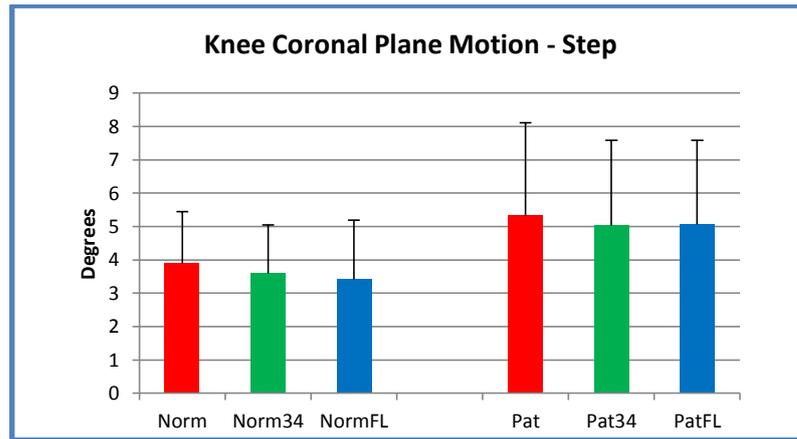


Fig.8.50. Range of coronal knee excursion during step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

The range of transverse knee excursion during the forward continuum phase was relatively small in both groups. However the patients demonstrated a significantly increased range that was almost an extra 30% (Fig.8.51.) the confidence intervals show that this was 1.5° maximum. The normal group displayed almost double the range seen in the lowering phase compared to the forward continuum phase, but the patient group mean was virtually the same. There was no significant difference between the groups which was confirmed by the results in Table 8.10.

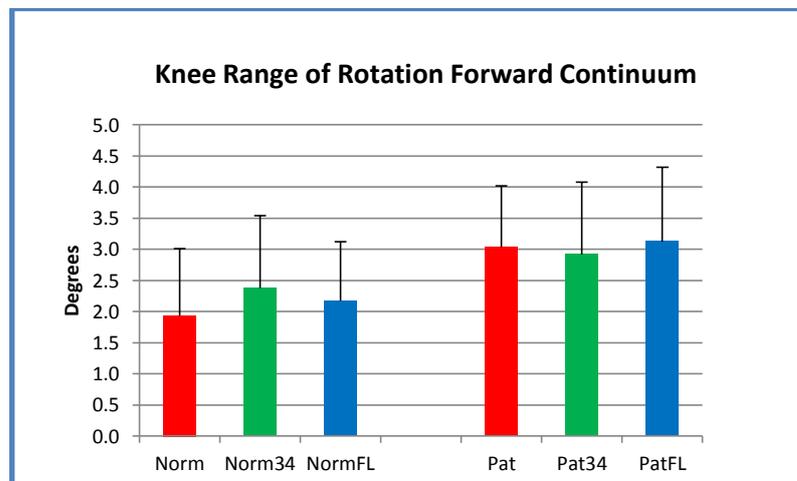


Fig.8.51. Range of transverse knee excursion during the forward continuum phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

Table 8.10. Maximum positions and mean ranges of motion of the knee comparing the normal and patient group data and recording any differences seen with the orthoses

Knee Range of Motion Step	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Maximum Flexion	-4.771	P=0.008	-8.252 to -1.289	0.237	0.164	P=0.912	P=0.939
Flexion Range	-6.496	P=0.000	-9.650 to -3.342	1.049	0.563	P=0.588	P=0.773
Maximum Adduction	3.274	P=0.000	1.851 to 4.697	-0.064	-0.623	P=0.942	P=0.480
Maximum Abduction	1.773	P=0.017	0.320 to 3.227	-0.234	-0.263	P=0.793	P=0.771
Coronal Plane ROM	-1.501	P=0.002	-2.148 to -0.583	0.298	0.361	P=0.598	P=0.526
Transverse Range F.C.	-0.865	P=0.000	-1.326 to -0.404	-0.167	-0.166	P=0.557	P=0.562
Transverse Range L.P.	0.034	P=0.940	-0.861 to 0.928	0.022	-0.159	P=0.968	P=0.774

8.2.5. Knee moments – step descent

When examining the maximum flexion moment during the forward continuum phase of the step cycle there were no differences seen between either of the groups or between the shoe only condition and either of the orthoses. Later in the step cycle maximum flexion moment was significantly larger in the patient group, this is illustrated in Fig.8.52. This chart also seems to show the ¾ orthoses increases the moment in both groups. This trend can be seen in the data in Table 8.11, however it was a long way off producing a significant result.

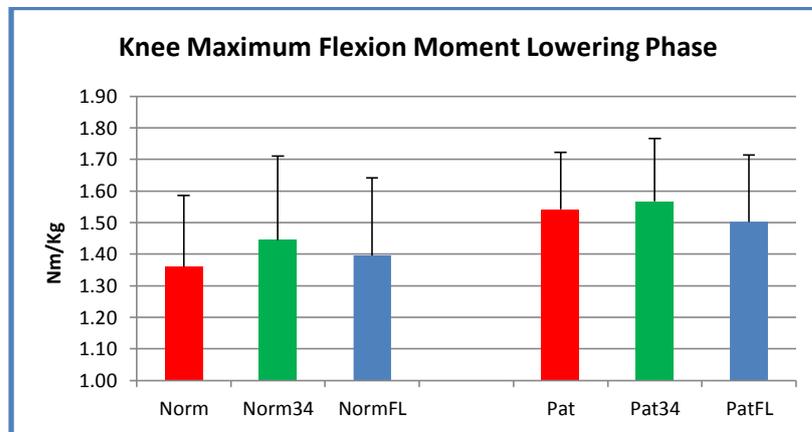


Fig.8.52. Maximum knee flexion moment during the lowering phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

All subjects in both groups remained in a flexion moment therefore the flexion range was smaller compared to the maximum moment. The moment excursion did reflect

the maximum and the patients still exhibited a significantly increased range compared to the normal group. This is illustrated in Fig.8.53. and confirmed in Table 8.11. where the confidence intervals reveal the maximum mean difference to be 0.228Nm/kg. There were no differences seen with the orthoses over the shoe only condition.

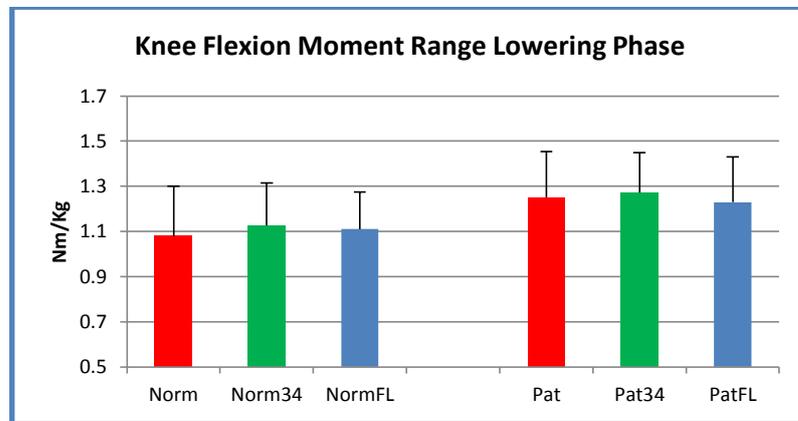


Fig.8.53. Knee flexion moment range during the lowering phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

The non-symptomatic group demonstrated a significantly reduced maximum adduction moment compared to the patient group during the forward continuum phase, this is presented in Fig.8.54. The mean difference was small and hence the significant result may suggest that this diversity was seen in most of the subjects. There were no differences recorded with the orthoses over the shoe only condition.

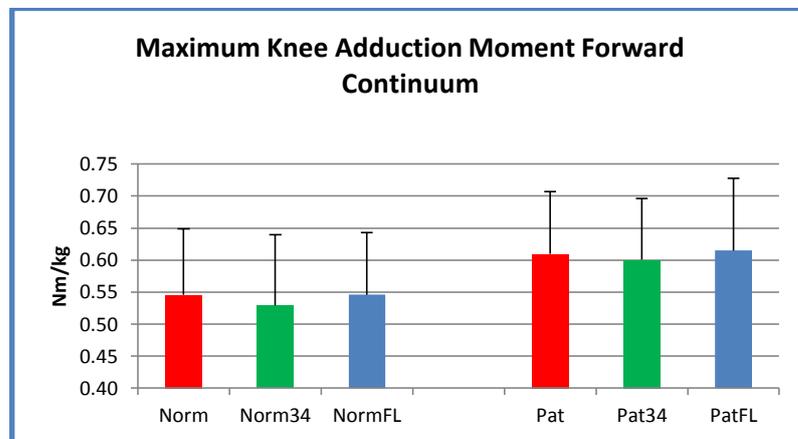


Fig.8.54. Maximum knee adduction moment during forward continuum phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

The significant mean difference between the groups was continued to the coronal moment range $p=0.001$ (see Table 8.11.). Fig.8.55. illustrates that the orthoses reduced the moment excursion in both groups and this time there was a significant reduction seen $p=0.019$ and $p=0.028$ for the $\frac{3}{4}$ and full length orthoses respectively (see Table 8.11.).

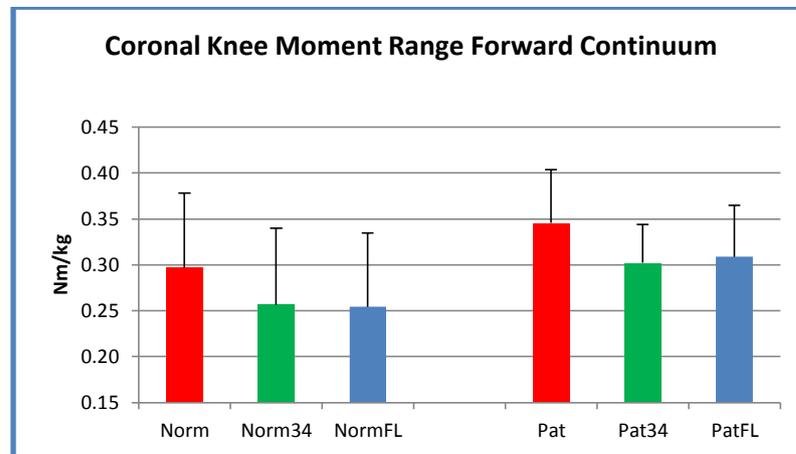


Fig.8.55. Coronal knee moment range during forward continuum phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

The maximum adduction moment during the lowering phase was almost half of what was recorded for the forward continuum phase, there was no difference recorded between the groups. When looking at the moment range during the later phase the symptomatic subjects did demonstrate a significantly increased range of moment over the pain free group $p=0.006$ (Table 8.11.). The orthoses made no significant changes to the maximum moment or the range of the moment.

In the transverse plane the maximum internal moment during the forward continuum phase was not significantly different between the groups. The full length orthoses tended to lower the P value but not enough to produce a significant result, there were no discernible trends displayed with the $\frac{3}{4}$ orthoses. The transverse moment range in this phase was significantly larger in the patient group, which is illustrated in Fig.8.56. It also shows the orthoses did tend to reduce the moment in both groups however this was not enough to be significant (see Table 8.11.).

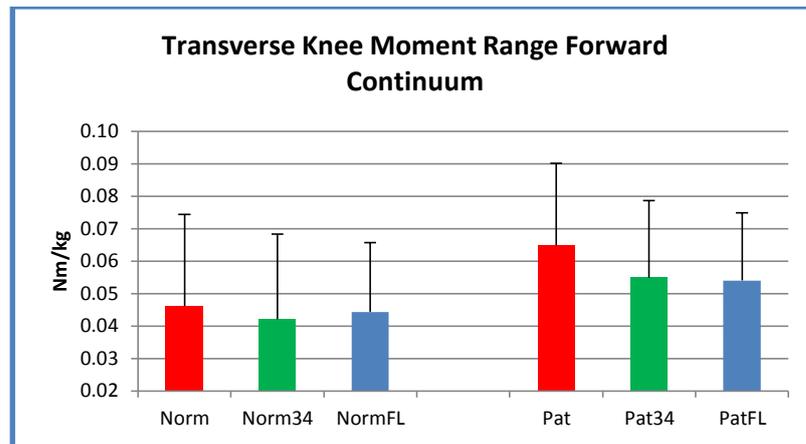


Fig.8.56. Transverse knee moment range during forward continuum phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

During the lowering phase the mean maximum internal knee moment and the moment range were not significantly different between the groups. Although the confidence intervals for the maximum moment did suggest there was a very mild trend towards the normal group having a larger value. This small mean increase (0.019Nm/kg.) is illustrated in Fig.8.57. which also shows the full length orthoses increased the maximum moment in both groups this was reflected in the data but was not a significant difference (Table 8.11.). This small increase did not have any influence on the rotational moment range and hence there were no significant changes noted.

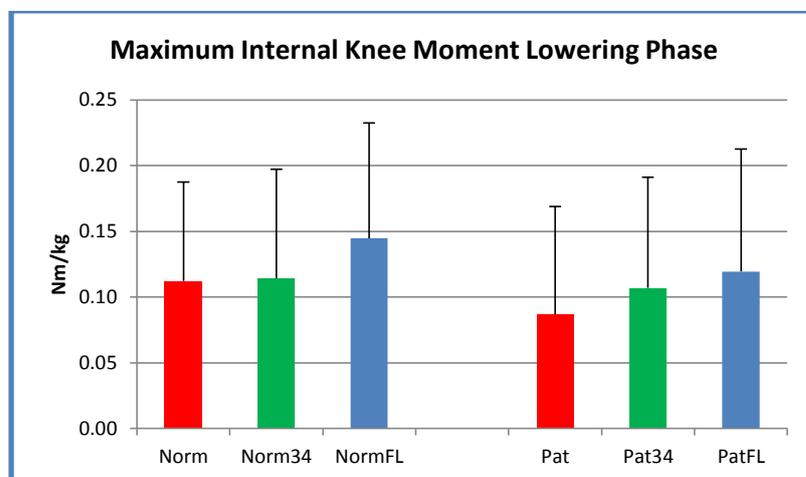


Fig.8.57. Maximum internal knee moment during lowering phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

Table 8.11. Maximum moments and mean ranges of moments of the knee comparing the normal and patient group data and recording any differences seen with the orthoses

Knee Moments Step	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Max Flexion M. F.C.	0.035	P=0.470	-0.61 to 0.131	0.003	0.021	P=0.963	P=0.720
Flexion M Range F.C.	0.008	P=0.741	-0.039 to 0.055	0.034	0.30	P=0.243	P=0.313
Max Flexion M. L.P.	-0.137	P=0.005	-0.231 -0.043	-0.055	0.002	P=0.343	P=0.975
Flexion M Range L.P.	-0.146	P=0.001	-0.228 to -0.064	-0.034	-0.005	P=0.493	P=0.927
Max Adduction M F.C.	-0.068	P=0.002	-0.112 to -0.025	0.013	-0.003	P=0.631	P=0.916
Coronal M Range F.C.	-0.050	P=0.001	-0.078 to -0.021	0.042	0.040	P=0.019	P=0.028
Max Adduction M L.P.	-0.011	P=0.608	-0.052 to 0.031	-0.032	-0.040	P=0.368	P=0.116
Coronal M Range L.P.	-0.044	P=0.006	-0.075 to -0.013	0.002	0.005	P=0.911	P=0.775
Max internal M. F.C.	0.002	P=0.818	-0.016 to 0.012	-0.005	-0.011	P=0.529	P=0.190
Transverse M Range F.C.	-0.014	P=0.009	-0.024 to -0.003	0.007	0.006	P=0.272	P=0.328
Max internal M. L.P.	0.019	P=0.284	-0.016 to 0.055	-0.011	-0.033	P=0.617	P=0.142
Transverse M Range L.P.	-0.002	P=0.787	-0.020 to 0.015	0.002	-0.008	P=0.837	P=0.441

8.2.6. Ankle moments – step descent

The mean difference of the maximum ankle dorsiflexion moment was very small between the groups (-0.056Nm/kg) but this was a significant difference (Table 8.12.). Fig.8.58. illustrates that this was mainly due to the larger increases seen with the orthoses in the normal group. The increases seen in both groups were also significant. However when the dorsiflexion moment range was considered there were no differences between the groups or any significant changes with the orthoses over the shoe only condition. There were small increases in the normal group which were not repeated in the symptomatic cohort.

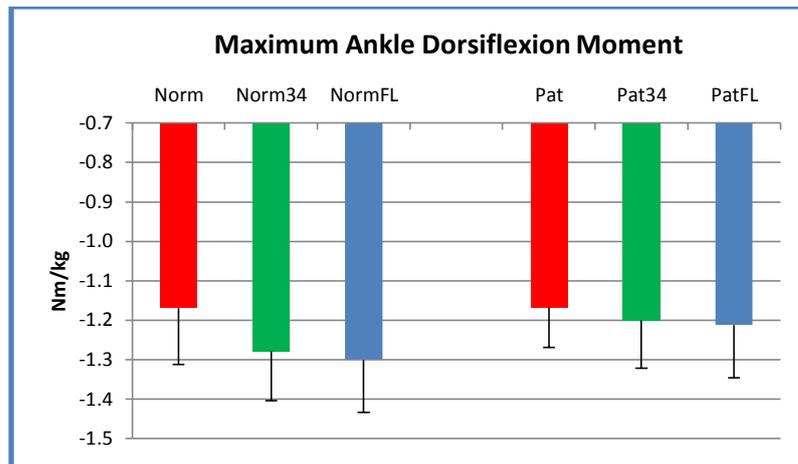


Fig.8.58. Maximum ankle dorsiflexion moment during step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

In the coronal plane the mean maximum inverting moment during the forwards continuum phase was significantly larger in the patient group compared to the non-symptomatic group (mean difference -0.160Nm/kg) however neither orthoses altered the moment (Table 8.12.). The mean difference when looking at the coronal moment range for the forward continuum phase remained similar (mean difference -0.133Nm/kg) and unsurprisingly significant. Fig.8.59. shows the orthoses reduced the range of moment in all cases, but only by a very small amount and as such did not produce any significant results; however there was a mild trend towards this reduction reflected in the data.

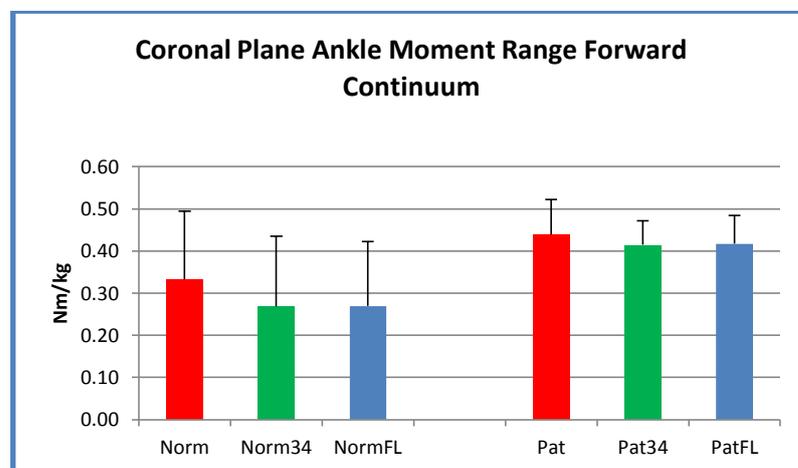


Fig.8.59. Coronal ankle moment range during forward continuum phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

During the lowering phase the maximum inverting moment was not significantly different between the groups and there were no changes produced by either of the orthoses. It was similar when looking at the coronal moment range, however the patient group did demonstrate a trend towards being increased; this was seen by the shift in the confidence intervals (Table 8.12.) and is illustrated in Fig.8.60.

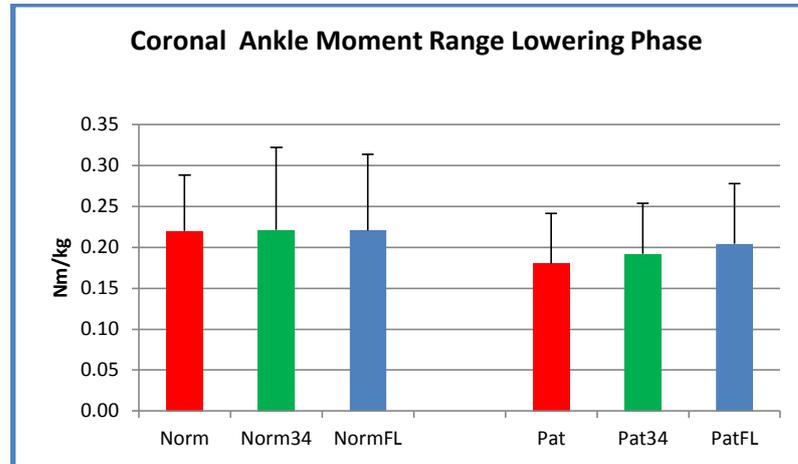


Fig.8.60. Ankle coronal moment range during lowering phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

The maximum external ankle moment during the forward continuum phase had a mean difference of -0.014Nm/kg which was not significant. Fig.8.61. illustrates that both orthoses did reduce the maximum moment in each group and this was reflected in the mean difference significance value. This reduction was not seen in the early phase transverse moment range where again the groups produced similar means. The orthoses did produce a very small reduction in the normals group, but made absolutely no difference in the symptomatic group (see Table 8.12.).

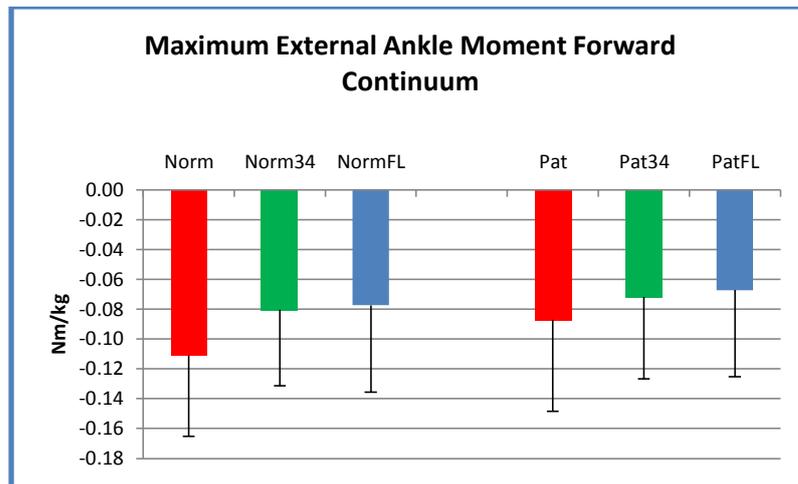


Fig.8.61. Maximum external ankle moment during forward continuum phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

When looking at the maximum external ankle moment during the lowering phase there was no significant differences between the groups but there was a definite trend in the data with the P value being just outside the level set for significance, and this was confirmed by the confidence intervals (see Table 8.12.). Similar to the forward continuum moment the orthoses did not produce a significant difference, but Fig.8.62. illustrates that there was a small reduction in each group with both orthoses over the shoe only condition.

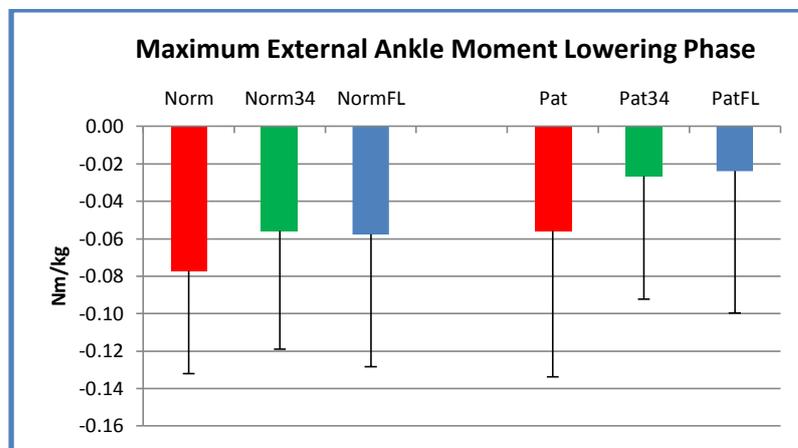


Fig.8.62. Maximum external ankle moment during the lowering phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

When looking at the transverse range of moment in the lowering phase there was a significant difference between the groups, the patient group demonstrating a larger range than the normal group. ($p=0.033$ see Table 8.12.). This time the $\frac{3}{4}$ orthoses did produce a significant decrease, Fig.8.63. illustrates that the main reduction was seen in the non-symptomatic group, the full length orthoses did not produce a significant result.

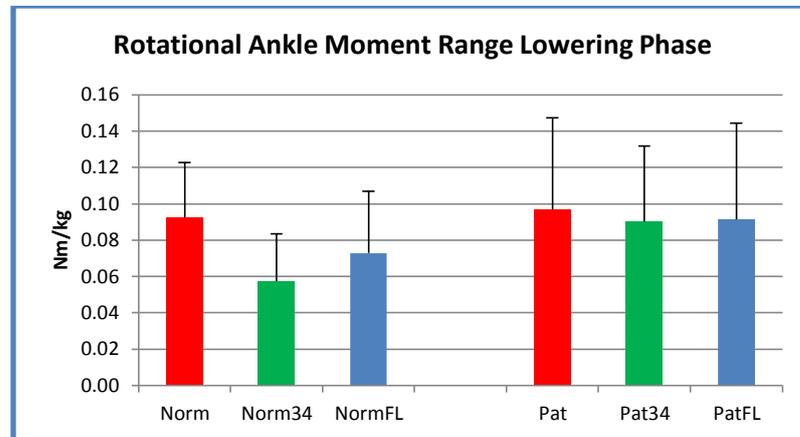


Fig.8.63. Ankle moment rotational range during the lowering phase of step descent illustrating the differences between the normals and patients and how each of the three shoe conditions affected each group (Error bars = standard deviation)

Table 8.12. Maximum moments and mean ranges of moments of the ankle comparing the normal and patient group data and recording any differences seen with the orthoses

Ankle Moments Step	Mean Difference Normals Vs Patients	Significance Normal Vs Patients	Confidence Intervals Normals Vs Patients	Mean Difference No Orthoses Vs Orthoses		Significance Shoe only condition Vs orthoses	
				3/4	FL	3/4	FL
Max Dorsiflexion M	-0.056	P=0.041	-0.110 to -0.002	0.072	0.088	P=0.032	P=0.010
Sagittal M Range	-0.021	P=0.585	-0.095 to 0.054	-0.011	-0.041	P=0.813	P=0.374
Max Inverting M. F.C.	-0.160	P=0.000	-0.210 to -0.110	0.007	0.015	P=0.809	P=0.374
Coronal M Range F.C.	-0.133	P=0.000	-0.186 to -0.081	0.045	0.043	P=0.166	P=0.187
Max Inverting M. L.P.	0.088	P=0.776	-0.048 to 0.064	-0.036	-0.036	P=0.299	P=0.302
Coronal M Range L.P.	0.028	P=0.089	-0.004 to 0.061	-0.006	-0.013	P=0.756	P=0.538
Max External M. F.C.	-0.014	P=0.238	-0.038 to 0.010	-0.023	-0.027	P=0.116	P=0.071
Transverse M Range F.C.	-0.004	P=0.643	-0.019 to 0.012	0.007	0.008	P=0.461	P=0.400
Max External M. L.P.	-0.028	P=0.056	-0.057 to 0.001	0.025	0.026	P=0.157	P=0.153
Transverse M Range L.P.	-0.019	P=0.033	-0.036 to -0.002	0.012	0.013	P=0.049	P=0.237

8.3. Summary of key findings Chapter 8:

Calcaneal segment:

- During walking the patient group demonstrated less dorsiflexion than the control group; this sagittal plane difference was not repeated during the step descent.
- Only the full length orthoses reduced maximum eversion and coronal plane range during the early stance phase of walking, although the $\frac{3}{4}$ orthoses did demonstrate a marked trend towards reducing maximum eversion in early stance.
- Minimum and maximum eversion in the forward continuum phase and maximum eversion in the lowering phase were decreased during the step trials, coronal plane range and transverse plane range were also increased in the patient group during the lowering phase.
- During the step trials both orthoses increased the range of dorsiflexion, but reduced the minimum and maximum everted position during the forward continuum phase. There were no significant results in the transverse plane.

Metatarsal segment:

- The patient group demonstrated significantly increased sagittal plane range during late stance and increased coronal plane range during early stance over the pain free group in the walking trials.
- There were no differences recorded during the step trials between the two groups.
- During the walking trials only the full length orthoses reduced the sagittal plane motion during late stance, both orthoses reduced the transverse range during early stance. Both orthoses significantly reduced the coronal plane range during late stance and this was also seen in the lowering phase of step descent. Similarly only the $\frac{3}{4}$ orthoses produced significant reductions in transverse plane rotation during late stance, while walking and during the lowering phase of step descent.

Knee kinematics:

- In the sagittal plane during walking, maximum extension only was significantly increased in the patient group. Maximum adduction was significantly increased, but abduction was decreased. In the early stance phase both maximum external position and transverse plane rotation range were increased in the patient group.
- During step descent the patient group demonstrated more significant changes; maximum flexion; flexion range; maximum adduction; coronal plane range and transverse plane range during the forward continuum phase were all increased. Only maximum adduction was significantly greater in the pain free group.
- There were no significant changes seen with the orthoses during either the walking or step descent trials.

Knee moments:

- During the walking trials the maximum flexion moment was increased in the patient group during late stance. Through early stance both the maximum abduction moment and the coronal moment range were reduced in the patient group.
- Similarly to the knee angular data during the step trials there were significant increases seen in the patient group. Maximum abduction moment, coronal moment range and transverse moment range were increased during the forward continuum phase, while in the lowering phase maximum flexion moment, flexion moment range and coronal moment range were increased.
- The orthoses did not record any significant changes during the walking trials, but there was a significant reduction seen with both in the coronal moment range during the forward continuum phase.

Ankle moments:

- The patient group demonstrated significant reductions in the maximum dorsiflexion moment, sagittal plane moment range and the maximum everting moment compared to the pain free group.

- The maximum dorsiflexion moment was also reduced in the step trials but there were increases seen in the coronal plane when looking at the maximum inverting moment and the moment range during the forward continuum phase, and the transverse moment range during the lowering phase.
- The orthoses did not record any significant changes during the walking trials but both orthoses increased the maximum dorsiflexion moment during the step down trials. Only the $\frac{3}{4}$ orthoses decreased the rotational moment range during the lowering phase.

Chapter 9. Discussion symptomatic subjects walking and step descent

This chapter discusses the similarities and differences, when walking and descending a step, between two groups of fifteen subjects: Group one, healthy subjects with no pain; Group two, subjects diagnosed with patellofemoral pain and excessively pronated feet. By using the previous methods and techniques addressed in this thesis (Chapters 4 and 5) differences in the kinematics of the knee and foot segments; the kinetics of the knee and ankle between the groups and the $\frac{3}{4}$ and full length orthoses are explored.

9.1. Rearfoot kinematics walking and step descent

9.1.1. Comparison of rearfoot kinematics non-symptomatic and symptomatic subjects - walking

There were no differences seen in the range of plantarflexion between the symptomatic and non-symptomatic subjects, however there was a significant reduction seen in the range of dorsiflexion in the patient group (mean difference 1.68° , CI 0.32-3.02). This was in contrast to the study undertaken by Levinger and Gilleard, (2007) who reported almost identical values for dorsiflexion between their groups of normals and patellofemoral patients, although Barton, et al. (2011) reported a non-significant reduction of -2.5° (CI-5.3 to 0.5) in their patient group leading them to report a trend towards significance ($p=0.097$). It must be emphasized that the subjects in both the latter studies were recorded barefoot. It is obviously important to establish foot movement in the specified groups, but it does not evaluate the subjects in their normal every day functioning state which for the vast majority in the western world will be shod.

Rearfoot eversion during the walking trials was not statistically significant between the groups; this was in agreement with both Levinger and Gilleard, (2007) and Barton, et al. (2011) although peak eversion in this study was higher than the two aforementioned studies, which may be as a consequence of recording the subjects shod this will be discussed later (Chapter 10). Prior to this Messier, et al. (1991) had noted that excess eversion of the heel had been held responsible for many lower limb injuries when running, as in the present study they found no differences between

their patellofemoral pain group and their normal group. As Messier, et al. used a single segment foot model it is not possible to directly compare their results with the present study, these differences will be discussed later (Chapter 10). The range of coronal plane excursion was also measured in the present study and this was considered as early and late stance phase. The early stance phase range demonstrated no differences, but the later range showed the patient group had an increased mean difference of 1.58° (CI -3.89- 0.73), though this was not significant change, it could be an indication of instability in some of the symptomatic subjects.

The transverse calcaneal segment rotation in early and late stance phase demonstrated no differences between the groups in this study, Levinger and Gillear, (2007) split their rotation into peak abduction and adduction; the former was significantly increased in the control group. Barton, et al. (2011) also found an increased peak internal rotation was higher in their control group, but this was not significant though the mean difference was 4.2° (CI -0.9 to 9.3). They also reported mean range; this was for the whole of stance phase, but was within half a degree of the early stance phase range of this study. The significant results of the previous studies compared to the present study are not too surprising when realising they both recorded their subjects barefoot. From the initial part of this work in Chapters 5 and 6 it was found that the shoes tended to increase the range of motion compared to the sandal condition with the markers placed directly on the heel. The shoe mounted markers could also mask any small systematic differences between the groups, or the shoes could have provided enough support to even the range between the groups.

9.1.2. Comparison of rearfoot kinematics non-symptomatic and symptomatic subjects - step descent

As there are no previous studies that have considered multi-segment foot motion during step descent in subjects with patellofemoral pain, there are no direct comparisons to make. In the sagittal plane there were no significant differences between the groups, however the symptomatic subjects did have an increased mean range of dorsiflexion of 1.72° , which demonstrated a shift in the confidence intervals (CI -4.132 to 0.688).

During the forward continuum phase it was the control group which demonstrated significantly more eversion at both the minimum position and the maximum position, although the total excursion was a little larger in the knee pain group this was not significant. During the early step phase the stance limb will not be under great eccentric strain as the swing limb is still within line with the top step, so the lack of extra excursion in the patient group was not unexpected. The reduction in maximum eversion recorded could be as a result of the patients trying to keep their feet as stable as possible, clinically it is believed that supination of the sub-talar joint is one method of making the foot more rigid.

Similar to the forward continuum phase, maximum eversion in the lowering phase was significantly greater in the non-symptomatic group; the mean difference was just over 1.7° (CI -4.13 to 0.69). When the coronal range was considered it was the patient group that demonstrated a significantly increased excursion (mean difference 2.0° , CI -3.31 to 0.75). The reduced maximum position in the patient group could be related to the earlier position, but equally could be a compensation mechanism to aid knee stability as the knee flexes to its deepest point just before the swing limb contacts the lower step.

The rotational range of the calcaneal segment on the shank was very similar between the groups in the forward continuum phase, however the patients did have significantly greater excursion in the lowering phase. The heel of the stance limb tended to lift just before the swing limb contacted the lower step, therefore it would be logical to assume the heel was fixed for most of the phase; this would suggest that the excursion was emanating from rotation of the shank on the calcaneal segment. As the patients in this study did not complain of pain during the trials there is no real argument for increased rotation being a compensation mechanism, however it could be an etiological factor leading to patellofemoral pain, which could be supported by documented evidence that tibial rotation can affect the contact areas of the patellofemoral joint (Hefzy, et al. 1992; Lee, et al. 2003, Chapter 3, Section 3.2.5.).

9.1.3. Effects of orthoses rearfoot kinematics - walking

The movement of the rearfoot was unaffected by the orthoses in the sagittal plane, there was a small trend seen with the $\frac{3}{4}$ orthoses which marginally increased the range of dorsiflexion in both groups. In contrast to the present study Chevalier and Chockalingam, (2012) reported that nine of their eleven custom made orthoses tended to decrease maximum dorsiflexion on a single medial knee pain subject, although the modified plimsolls with no heel they used to “contain” the orthoses could account for some of the differences. They used the Oxford foot model and stated that it was changes in the sagittal plane that were the most systematic, as the results in the other planes were not as consistent. However Chevalier and Chockalingam, only reported on maximum positions, whereas Wilken, et al. (2012) noted that minimal detectable change would be decreased considerably when range of motion is considered.

In this study the coronal plane maximum eversion in early stance was reduced significantly by the full length orthoses and the $\frac{3}{4}$ orthoses produced a definite trend in the same direction ($p=0.062$, CI -0.8- 3.26). A 5° medial wedge plus the arch support would be prescribed clinically to resist eversion; hence this reduction was predictable during the contact phase. The full length wedge orthoses produced a mean difference of just less than 2° which is a reduction of over 35%. Clinically some patients may report a 2° difference, while others may not; this could be due to a number of factors, such as foot type or proprioception. In a static standing trial Payne, et al. (2002) reported that their 6 prefabricated orthoses with different arch heights reduced calcaneal eversion by 1.99° to 3.05° , the latter change being as a result of using the highest most rigid device. Further comparisons with other studies are difficult due to foot modelling or the lack of footwear.

The range of coronal plane motion in early stance was also reduced significantly by the full length orthoses, but this time the $\frac{3}{4}$ orthoses did not display the same trend in the mean data. The wedge under the forefoot plus the arch support would be assumed to have more of an effect after forefoot loading and hence as stance phase was split at 50% this result could have been expected. The lack of significant difference in late stance phase would also be expected as the effect of the arch

support is lost as the heel lifts. There were no changes seen in transverse rotation of the calcaneal segment of the shank with either orthoses.

9.1.4. Effects of orthoses rearfoot kinematics - step descent

There were trends seen in the data towards both orthoses increasing the maximum dorsiflexed position of the calcaneal segment on the shank. However, when considering the total range of dorsiflexion, this was significantly increased by both orthoses by approximately 3.5° (CI -6.68 to -0.81 and -6.36 to -0.44, $\frac{3}{4}$ and FL respectively) and this may be as a consequence of reducing coronal plane range.

The rearfoot remained everted throughout step descent during the forward continuum phase, both minimum and maximum eversion was reduced significantly by each orthosis; this however did not translate to a range reduction. Maximum eversion in the lowering phase did show trends for reduction with both orthoses, though neither reached a significant level, the full length orthoses was just outside the level set ($p=0.063$, CI -0.11 to 3.96). A medial wedge decreasing eversion would seem logical at first glance, but further consideration would seem to contradict the clinically held belief that medially wedging the forefoot would unlock the midfoot causing late stage pronation. The range of eversion during the lowering phase was unaffected by the orthoses as was the rotation range during both the forward continuum and lowering phases.

9.1.5. Summary rearfoot kinematics

When walking the only significant difference seen at the rearfoot between the groups was a decrease in dorsiflexion in the patient group. Initially this could have been attributed to a limitation in movement, but considering there was a non-significant increase in dorsiflexion during the step trials it is probably more related to a reduced step length. Powers, et al. (1999) suggested that a reduced step length may be used to reduce knee flexion. It is interesting that during the walking the significant reduction in dorsiflexion was related to a non-significant increase in eversion in the patient group during propulsion. While on the step trials there was a non-significant increase in dorsiflexion but a significant decrease in eversion.

When the effects of the orthoses were considered they only seemed to affect the coronal plane motion, but during the step descent trials there was a significant change in the sagittal plane also. The increase in dorsiflexion cannot really be explained looking solely at the sagittal plane, however when considering the effect of the orthoses on the coronal plane motion of the calcaneal segment, significant reductions were seen in the minimum and maximum everted positions during the forward continuum phase while the heel was still on the step. This limitation could certainly account for the increase in the sagittal plane. Abboud, (2002) recognised that pronation of the subtalar joint consists of abduction, eversion and dorsiflexion; logically reducing one motion could produce the need to find compensation from another plane. Clinicians need to consider that reducing joint motion in one plane may put extra demands on that joint in the other planes.

9.2. Metatarsal segment kinematics walking and step descent

9.2.1. Comparison of metatarsal kinematics non-symptomatic and symptomatic subjects - walking

All parameters for the metatarsal segment were considered in two phases, early and late stance, this was achieved by splitting the stance phase in two. From the results in Chapter 5, the sagittal plane was split at 30% this was to capture the motion from heel strike to forefoot loading, then the second phase was to capture midstance and propulsion. In the coronal and transverse planes it was felt bisecting the stance phase would make certain of capturing the eccentric and propulsive phases.

In the sagittal plane there were no differences seen between the groups during the eccentric phase of early stance, which would be expected, as their forefoot will not be in contact with the ground. In the propulsive phase there was a significant increase seen in the patient group, this was less than a degree and therefore it would be unlikely to be recorded clinically. It is difficult to compare any of the metatarsal segment results with previous papers due to the present study using shod subjects. Barton, et al. (2012) reported a mean range of 7.9° for their 25 barefoot patellofemoral subjects, it was expected that the range for the present work would be

less due to the influence of the shoes, however, the mean range for the patient group was 6.4° (CI 5.61-7.18) for the propulsive phase; closer than anticipated.

Barton, et al. (2011) reported on the whole of stance phase and recorded peak dorsiflexion of the forefoot on the rearfoot as 12.5° . Their symptomatic cohort demonstrated 0.9° more than their controls; this was similar to the present study, Barton, et al. recorded this as a non-significant difference. Leitch, et al. (2012) published a short study looking at twelve midfoot striking runners, six with patellofemoral pain and six controls. Running barefoot on a treadmill, using the Oxford foot model, they also reported a non-significant reduction of peak forefoot dorsiflexion in the control group. Noehren, et al. (2011) also compared two groups of runners with and without patellofemoral pain; peak forefoot dorsiflexion was within half a degree and therefore was not significant, which was in contrast to the present study. Running studies cannot be used as a direct comparison with the present study; however it is the only other evidence that has been reported to date on midfoot motion between non-symptomatic subjects and patellofemoral pain patients. Leitch, et al. found the symptomatic group demonstrated significantly greater rearfoot eversion and suggested that this lead to greater forefoot dorsiflexion.

In the present study the patient group did demonstrate a significantly increased coronal plane range over the normal group; the total range was 1.5° (CI 1.19-1.74) on the calcaneal segment which will include forefoot loading. Later in the propulsive phase the range was still small at just over 2° (CI 1.89- 2.65), but at this stage of the gait cycle there were no differences seen between the groups. To the authors knowledge no previous studies have split the metatarsal segment motion into early and late stance. Barton, et al. (2011) reported greater peak forefoot supination relative to the rearfoot in their control group, but the mean difference was only half a degree. This was not a significant result and once again it must be highlighted that their study was conducted barefoot.

In the present study there were no differences seen in the transverse plane, the range of abduction/adduction was similar in both groups during early and late stance phase. Barton, et al. (2011) and Noehren, et al. (2011) both reported their control group had a non-significant increase in mean peak abduction.

9.2.2. Comparison of metatarsal kinematics non-symptomatic and symptomatic subjects - step descent

There were no significant differences seen between the symptomatic and control groups when looking at the maximum positions or excursions of the metatarsal segment with respect to the calcaneal segment.

9.2.3. Effects of orthoses metatarsal kinematics - walking

During the contact phase of stance neither orthoses altered the sagittal plane range of the metatarsal segment on the calcaneal segment. As mentioned previously, during this stage of stance the forefoot is not loaded. It would therefore not be unreasonable to assume an arch support or wedge would have little effect on the dorsiflexion/plantarflexion of the metatarsal segment. During midstance and propulsion the full length orthoses did produce a significant reduction compared to the non-orthoses condition, however it must be highlighted that though significant the mean difference was just less than 0.75° (CI 0.064- 1.43). It is unlikely that this difference would be noticed in the clinical situation though it is not known whether this could be clinically relevant.

The exact mechanism of this reduction is unclear, it could be due to a number of factors: the material of the full length wedge may be stiffening the shoe or functionally the arch support could be reducing the maximum motion towards dorsiflexion at midstance before the heel lifts and the metatarsal segment plantarflexes to its maximum position, hence reducing the range of motion. Ferber and Benson, (2011) also reported a significant reduction in plantar fascia strain, but no difference in medial longitudinal arch angle. Although their method of constructing a triangle on the medial side of the foot cannot be directly compared to the present study, it may be revealing a common effect.

During early stance the forefoot would be loaded and both orthoses demonstrated a trend towards reducing the range of motion of the metatarsal segment on the rearfoot segment; however the mean differences were small. From midstance to the end of propulsion both orthoses produced a highly significant reduction in the range of motion ($p=0.000$). The mean difference was still small around 0.6° (CI $\frac{3}{4}$ 0.27-0.92, FL 0.30-0.96) with both orthoses. On initial consideration this would appear

insignificant, however it equates to around a 25% reduction in range. The relatively small mean difference would suggest the effect was systematic throughout the subjects. Williams, et al. (2003) looked at how orthoses affected rearfoot motion with a single segment foot model. Due to varying results within their cohort they hypothesized that orthoses may provide the most control at the midfoot. This view was echoed by Mündermann, et al. (2003) who used a single segment model and stated that their moulded orthoses increased maximum inversion and inversion foot velocity, hence leading them to state orthoses have a significant role during late stance phase. The present study is the first to look at control of orthoses on the metatarsal segment in normal footwear. It appears the method used is sensitive to pick up small changes in range of motion. Although this may be the first indication of a minimal detectable change (MDC) it is unknown what the minimal clinical important difference (MCID) would be.

Both orthoses significantly reduced the range of transverse plane motion of the metatarsal segment on the calcaneal segment in the early phase of stance. It could be that the arch support of both orthoses was limiting the movement towards abduction in midstance that produced this result. Once again the mean difference for both orthoses was small (0.53° , CI 0.15-0.91 and 0.77° , CI 0.39-1.15) but 0.5° is a 20% reduction during early stance. Chevalier and Chockalingam, (2012) also reported, on the whole, reductions in range for their single subject with different orthoses, there were only 2 increases out of the 22 orthoses. The maximum increase was 2.9° while the maximum reduction was 4.1° . In late stance it was only the $\frac{3}{4}$ orthoses that reduced the range significantly; this would seem to discount that this was anything to do with making the shoe stiffer with the longer wedge. From a clinical theory perspective it may be the medial wedge under the metatarsal heads effectively inverting the mid-tarsal joint unlocking the forefoot and hence reduced the stabilising effect.

9.2.4. Effects of orthoses metatarsal kinematics - step descent

The orthoses had no significant effect on the sagittal plane range over the shoe only condition; this lack of effect was also seen in the forward continuum phase in the coronal plane. It must be remembered that metatarsal segment motion is measured

with respect to the calcaneal segment, therefore if both segments move together there will be no net motion. Chang, et al. (2008) investigated forefoot motion with respect to the rearfoot (3 healthy subjects). When the forefoot inverted at the same time as the rearfoot they termed this "in-phase", eversion of the forefoot on an inverting rearfoot would be termed "anti-phase". Chang, et al. recorded simultaneous forefoot inversion and adduction. This would be contradictory to the clinically hypothesis leading them to conclude that previous descriptions of forefoot to rearfoot motion devised from cadaver experiments "may oversimplify the complexity of the interactions".

The only significant reaction to the orthoses in the present study was seen in the lowering phase where eversion of the metatarsal segment was reduced. The mean difference was less than 0.5° (CI 0.13 to 0.76, 0.14 to 0.78 34 and FL respectively) but again this equated to almost a 60% reduction. This result would be expected with the full length wedge. The $\frac{3}{4}$ orthoses may have been expected to demonstrate a little less control as the GRF moved towards the phalangeal segment, this may suggest that the arch support of the orthoses is the most important factor here. There were no significant effects in the transverse plane, but the $\frac{3}{4}$ orthoses did demonstrate a trend towards reducing the range of adduction during the lowering phase.

9.2.5. Summary metatarsal kinematics.

When walking the patellofemoral pain group did demonstrate a significant increase in propulsive phase sagittal plane range and early phase coronal range compared to the normal group. Certainly the increase in coronal plane motion during early stance could be affecting the sagittal range in the later phase. The increased sagittal plane range in the metatarsal segment could also be related to the significant reduction of dorsiflexion of the rearfoot. The complete lack of differences seen in the metatarsal segment results during step down would suggest this segment is not responsible for either cause or compensation during this task.

During walking only the full length orthoses significantly reduced the range of sagittal plane motion in the propulsive phase. Both orthoses demonstrated highly

significant reductions in the coronal plane during propulsion and though the mean difference was small the percentage effect was substantial (approximately 25%). Both orthoses significantly reduced the range abduction/adduction in the early phase of stance, but it was only the ¾ orthoses that reduced the range in the later phase. This was attributed to the arch support before the heel lifted and it may be that the medial forefoot wedge of the full length orthoses did tend to unlock the midfoot after heel lift. The only effect seen with the orthoses during step descent was in the lowering phase where again there was a significant reduction, this may be related to increasing stability, but may also be as a result of reduced rearfoot motion not “twisting” the forefoot as much.

9.3. Phalangeal segment kinematics walking and step descent

9.3.1. Comparison of phalangeal kinematics non-symptomatic and symptomatic subjects - walking

Maximum dorsiflexion of the toe segment on the metatarsal segment did exhibit a non-significant reduction in the patient group compared to the normal group (mean difference 2.2°, CI -1.10- 5.44) however this was not reflected in the range of motion. Danenberg, (1993) did speculate that hallux limitus can be associated with knee arthritis, though there was no further explanation involving the knee. The lack of difference seen between the groups in this study would suggest that this was not a main contributing factor. It must be noted that this study could not look solely at first metatarsophalangeal joint function due to the modelling of the foot and the positioning markers on the shoes.

9.3.2. Comparison of phalangeal kinematics non-symptomatic and symptomatic subjects - step descent

There were no significant differences seen between the symptomatic and control groups when looking at the maximum dorsiflexion or sagittal plane excursion of the phalangeal segment on the metatarsal segment during step descent.

9.3.3. Effects of orthoses phalangeal kinematics - walking

There was a definite trend towards reduction of maximum dorsiflexion of the phalangeal segment on the metatarsal segment with the full length orthoses ($p=0.060$) leading to a mean difference of 3.8° (CI -1.7- 7.86). This was not repeated with the $\frac{3}{4}$ orthoses over the shoe only condition; this could be due to the extra stiffness of the material under the toes or it could be interference with the windlass mechanism (as discussed in Chapter 6). When the range of phalangeal motion was considered the full length orthoses did restrict the excursion significantly; the mean difference was just under 4.5° (CI 1.61- 7.27).

Clinically wedging under the first ray is thought to restrict first metatarsophalangeal joint dorsiflexion by not allowing the proximal phalanx of the hallux to access the dorsal articular surface of the metatarsal, hence interfering with the windlass mechanism. Lafuente, et al. (2011) stated that if the hallux cannot dorsiflex adequately it may cause supination of the forefoot, collapse of the medial arch and late stage pronation. Although the subjects in this study demonstrated limitation in phalangeal segment excursion, calcaneal eversion was decreased; transverse rotation and dorsiflexion remained the same with the orthoses. These results may infer that there is no direct connection between limitation of toe extension and rearfoot pronation, further research is required to establish if there is a threshold value for restriction which is clinically detrimental.

9.3.4. Effects of orthoses phalangeal kinematics - step descent

Unlike the walking trials there was no trend seen towards reducing peak dorsiflexion during step descent with either orthoses. However phalangeal segment range with respect to the metatarsal segment was only just outside the level set for significance ($p=0.054$, CI -0.5 to 6.89 and $p=0.051$, CI -0.19 to 6.99, $\frac{3}{4}$ and full length respectively). This suggests that the phalangeal segment started in a more dorsiflexed position and hence reduced the excursion; the mechanism by which this occurred was not determined in this study.

9.3.5. Summary Phalangeal segment kinematics.

The maximum position or range of motion of the phalangeal segment was not significantly different between the symptomatic and non-symptomatic groups; this

may suggest that sagittal plane motion is not contributing to knee function as either an aetiology or a compensatory mechanism during walking or step descent.

The full length wedged orthoses did tend to reduce the phalangeal segment range of motion which may be as a result of interference with the windlass mechanism or due to the stiffness of the extra material. If the windlass mechanism was compromised this study could not support the clinically held belief that this can lead to excess pronation of the rearfoot.

9.4. Knee kinematics during walking and step descent

9.4.1. Comparison of knee kinematics non-symptomatic and symptomatic subjects - walking

The patient group in this study did not demonstrate any difference in knee flexion angle at heel strike compared to the non-symptomatic cohort. This was also reported by Powers, et al. (1999) and Paoloni, et al. (2010) the latter investigators suggested that sagittal plane motion may not be the dominant factor in the development of patellofemoral pain. Paoloni, et al. also noted that level walking may not load the knee enough to induce the compensation strategies used to reduce the symptoms. Wilson and Davis, (2008) had previously suggested that during low impact activities subjects with patellofemoral pain may possess enough strength to produce normal kinematics, though this may not be the case when undertaking more demanding tasks.

There was a significant difference in knee extension during midstance between the two groups; with a mean difference of 2.2° (CI 0.69 to 3.74), the patient group extended the knee further than the controls. Farrokhi, et al. (2011) demonstrated with a finite element experiment that greater flexion angle of the knee lead to greater compression within the patellofemoral joint. It could be postulated that this significant reduction in the flexion angle was an attempt by the patients to reduce the joint load. Previous studies have not highlighted this and therefore could be an anomaly within the present cohort. There was a trend seen in the loading response towards the patient group flexing less (mean difference 1.5° , CI -0.37 to 3.40) but this was not significant. No other studies have reported significant changes in this

parameter during normal walking, although Powers, et al. (1999) did report a significant reduction in the flexion angle of their patellofemoral cohort compared to their controls during fast walking (16.9° vs 21.6° $P=0.04$).

In the coronal plane the patient group demonstrated a significant increase in peak adduction, while the normal subjects revealed significantly greater peak abduction; this led to the range being similar, but there was a trend towards the patients having a larger mean excursion. Paoloni, et al. (2010) also reported a significant peak adduction angle in their symptomatic group, they stated this was during the loading phase. No such increase was found in the peak motion by Barton, et al. (2011) both these studies used a Helen Hayes marker set to model the knee as opposed to the CAST technique used in this study.

In this subject sample the absence of any link between coronal plane motion of the calcaneal and metatarsal segments may suggest that coronal plane motion of the knee is not influenced by foot motion. Pitman and Jack, (2000) suggested patellofemoral pain could be caused by dynamically increased Q-angle; the increase in peak adduction found in the present cohort would seem to be in contradiction to this theory. However measuring tibiofemoral motion may not be directly comparable to patella motion. Powers, (2003) discussed knee valgus, but recognised that coronal plane motion of the knee may be under proximal as well as distal control; hence foot position may not be the only controlling factor in coronal plane knee mechanics.

In this study both maximum external rotation of the knee and transverse knee excursion were significantly higher in the patient group during early stance, although this cannot be linked with either transverse or coronal plane motion of the foot segments. The mean difference of 4.4° (CI -6.59 to -2.16) seen in the maximum position would probably be seen clinically and could possibly be linked to greater patellofemoral pressure (Chapter 3.2.5. tibial rotation). Noehren, et al. (2012) stated they expected to find increased external rotation in their symptomatic running group, but in contrast to this study they found more of a trend towards internal rotation. They postulated that this may have been a compensation mechanism to reduce pain. A non-significant increase in internal rotation was also reported by Leitch, et al. (2012) in their runners with patellofemoral pain. They speculatively proposed that

this rotation in combination with the significantly greater rearfoot eversion at toe off supported the theory that prolonged rearfoot eversion causes prolonged internal rotation of the tibia. Certainly there was no evidence in the present study to support this theory; however the differences between barefoot running and shod walking may go some way towards explaining the contrary findings of the two studies. Splitting the stance phase into early and late phases may have also uncovered some differences, although there were no significant rotational differences between the groups in the late stance phase.

9.4.2. Comparison of knee kinematics non-symptomatic and symptomatic subjects - step descent

Both maximum flexion of the knee and sagittal plane excursion of the knee were found to be significantly increased in the patellofemoral pain group over the control group. The peak angle in the present study was much less than that reported by Salsich, et al. (2001) they did not report any difference between their symptomatic and non-symptomatic groups, this was not discussed as their primary focus was on the joint moments. Brechter, et al. (2002) found their symptomatic group to have a reduced cadence when descending steps, but they also reported that there was no significance between the groups regarding knee kinematics.

In contrast to the present study Crossley, et al. (2004) reported that their symptomatic group demonstrated significantly reduced peak knee flexion. The Figures published were almost half of the present study though the step height was the same; the kinematic data was recorded in a 2D format which may have induced some differences. This result lead them to conclude their knee pain group flexed their knees less in an attempt to reduce joint pressure, though due to discrepancies in the pain measures and the kinematics they suggested that individuals may adopt different strategies to accomplish their pain reduction. Grenholm, et al. (2009) also reported a minor non-significant reduction in the knee flexion of their symptomatic cohort over their control group. They recognised that their hypothesis of reduced knee flexion could not be accepted; this was explained by suggesting reduced patellofemoral joint loading could be achieved by greater plantarflexion of the ankle of the swing limb, hence loading the other limb earlier. Due to most subjects not

actually having pain descending the steps in this study the extra flexion seen in this study may be more of a “cause” rather than a compensation strategy.

Although there are many references to sagittal plane motion of the knee during step descent, fewer studies deal with coronal plane motion. Yu, et al. (1997) compared level walking with stair ascent and descent. They reported that all three tasks were undertaken with the knee in a varus position, but peak angle was greater when on stairs; they reported there was no difference between ascent and descent. This was in contrast to the present results where walking tended to demonstrate a mean increase of 1° over the step descent trials.

In the present study maximum adduction and coronal plane excursion were significantly larger in the patient group, but maximum abduction was significantly greater in the control group. Range of mean coronal plane motion was just over 5° (CI 4.49 to 5.78) in the patient cohort. This was a little more than the range illustrated in Selfe, et al. (2007) however this was for a normal sample and therefore was more comparable to the control group in this study. In a later study by Selfe, et al. (2011) they reported coronal plane excursion of 8.9° which was a lot higher than the patient group in this study, however their subjects were performing a slow descent.

Knee motion in the coronal plane can be mapped to motion of the calcaneal segment in this study. The patient group demonstrated less maximum eversion than the controls, but a similar comparison at the knee shows greater adduction in the patient group. Fig. 9.1 shows two subjects on the step descent trials at a similar point in the lowering phase. A. is a symptomatic subject and is demonstrating greater knee adduction. Neither subject has any orthoses in their shoes in this illustration therefore the difference seen could be due to compensation strategy in an attempt to reduce pain.

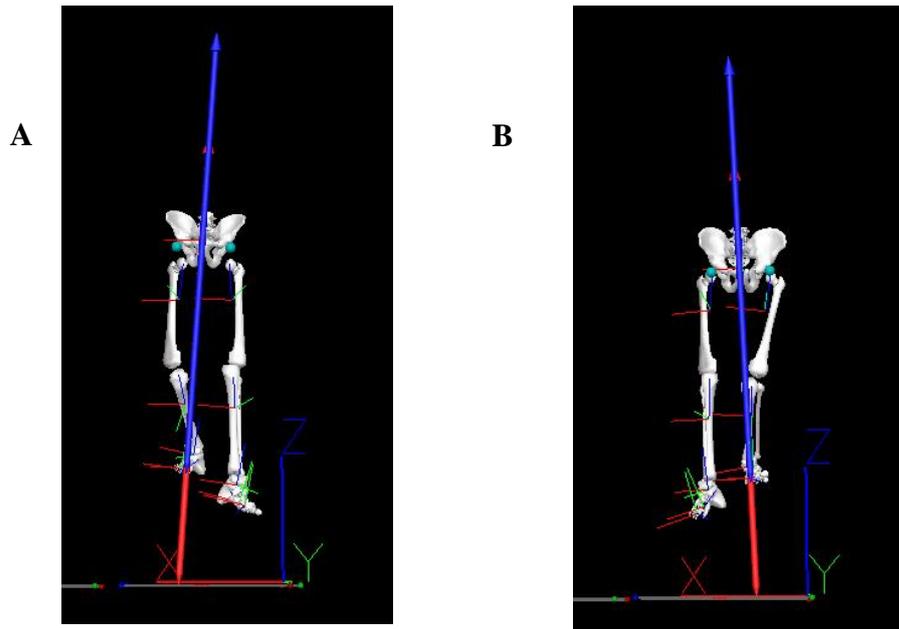


Fig. 9.1. Illustration of step descent at similar frame A. patient demonstrating more knee adduction than control in B.

In the transverse plane the range of rotation was significantly higher in the patient group during the forward continuum phase of step descent. Few studies have dealt with transverse rotation of the knee during step descent at normal speed. Selfe, et al. (2011) recorded a slow controlled step descent with a group of 13 subjects with patellofemoral pain. They reported a rotation excursion of 6.79° , this was almost double the range reported in the present study, however this was for the whole of the step down and the slow flexion rate may have challenged the stability of the knee further.

Similar to the later propulsion phase of walking, the lowering phase of step descent did not illustrate any differences between the groups. In this study it would seem that the patellofemoral pain subjects did demonstrate both increased ranges of calcaneal segment eversion and an increased range of knee rotation. However this was only true for the forward continuum phase; once the ground reaction force moved forwards and the heel prepared to lift into the lowering phase the groups tended not to reveal any differences. The lack of difference seen in the patient and control groups in the metatarsal segment would suggest that it has little influence over the knee, at least during step descent.

9.4.3. Effects of orthoses knee kinematics - walking

When compared to the shoe only condition neither orthoses made any significant changes to the mean kinematics of the knee in any plane, nor were there any trends to report. Pitman and Jack, (2000) suggested that biomechanical foot orthoses can be an effective first line treatment of patellofemoral pain. In their study over 60% of their subjects, who were grouped by age and gender, reported improvements in pain, the average time before the improvement was noted was between one to four weeks.

Collins, et al. (2008) undertook a randomised clinical trial which compared prefabricated orthoses, flat insoles, physiotherapy alone, foot orthoses and physiotherapy combined. The pain scales of 100 women were recorded before and after the interventions. They concluded from the results of the pain scales that recovery may be hastened by providing foot orthoses. Barton, et al. (2010) conducted a study whereby 60 patellofemoral pain patients were issued with a pair of orthoses then completed three challenges. The number of “pain free” step downs, single leg rises from sitting and single leg squats were recorded with reference to their subjective pain scores. Barton, et al. concluded at 12 weeks improvements in performance were better than the immediate results due to the fact the subjects were able to perform more pain free trials. In the present study only two of the subjects used in this study had mild knee pain during the walking trials; perhaps compensation strategies would have been more noticeable with a cohort with greater pain.

The findings of the present study agree with Boldt, et al. (2013) who tested medially wedged orthoses on a sample of 40 females, 20 with patellofemoral pain and 20 controls, while running. They reported that the orthoses had little effect on knee or hip joint mechanics when running. They also found that their two groups of subjects tended to react in a similar manner to the orthoses; the orthoses they used comprised a full length 5° wedge fitted into a standard shoe.

9.4.4. Effects of orthoses knee kinematics - step down

There were no significant changes or trends seen within the knee data, considering the alterations seen within the foot segments this must suggest that it is the foot and ankle joints which “absorb” the effects of the orthoses. In their review of the effects of orthoses Gross and Foxworth, (2003) suggested that there appeared to be limitation of tibial internal rotation which in turn could decrease the Q-angle. They were not looking at experiments on step down, but there was no evidence to support either of these effects in the present study.

There are no other experiments in the literature that investigate the effect of foot orthoses on knee kinematics during step down, the only statement that can be made is the foot orthoses used in this experiment did not affect knee kinematics in this cohort of patellofemoral pain and normal subjects. It may be that the hip musculature is dominant during step descent. Perhaps in the future a similar study to that conducted by Bellchamber and van Den Bogart, (2000) could be useful to look at power flow to determine if it is the hip or the foot that influences knee rotation during step descent. Further research may find that the significant adduction seen in the patient group in this study could be a dominant compensatory mechanism during step descent to aid the unloading of the lateral facet of the patella.

9.4.5. Summary knee kinematics.

There were more significant differences seen between the symptomatic and non-symptomatic groups in this study than had previously been reported when walking (Powers, et al. 1999; Paoloni, et al. 2010; Noehren, et al. 2012; Leitch, et al. 2012). These differences were seen in all three planes, maximum extension, adduction, external rotation and transverse excursion were increased in the patient group. This would question the statement that walking is not “strenuous” enough to induce compensation mechanisms (Crossley, et al. 2004; Wilson and Davis, 2008).

Similarly when looking at the step descent results there were significant differences to report in all three planes. Maximum flexion and flexion range were increased in the patient group; there is no logical argument to define this as a compensation

strategy leading to the assumption it may be a factor in the instigation of pain. Similar trends to the walking trials were seen in the coronal and transverse planes.

There were no significant differences produced with either orthoses when walking or during step descent. This may indicate that it is the hip musculature that is dominant in this group of patients; it may also be that individual reaction to the orthoses as suggested by Payne, et al. (2002) may be disguising any subtle differences/effects.

9.5. Knee moments during walking and step descent

9.5.1. Comparison of knee moments non-symptomatic and symptomatic subjects - walking

The maximum flexion moment in early stance and the maximum extension moment did not demonstrate any significant differences between the groups. There was a small shift in the confidence intervals for the extension moment, but this did not reflect the significant difference seen in the angular data. Paoloni, et al. (2010) did find a significant difference in the extensor moment, the mean peak moment was very similar to this study (0.083Nm/kg to 0.085Nm/kg, CI 0.77 to 0.92 respectively) however their control group was higher (0.329Nm/kg).

Peak knee extensor moment was also reported by Manetta, et al. (2002) they did not find any difference between their knee pain and control groups, but they were 0.250Nm/kg for the patient group and less for the controls (0.233Nm/kg). Manetta, et al. reported on the whole of stance phase whereas the result for the present study was only for the first half of stance phase, however this does not really explain the wide variation in results. Interestingly the flexion moment in late stance between the groups in the present study was significant, yet the mean difference was almost identical to the extension moment mean difference, although this was not significant; this would suggest the difference was more systematic, the patient group demonstrating the larger moment.

It was the smaller of the coronal plane moments, the abduction moment during early stance that demonstrated a significant difference; the patient group having the

smaller mean value. Paoloni, et al. (2010) only reported on the abductor moment in the coronal plane though similarly this was split into the loading response and terminal stance. The results they reported were larger than the present study (0.555Nm/kg versus 0.43Nm/kg, CI 0.39 to 0.46). Their experimental group had a significantly larger moment recorded than the controls in early stance. The adduction moment in the present study demonstrated no significant increase and if anything there was a trend towards the patient group having a reduced moment. The differences seen between the studies could be due to the method of recording the data (which will be discussed later Chapter 10.1.5.) or possibly it is highlighting subpopulations.

Both the maximum abduction and adduction moments were marginally greater in the non-symptomatic group during early stance phase. This resulted in the coronal moment range being significantly smaller in the patient group of this study. Considering the results of the coronal plane knee kinematics; excursion was not significantly larger in the patient group. Maximum adduction was significantly increased while maximum abduction was significantly reduced. In theory this could lead to the GRF passing closer medially to the joint centre in the patient group, hence reducing the moment range and may indicate a compensation strategy.

In late stance the control group once again demonstrated a trend towards having a greater adduction moment; this was in contrast to Paoloni, et al. (2010) who reported a trend in the opposite direction. Magnitudes in this phase were much closer between the two studies (0.477Nm/kg versus 0.43Nm/kg, CI 0.39 to 0.46) therefore it is less likely that it is the method which is the difference in this instance. It is more likely it has something to do with foot type which was included in the present study; Paoloni, et al. was not interested in foot type or function.

The present study did not find any differences in the transverse moments of the knee; they were reported for the whole stance phase. Paoloni, et al. (2010) looked at loading and terminal stance and did report an increased moment in the symptomatic group for the initial phase. The magnitude of mean internal rotation moment in this study (0.17Nm/kg, CI 0.16 to 0.19) was much larger than the external rotator moment published by Paoloni, et al. (0.071Nm/kg). It was stated that they found an

internal rotator moment in late stance of equal magnitude, whereas the mean transverse rotation moment in this study remained internal. Again this could be as a result of the excessively pronated foot type included in this study.

9.5.2. Comparison of knee moments non-symptomatic and symptomatic subjects - step descent

The present study found that during the forward continuum phase of step descent the maximum flexion moment and flexion moment range were very similar between the subject groups; however during the lowering phase the patient group demonstrated significant increases in both parameters. This may be predicted on two counts, it gives a convenient explanation for increased patellofemoral joint stress in the symptomatic cohort and the lowering phase will be the point at which the knee will be under its greatest eccentric load, hence it would be expected any instability would be evident at this point.

The increased flexion moment in this study was in contrast to Salsich, et al. (2001) who reported a reduced peak flexion moment in their patellofemoral pain group compared to their control group (0.50Nm/kg vs 0.78Nm/kg). They hypothesised that this was due to the symptomatic subjects altering their centre of mass by leaning the trunk anteriorly. This may have been a factor in Salsich, et al. study, however, the lack of agreement between the studies could be due to: contrasts in the magnitude of pain (Chapter 10.1.1.) plus they calculated the knee moments with respect to a global coordinate system. In the present study moments were calculated with respect to the distal segment (the consequences of which will be discussed later Chapter 10.1.5.).

Brechter and Powers, (2002) reported a trend towards their patient group demonstrating a lower knee extensor moment (0.98Nm/kg vs. 1.12Nm/kg) although this was just outside the level of significance ($p=0.09$). Both Brechter and Powers, and Salsich, et al. (2001) reported lower results than the present study (1.36Nm/kg, CI 1.25 to 1.47 vs. 1.54Nm/kg, CI 1.33 to 1.56). This is not easily explained due to the step height, as all were within 5mm. Possibly the segment modelling used produced some of the differences, however the shoes worn in the present study may also have had an effect.

During the forward continuum phase the maximum adduction moment and the coronal moment range were significantly increased in the patient group of the present study. The magnitude of the results obtained in this study compared well to Kowalk, et al. (1996) who referenced to a proximal coordinate frame. They recognised that the sagittal plane moments will be a natural focus during stair use. They reflected that the coronal plane moments provided not only medio-lateral stability, but also contributed to propulsion. Yu, et al. (1997) stated that it was the vertical and medio-lateral ground reaction forces that provided the predictors for the valgus knee moment in both walking and descending stairs. Aminaka, et al. (2011) reported no differences in coronal plane moments between their patellofemoral group and their controls, though the peak moments reported seemed exceedingly low (-0.0033Nm/kg) this may be attributed to them using the second step and a different marker system; it was not stated how the moments were referenced.

In the present study there was no difference seen in the maximum knee adduction moment during the lowering phase, but the patient group did have an increased moment range. No comparison can be made with other studies as there have been no attempts made to look at the different phases previously, this result could suggest it is an attempt to try and stabilise the knee while it is under its greatest load.

It is again difficult to compare the rotational moments of the knee with previous studies as there is a preoccupation with the sagittal plane estimating patellofemoral compression. Selfe, et al. (2011) investigated the coronal and transverse kinematics and kinetics suggesting that reducing both the ranges of motion and the range of moments could help with the treatment of patellofemoral pain. It was the mean differences between the interventions they reported. The transverse moment graph published a range of 0.05Nm/kg which was similar to the forward continuum phase of this study, but was smaller than the lowering phase.

In the present study the internal rotation moment dominated in both the forward continuum and lowering phases; the maximum moments were similar between the groups in both phases as was the moment range in the lowering phase. However, in the forward continuum phase the moment range was significantly greater in the

patient group, this can be related to the increased range of motion, or greater instability seen in the kinematics.

9.5.3. Effects of orthoses knee moments - walking

There were no significant differences seen with either of the orthoses for any of the moments measured in the present study. There was a slight trend in the maximum extension moment where there was a non-significant reduction seen in both groups with the $\frac{3}{4}$ orthoses. Possibly more of a surprise was that there was no significant difference seen with the wedged orthoses in the coronal plane, it is a clinically held belief that medially wedging a foot will have the effect of increasing the adduction moment. In the present study the adduction moment remained more or less unaffected during early stance phase, with only a tendency for the maximum adduction moment to be increased by the full length orthoses in the propulsive phase. This was more evident in the control group suggesting the orthoses were having a reduced effect possibly due to their more pronated foot type or due to compensation strategies from a proximal source.

These results are in contrast to Nester, et al. (2003) who reported that the 10° wedge used in their study did increase the adduction moment, however there were no differences reported in the kinematics. It is possible that the steeper wedge had more effect, plus Nester's cohort was all pain free. Nigg, et al. (2003) conducted a study on 15 male runners with four different orthoses, they reported the only significant group change was in the maximum external rotation moment which was increased over the control insole condition. They reported that reaction to the orthoses tended to be subject specific not creating systematic changes in all subjects.

9.5.4. Effects of orthoses knee moments - step descent

The flexion moments seen in both phases of step descent were unaffected by either of the orthoses, which was similar to the walking trials; again this cannot be compared to other step down studies. The maximum knee adduction moment was also unaffected by the orthoses in the forward continuum phase, but the range of moment was significantly reduced by both orthoses in the same phase. As there was nothing to compliment this change in the knee kinematics it has to be assumed that

this was due to an alteration in the GRF passing closer to the knee joint centre, this could be linked to the foot being less everted and possibly more stable with the orthoses. Further studies are required to investigate whether this change is apparent when using individually prescribed orthoses to patellofemoral pain subjects.

There were no further significant changes seen in the coronal plane during the lowering phase though the full length orthoses did demonstrate a mild increase in the maximum adduction moment. This is logical as the GRF moves forwards towards the phalangeal segment and the heel lifts. The wedge under the front of the foot will have more of an effect; this may not be desirable on a subject who already adopts a more adducted position as in Fig.9.1. No significant changes were seen in the transverse plane, though the full length orthoses did produce a non-significant increase in the maximum internal moment of the lowering phase. This cannot be linked to the knee or rearfoot kinematics therefore it must be presumed that this is again related to an alteration in the GRF due to the wedge under the front of the foot.

9.4.5. Summary knee moments

During the walking trials the maximum flexion moment during late stance was increased in the patient group compared to the controls. In the coronal plane the symptomatic subjects demonstrated reduced knee adduction moment and coronal moment range during early stance phase. It was suggested that this change in the coronal moments was a compensation strategy to increase the stability of the knee, the external moments did not reveal any significant differences. The step down trials revealed higher maximum and flexion moment range which was hypothetically linked to greater patellofemoral compression. In the coronal plane maximum adduction moment and coronal moment range were increased in the patient group during the forward continuum phase of step down. This was related to the greater adduction seen in the kinematics, though during the lowering phase only the moment excursion remained significantly increased. Only the rotational moment range during the forward continuum phase was significantly increased in the patellofemoral group. The rotational moments did not demonstrate any differences between the groups in the lowering phase.

The orthoses did not affect the knee moments in any way during the walking trials; however the coronal moment range during the forward continuum phase of step descent was reduced significantly. As there were no kinematic changes seen at the knee with the orthoses this was attributed to an alteration of the GRF being angled closer to the joint centre.

9.6. Ankle Moments during walking and step descent

9.6.1 Comparison of ankle moments non-symptomatic and symptomatic subjects - walking

Both maximum ankle dorsiflexion moment and sagittal moment range were significantly smaller in the patellofemoral pain group over the control group in this study. Maximum ankle dorsiflexion moment was larger than the plantarflexion moment and was similar in magnitude to the value published by Nadeau, et al. (1997). They reported no differences between their groups, however it must be highlighted that their symptomatic group had knee osteoarthritis rather than patellofemoral pain. Robon, et al. (2000) also used subjects with knee arthritis and controls; they reported that the group with pain did demonstrate a significant reduction in ankle plantar flexor moment.

Rouhani, et al. (2011) noted that it was only the sagittal plane moment that demonstrated any consistency, both the transverse and especially the coronal plane were less consistent between both studies and subjects. This may be one reason for few studies publishing inverting and everting moments, certainly there was variation seen between subjects in both groups of the present study as is illustrated by the error bars of the coronal plane bar charts Fig. 8.31. and Fig.8.32. in Chapter 8.

The maximum everting moment was significantly reduced in the patellofemoral pain group; this moment was prevalent during early stance phase when the coronal plane knee moment was shifting from maximum abduction to adduction. There was a significant reduction in both maximum abduction and coronal moment range at the knee seen in the symptomatic patients, though this is interesting in isolation, this study is unable to decipher if the ankle is controlling the knee or vice versa.

In the transverse plane there were no differences seen between the two groups of subjects, the internally rotating moment was dominant from before half way through stance phase, this was in agreement with Bellchamber and van den Bogert, (2000) who suggested the internal rotating moment was evident for at least 70% of stance.

9.6.2. Comparison of ankle moments non-symptomatic and symptomatic subjects - step down

When both sets of subjects were descending in the shoe only condition the peak dorsiflexion moment was equal between the groups. This was lower than the mean quoted by Salsich, et al. (2001) though this could be due to the fact the ankle moment in the present study was referenced to the calcaneal segment rather than the whole foot, equally they may have used a global reference frame, this was not made obvious (the effects of this will be discussed later Chapter 10.1.6.i.). Salsich, et al. reported a non-significant decrease in their patellofemoral group, they suggested that the reduced knee moment they obtained in their symptomatic cohort was not compensated for at the ankle.

The peak dorsiflexion moment obtained in this study (-1.17Nm/kg, CI -1.21 to -1.12) was also less than the 1.38Nm/kg obtained by Protopapadaki, et al. (2007) although they calculated the moments with the link-segment method. Again this was using a single segment foot model and the subjects were barefoot which may have had some bearing on the results. Bruening, et al. (2011) noted that sagittal plane motion and peak power generation are overestimated by using a single segment foot model. Care must be taken not to draw any conclusions from the significant result in the maximum mean moment obtained in this study; it would appear that it was the orthoses that were having the greatest effect on the non-symptomatic group. When the looking at the moment range in the sagittal plane no difference or trends were seen in the data.

Most studies dealing with step down have tended to avoid reporting moments in the coronal plane. Cluff and Robertson, (2011) reviewed 17 studies investigating stair descent none of which reported coronal moments. In the present study during the forward continuum phase the patient group demonstrated significantly increased

peak inverting moment and coronal moment range. This may relate to the increased eversion seen in the kinematics. Aliberti, et al. (2010) reported a group of patellofemoral pain subjects demonstrated a significantly greater medial contact area at the medial rearfoot and midfoot over a group of matched controls, this may relate to the increased eversion in this work. In the present study the lowering phase did not highlight any significant differences between the groups, however there was a trend towards the moment range being a little less in the patient group.

In the transverse plane the results were converse, in that the forward continuum phase did not demonstrate any differences between the groups. However in the lowering phase there was a trend for the peak external moment to be to be greater in the normal cohort. The rotational moment range was significantly greater in the patient group; a similar result was also seen in the rotational excursion of the calcaneal segment with respect to the shank. Again care must be taken not to make sweeping judgements from this result as when looking at the shoe only condition between the two groups the difference was minimal, it was mainly the increases seen with the orthoses that produced the significant result.

9.6.3. Effects of orthoses ankle moments - walking

Similar to the knee there were no significant changes seen in the ankle moments with either of the orthoses used in this study. Nigg, et al. (2003) reported maximum inversion moments were significantly smaller with the full length medial insert compared to a flat control insole, although this was for 15 subjects running, they did not report any other changes at the ankle. Nester, et al. (2003) reported that medially wedged orthoses increased the first peak external rotation moment significantly; they also noted that the adduction moment of the rearfoot complex was increased however this did not appear to be a significant change. Nigg, et al. and Nester, et al. used healthy pain free subjects and did not report on foot type, making comparisons with this study difficult.

In the present study the full length orthoses did demonstrate a trend towards reducing the maximum everting moment and the maximum external rotating moment. This demonstrates there were some similarities with Nester, et al. (2003) however; the

lack of significance could be due to the moments being referenced to the distal segment in this study, rather than the global system used by Nester, et al.

9.6.4. Effects of orthoses ankle moments - step down

The maximum ankle dorsiflexion moment was significantly increased by both orthoses; this would appear to relate to the non-significant increases in maximum dorsiflexion seen in the kinematics, however the significant increase in calcaneal segment excursion was not repeated in the moment range. Even though dorsiflexion range was increased the GRF may remain a similar distance from the joint centre leading to a lack of change in the moments.

In the forward continuum phase the coronal moment range did show a weak trend towards reduction with both orthoses and even though not significant may add evidence to support the theory that the significant reduction seen at the knee in the coronal plane is likely to be due to a change in the GRF. There were no further trends to report in the lowering phase which was again similar to the knee. The transverse moments did not relate to the kinematics of the calcaneal segment, where there were no differences to note. The maximum external moment in the forward continuum phase was not significantly decreased by either orthoses but they did demonstrate trends, this was more evident with the full length wedges. The maximum ankle external moment also demonstrated the similar trend in the lowering phase, but this time there was a significant decrease being produced by the $\frac{3}{4}$ orthosis, however examination of the data shows this was mainly due to the reduction seen in the control group. Although this is a significant result the lack of changes in the knee moments and the main influence being seen in the control group must question the clinical relevance of this result.

9.6.5. Summary ankle moments.

The present study revealed the patient group demonstrated decreased dorsiflexion moment and sagittal plane moment range over the control group and this is more than likely related to the reduced amount of dorsiflexion reported in the rearfoot kinematics. The everting moment was similarly reduced, but this could not be

mapped to the coronal kinematics of the foot; all the rotational moments recorded were similar between the groups. During step down the dorsiflexion moment mirrored the walking trials in that it was significantly lower in the patient group. However in the coronal plane the inversion moment and the coronal moment range were increased in the patient group, this was attributed to the patients attempting to stabilise their knees in greater adduction.

There were no significant changes in the ankle moments induced by the orthoses when walking. During step descent the maximum dorsiflexion moment was increased by both orthoses while the $\frac{3}{4}$ orthoses reduced the transverse moment range during the lowering phase.

Chapter 10. Study limitations and conclusion

10.1. Study Limitations and further considerations

10.1.1. Lack of pain during data collection

During the walking trials in this study only 2 of the 15 patellofemoral pain subjects complained of any pain. During the step down trials this increased to 4 out of the cohort, however only one of the 4 complained of pain 3/10 on a numeric pain rating scale; one subject only complained of pain on their last set of step down trials and suggested that this was due to the number of step downs rather than any intervention. Due to the lack of pain questions must be raised, were the patient group using any pain avoidance compensation mechanisms, or were there changes in the data linked to causative factors. The literature states that the incidence of patellofemoral pain is more prevalent in the female population; however more male subjects volunteered in the present study.

10.1.2. Using a static measure as a parameter for inclusion criteria

The foot posture index (Redmond, et al. 2006) provides a simple and effective solution to defining the position the foot adopts in static stance. It is the only validated method of defining foot position without the use of specialist equipment. The scores given for positional observations, “level” the investigators opinions. However clinically it is recognised that just because a patient stands with a certain foot position it may not predict how the foot functions during dynamic tasks. A dynamic assessment tool is needed which reflects these attributes of the FPI; however the wide variation in individuals gait makes this an almost impossible task. The development of such an assessment tool could lead to the unlocking of the sub-populations associated with this disorder.

Neilsen, et al. (2010) considered the use of a static measure to recruit subjects in relation to midfoot function, they concluded that the FPI does demonstrate significant relationships with dynamic motion, but it is not possible to predict individual movement based on this measure. The video based system they

recommended would be impractical for most subject recruitment due to the time and costs involved. Barton, et al. (2011) considered foot function and FPI on a patellofemoral pain group compared to a control group. They reported significant correlations of forefoot abduction and rearfoot eversion in relation to the laboratory axis. However no correlation was found between FPI and rearfoot eversion when referenced to the tibia, this may be corroborated by the present study with there being no significant differences between the groups in the coronal plane.

10.1.3. Standard wedge under orthoses

The customisation of the orthoses in the present study was limited to moulding the devices to the arch profile, beyond this a standard 5° wedge was used either extending to the metatarsal heads (3/4 orthoses) or to the end of the insole (FL orthoses). Although this helps to standardize the research protocol this may not be the optimum prescription for an individual subject, this work did measure significant biomechanical changes in foot function. However further work is required to investigate different levels of orthosis prescription customisation to symptomatic subjects.

10.1.4. Using initial step

All subjects in this work were recorded as they stepped off a higher step onto a lower one. Cluff, et al. (2011) pointed out that this method would not lead to a subject reaching their steady state (Chapter 2, section 2.5.1.). Although Cluff, et al. focussed on the impact of the swing limb which may also affect the results of the stance limb, further studies are needed to investigate if orthoses have any other effects at this “steady” rate of descent.

10.1.5. Research methods

10.1.5.1. Kinematic data

When comparing data between studies it was highlighted previously that sometimes the methods used in data collection can affect the reported results. Many authors use a Helen Hayes marker set which was developed in 1990 by Kabada, et al. a similar marker set was published by Davis, et al. (1991) which is illustrated in Fig. 10.1.

Levine, et al. (2012) stated that the tibial and femoral wand markers allowed rotations of the thigh and tibial segments to be measured for the first time. The thigh and shank are modelled as equilateral triangles and the knee joint centre is estimated as half the knee width in alignment from a single lateral marker. In the CAST proposed by Cappozzo, et al. (1995) the static marker set (see Fig.10.1.) has markers on either side of the knee. Levine, et al. suggested this allows the joint centre to be estimated more accurately; which could have a significant influence on the coronal and transverse plane measurements. This may be a reason why the present study detected small significant changes with some of the comparisons made. The present study also focussed on range of motion which illustrated many of the changes between the groups, rather than just looking at maximum positions attained during a movement task (Wilken, et al. 2012).

It has been highlighted that the data in the second study presented in this thesis was recorded shod (Chapter 8). It could be argued makes it more realistic as most patients will wear their orthoses in normal footwear. However the effect of heel height, support of the heel counter and slippage within the shoe will all have effects on the results, Bishop, et al. (2013) described this as the foot-shoe complex. Therefore care must be taken when comparing the results with barefoot studies and studies where holes cut in the shoes may reduce the shoes effect. Similarly studies using single segment foot models provide poor comparisons with the present study, but the majority of information in the literature used this form of protocol as multi-segment foot modelling is still under development with little agreement on the best model.

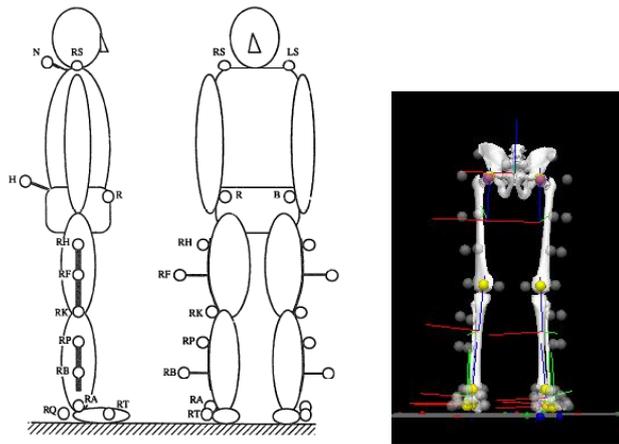


Fig. 10.1. Marker sets from Davis (1991) and the CAST static trial from the present study.

10.1.5.2. Kinetic data

In the present study all the moments presented were referenced to the distal segment. Liu and Lockhart, (2006) compared moments measured from local and global coordinate systems they found that how the moments were referenced did have a significant effect on the results; the global results tended to underestimate the coronal plane results particularly at the ankle. It was hypothesised that the locally expressed moments were more “meaningful”.

Another confounding problem with the ankle joint moments in the comparison with other studies is the use of a multi-segment foot model. The ankle joint moments were referenced to the calcaneal segment in this work; hence the moment distance will be calculated from the centre of the segment to the ankle joint centre. Most previous studies used a single segment foot, so even if the authors have calculated their moments from a local coordinate frame it will have the effect of increasing the distance from the ankle joint centre (see Fig 10.2.).

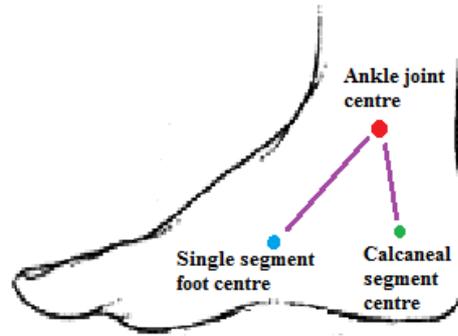


Fig.10.2. Diagrammatic representation of calculation of ankle joint moments using single segment foot and multi-segment foot models

10.3. Conclusions

In conclusion this thesis was structured in two parts, the primary focus of the initial study was to investigate whether using the calibrated anatomical system technique (CAST) to split the foot into 3 segments presented the ability to detect small changes in foot function, even when the foot was shod in normal every day shoes, and using two different foot orthoses. This was achieved by comparing these results with data recorded with specially developed sandals allowing fixation of the markers onto the skin and fitment of the two types of orthoses. The shod data for the lower limb was then used as the control to compare the knee and foot mechanics to a second cohort of subjects who had patellofemoral pain in the second part of the study.

During the walking trials the shoes increased both sagittal and transverse plane excursion of the rearfoot compared to the sandal condition, while the coronal plane remained unchanged. The introduction of the orthoses tended to reduce the range of eversion. At the midfoot the shoes demonstrated the greatest reductions in all planes compared to the sandals and though this could not be definitely attributed to support or slippage the orthoses did reduce the range significantly during the propulsive phase which was as expected. All ankle joint moments were significantly affected by the shoes compared to the sandals barring dorsiflexion moment, however these changes tended not to be reflected at the knee. These results suggested that the techniques employed were sensitive enough to pick up small changes in foot mechanics, when using the subject's normal footwear.

During step down only the midfoot repeated the significant changes in all planes seen in the walking trials, the rearfoot and knee mechanics were unaffected by the footwear, or the orthoses. The transverse ankle moments were all affected by the shoes over the sandals, while both orthoses only decreased the maximum external moment. The full length orthoses also managed to increase the minimum inverting and the maximum internal moment. These initial step trials lead to the supposition it would be beneficial to look at the forward continuum and lowering phases as separate entities.

When comparing the walking foot kinematics of the patellofemoral group and the control group the symptomatic subjects demonstrated a greater range of dorsiflexion, while the metatarsal segment also highlighted a greater range in the propulsive phase. The knee revealed significant differences at maximum extension, maximum adduction, maximum abduction, maximum external rotation and rotation excursion during early stance, some of these differences were repeated in the kinetics. This may question the belief that it is only during increased demand that differences can be ascertained, however this work cannot establish if the changes in gait are due to cause or effect. Further investigations are necessary to establish why these effects are seen and if different sub groups can be defined from similar data.

Appendix 1.

Information Sheet and Consent Form

Title of Study:

Pilot Study: To Demonstrate the Possibility of Using Multi-Segment foot Model in Footwear With and Without Orthoses when Walking Over Level Ground and On Stair Descent

Aim of Study:

This study will look at the effects of insoles on the foot and knee mechanics, when walking and descending steps. This study will find the best method to look at the effect orthoses on the arches of the foot.

What you will be asked to do:

Reflective markers will be attached to your feet and legs with double sided tape; this allows the movement of your lower limbs to be accurately measured. You will then be asked to walk down the laboratory with one of the footwear conditions, then you will be asked to ascend and descend a set of three steps, this will take place in a controlled manor so any possibility of tripping or falling is minimised. In the unlikely event that you experience any pain or discomfort you will be advised to stop the trial immediately and you will be withdrawn from the study, the whole process will take approximately 90 minutes.

You will be asked to complete the trials barefooted, wearing a specially constructed sandal with and without orthoses, and some of your own flexible shoes with and without orthoses.

Foot orthoses would only normally be worn by subjects who have pain, due to the nature of the prescription this should not exceed the average range of movement or forces that are experienced by most people in everyday living; therefore wearing the orthoses for a short period of time should create no undesirable effects on the joints or muscles of the lower limbs.

How will the information be used?

The data we collect may be used for research publications and conference presentations. All data will be coded and no names will be associated with any of the data presented. Data will be stored for 5 years after the completion of the study after which time it will be deleted and destroyed

To be included in these studies please consider and answer the questions below:

	Initial Box	Yes
No		
Do you have any pain or problems with your lower limbs or back?	<input type="checkbox"/>	<input type="checkbox"/>
Have you ever had knee pain?	<input type="checkbox"/>	<input type="checkbox"/>
Do you wear foot orthoses regularly?	<input type="checkbox"/>	<input type="checkbox"/>
If you have answered yes to any of the questions above unfortunately you will not be able to take part in the study.		
Have you read the information above and do you understand it?	<input type="checkbox"/>	<input type="checkbox"/>

11th February 2009
Jim Richards/John Burston/James Selfe
PHACS
University of Central Lancashire
Dear Jim, John and James

Re: Faculty of Health Ethics Committee (FHEC) Application - (Proposal No.321)

Following review of your proposal 'Study to demonstrate the possibility of using multi-segment foot model in footwear With and Without orthoses when walking over level ground and on stair descent', the FHEC has requested that the attached *conditions* be addressed prior to further consideration of the approval of the project. If *recommendations* are also listed, the FHEC would prefer that they are addressed, but approval would not be withheld should you decide not to address one or more of these recommendations.

In your response to FHEC, please ensure that, in addition to including updated documentation (including a new application form) you complete the attached grid, indicating:

- how you have responded to the conditions
- whether you have adopted any of the recommendations, and, if so, how you have addressed these.

Please do not resubmit documentation which you have *not* amended.

Please number your documentation submitted to address these conditions (and recommendations, if appropriate) as Version (Number).3

Yours sincerely

Damien McElvenny
Chair
Faculty of Health Ethics Committee

Response to FHEC Application - Proposal No 321 (Version No.3)

Condition	Applicant Response
1. Full contact details should be given for the lead researcher, not just an email address	Added
2. What happens to the study data after 5 years needs explanation.	Included: Data will be stored for 5 years after the completion of the study after which time it will be deleted and destroyed
3. By explaining what is meant by "normal movement", the committee simply wanted it making clear whether the term "normal" was applied to the individual in the study or to the general population?	Sentence reworded" due to the nature of the prescription this should not exceed the average range of movement or forces that are experienced by most people in everyday living;"
4. The term "adverse effects" needs explaining.	Sentence reworded to: therefore wearing the orthoses for a short period of time should create no undesirable effects on the joints or muscles of the lower limbs.
5 Will a subject be withdrawn from the study if they feel discomfort rather than pain? This has not been addressed from the original conditions.	Sentence altered to include discomfort

The kinematic and Kinetic effects of three quarter and full length foot orthoses on subjects with anterior knee pain when walking and descending stairs

John Burston

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Yorkshire*

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University of Central Lancashire, Preston, UK*

Protocol (version 1)

17 May 2011

Participants:

All participants will have been diagnosed as having anterior knee pain and will not have had any previous orthopaedic interventions or a limb length discrepancy. All subjects should have no other lower limb or back pain at the time of testing.

Design:

A within subjects repeated measures design will be used to analyse the changes in gait under the various testing conditions. A Generalized Linear Model repeated measures ANOVA with post hoc pairwise comparisons with Least Squares Difference will be used to determine the changes in movement analysis between the different conditions.

Methods:

The proposed tests will make use of a highly accurate motion analysis system (Qualisys medical, Sweden) in conjunction with force platform analysis to allow for inverse dynamic analysis to calculate joint kinetics and joint kinematics.

The testing session will be broken down in to 3 distinct sections: Marking up and collection of subject data, gait tests with and without orthoses in shoes and sandals and step descents with the differing footwear conditions. Before any work commences the subject will be required to complete an informed consent form..

Marking up and collection of subject data

This stage involves the attachment of 14mm diameter retro-reflective markers to anatomical landmarks of the legs and 10mm markers attached to the feet to define

the position and orientation in space of limbs and associated joint centres. The markers attached to the anatomical landmarks are done so using dermatological friendly double sided adhesive tape. Once the anatomical frame markers are attached small thermoplastic plates are attached and strapped to the femoral and tibial segments, these plates have four, 14mm diameter retro reflective markers attached. These clusters are used to define a rigid body with 6 degrees of freedom and are referred to as tracking markers, Figure 1. This allows for rotational and translational analysis. At this point the calibration frame is collected, which requires the subject to stand in the centre of the motion capture area for approximately 2 seconds whilst a 3 dimensional image is constructed. Following this stage the subject's height and weight will be recorded. All data of this type will be encrypted on a non networked PC.

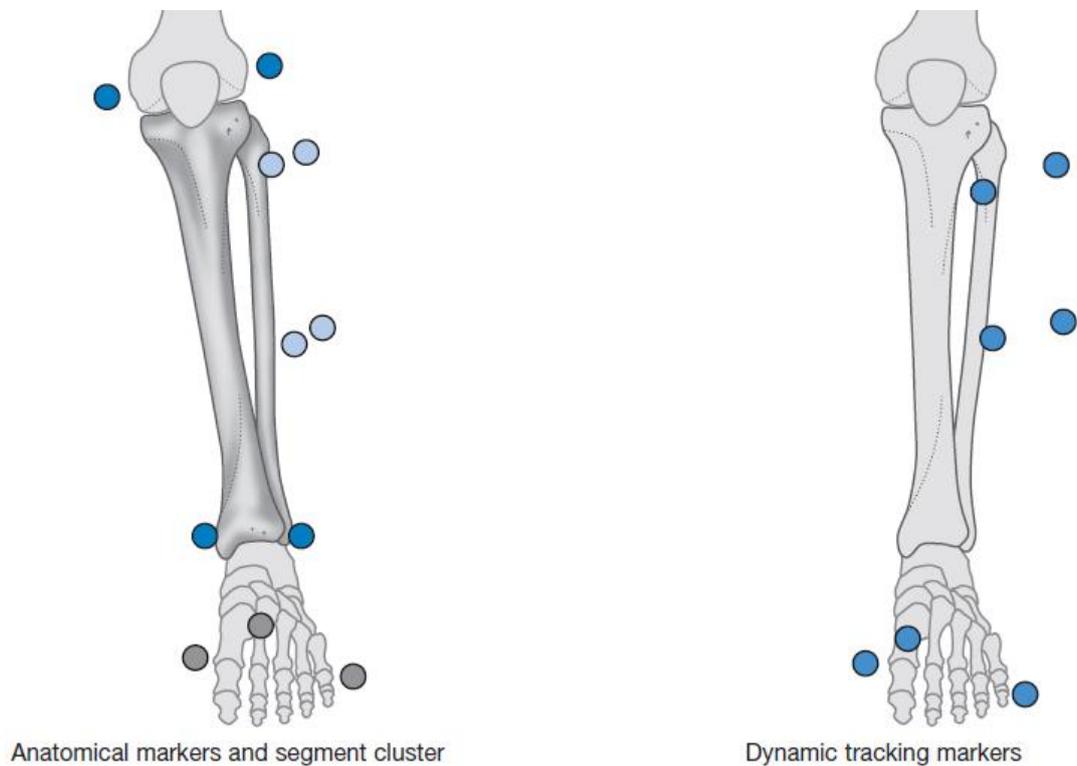


Figure 1: Anatomical and Tracking Markers

Assessment

Each subject will perform 5 walks of 20 metres and descend a 20cm step three times under the following randomised conditions: Sandals only, shoes only, sandals with $\frac{3}{4}$ orthoses, shoe with $\frac{3}{4}$ orthoses, sandals with full length orthoses, and shoes with full length orthoses. The sagittal, coronal and transverse plane mechanics of the knee and

foot will be investigated concurrently. This will allow an assessment of the effect of the orthoses and the footwear on the lower limb.

Modelling:

The phalangeal, metatarsal and calcaneal segments of the foot, shank (tibia), thigh and pelvic segments will be modelled in six degrees of freedom using the Calibrated Anatomical System Technique (CAST). On the foot similar markers sets will be used directly on the subject's skin with the sandals or over the same landmarks on the shoe this will lead to only the range of motion being reported. Figure 2..

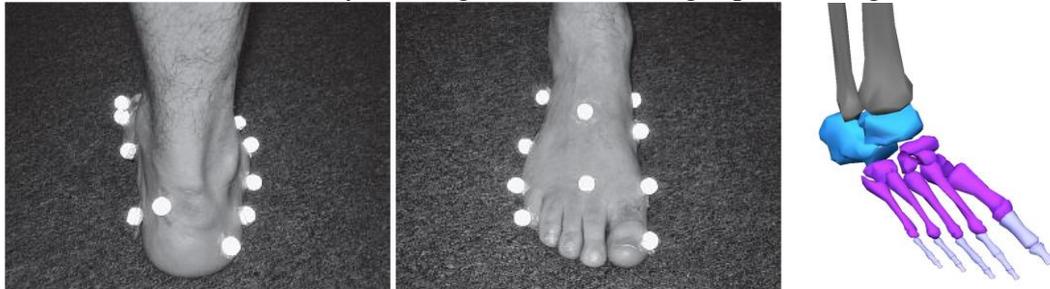


Figure 2: Model of phalangeal, metatarsal and calcaneal segments of the foot

Analysis:

Analysis of kinematic and kinetic variables will be done using Visual 3D (C-motion Inc, USA) and Microsoft excel. All statistical analysis will be carried out in SPSS.



Version 2 – Tuesday 13 September 2011

Dear Sir/Madam

You have been diagnosed as having anterior knee pain (pain around the knee cap during or after exercise) as such you fall in to a patient group whom the University of Central Lancashire are particularly interested in for research. The research involves gait analysis, the measurement of how you walk, balance and descend steps; this will ultimately demonstrate how effective orthoses treatment (shoe inserts) are for this condition.

If you are willing to participate in this research it may have a beneficial role in your rehabilitation by providing you with a pair of foot orthoses, and allowing the therapy team to better understand the mechanics associated with your knee.

If you are willing to participate in these tests, you will be required to attend two sessions one of around 30 minutes to be assessed by the researcher and to have some insoles moulded to your feet then a second a week or two later to complete the walking and step descent trials (90 minutes). If you wish to take part you can either call the researcher on 01226 794922 or email: jburston@uclan.ac.uk.

At this point it is conventional to explain the risks associated with the research; however these will be minimal as the activities are daily tasks such as walking and descending steps. However with all research there is an associated risk, we feel that we have done all that is possible to minimise this risk. Added to this ethical approval has been granted by both the University of Central Lancashire and NRES Committee North West – Greater Manchester South.

Regards

John Burston
Lead Podiatrist Barnsley PCT
University of Central Lancashire



National Research Ethics Service

NRES Committee North West - Greater Manchester South

Northwest Centre for Research Ethics Committees
3rd Floor - Barlow House
4 Minshull Street
Manchester
M1 3DZ

Telephone: 0161 625 7830
Facsimile:

14 October 2011

Mr John Burston
Lead Podiatrist Biomechanics
Barnsley PCT
New Street Clinic, New street,
Barnsley, South Yorkshire
S70 1LP

Dear Mr Burston

Study title: The kinematic and kinetic effects of three quarter and full length foot orthoses on subjects with anterior knee pain when walking and descending stairs
REC reference: 11/NW/0362

Thank you for your letter of 22nd September, responding to the Committee's request for further information on the above research and submitting revised documentation.

The further information has been considered on behalf of the Committee by the Chair.

Confirmation of ethical opinion

On behalf of the Committee, I am pleased to confirm a favourable ethical opinion for the above research on the basis described in the application form, protocol and supporting documentation as revised, subject to the conditions specified below.

Ethical review of research sites

NHS sites

The favourable opinion applies to all NHS sites taking part in the study, subject to management permission being obtained from the NHS/HSC R&D office prior to the start of the study (see "Conditions of the favourable opinion" below).

Non-NHS sites

The Committee has not yet been notified of the outcome of any site-specific assessment (SSA) for the non-NHS research site(s) taking part in this study. The favourable opinion does not therefore apply to any non-NHS site at present. We will write to you again as soon as one Research Ethics Committee has notified the outcome of a SSA. In the meantime no study procedures should be initiated at non-NHS sites.

Conditions of the favourable opinion

The favourable opinion is subject to the following conditions being met prior to the start of

This Research Ethics Committee is an advisory committee to the North West Strategic Health Authority
The National Research Ethics Service (NRES) represents the NRES Directorate within
the National Patient Safety Agency and Research Ethics Committees in England

the study.

Management permission or approval must be obtained from each host organisation prior to the start of the study at the site concerned.

Management permission ("R&D approval") should be sought from all NHS organisations involved in the study in accordance with NHS research governance arrangements.

Guidance on applying for NHS permission for research is available in the Integrated Research Application System or at <http://www.rdforum.nhs.uk>.

Where a NHS organisation's role in the study is limited to identifying and referring potential participants to research sites ("participant identification centre"), guidance should be sought from the R&D office on the information it requires to give permission for this activity.

For non-NHS sites, site management permission should be obtained in accordance with the procedures of the relevant host organisation.

Sponsors are not required to notify the Committee of approvals from host organisations

Notice of no objection must be obtained from the Medicines and Healthcare products Regulatory Agency (MHRA).

The sponsor is asked to provide the Committee with a copy of the notice from the MHRA, either confirming no objection or giving grounds for objection, as soon as this is available.

It is the responsibility of the sponsor to ensure that all the conditions are complied with before the start of the study or its initiation at a particular site (as applicable).

Approved documents

The final list of documents reviewed and approved by the Committee is as follows:

Document	Version	Date
Advertisement	2	
Covering Letter		
Evidence of insurance or indemnity		02 August 2010
Investigator CV		17 May 2011
Letter of invitation to participant	1	17 May 2011
Letter of invitation to participant	2	13 September 2011
Other: CV for James Selfe		
Other: CV for Jim Richards		
Participant Consent Form	1	17 May 2011
Participant Consent Form	2	
Participant Information Sheet	1	17 May 2011
Participant Information Sheet	2	
Protocol	1	17 May 2011
REC application	31	18 May 2011
Response to Request for Further Information	1	

Statement of compliance

The Committee is constituted in accordance with the Governance Arrangements for Research Ethics Committees (July 2001) and complies fully with the Standard Operating Procedures for Research Ethics Committees in the UK.

After ethical review

Reporting requirements

The attached document "*After ethical review – guidance for researchers*" gives detailed guidance on reporting requirements for studies with a favourable opinion, including:

- Notifying substantial amendments
- Adding new sites and investigators
- Notification of serious breaches of the protocol
- Progress and safety reports
- Notifying the end of the study

The NRES website also provides guidance on these topics, which is updated in the light of changes in reporting requirements or procedures.

Feedback

You are invited to give your view of the service that you have received from the National Research Ethics Service and the application procedure. If you wish to make your views known please use the feedback form available on the website.

Further information is available at National Research Ethics Service website > After Review

11/NW/0362	Please quote this number on all correspondence
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With the Committee's best wishes for the success of this project

Yours sincerely

PP 

Dr Ann Wakefield
Chair

Email: rowen.callaghan@northwest.nhs.uk

Enclosures: "After ethical review – guidance for researchers"

Copy to: Prof Jim Richards

19th June 2012

John Burston
School of Sports Tourism and the Outdoors
University of Central Lancashire

Dear John

Re: BuSH Ethics Committee Application
Unique reference Number: BuSH 082

The BuSH ethics committee has granted approval of your proposal application ‘The kinematic effects of three quarter and full length foot orthoses on anterior knee pain sufferers when walking and descending stairs.’

Please note that approval is granted up to the end of project date or for 5 years, whichever is the longer. This is on the assumption that the project does not significantly change in which case, you should check whether further ethical clearance is required.

We shall e-mail you a copy of the end-of-project report form to complete within a month of the anticipated date of project completion you specified on your application form. This should be completed, within 3 months, to complete the ethics governance procedures or, alternatively, an amended end-of-project date forwarded to roffice@uclan.ac.uk together with reason for the extension.

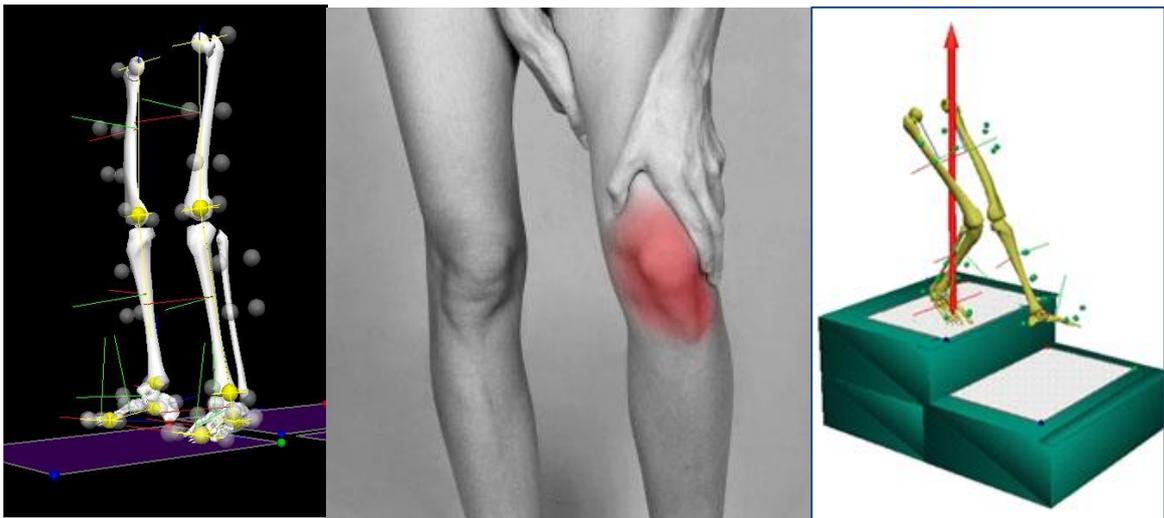
Please also note that it is the responsibility of the applicant to ensure that the ethics committee that has already approved this application is either run under the auspices of the National Research Ethics Service or is a fully constituted ethics committee, including at least one member independent of the organisation or professional group.

Yours sincerely

Denise Forshaw
Chair
BuSH Ethics Committee

Do you have knee pain after descending the stairs or sitting?

Are you under 30 years old?



We are looking for volunteers to take part in a research project which is looking at the effect different insoles have on knee function when walking and descending steps

**If you are interested please contact John Burston
tel. 01226 794922 or
email. j.burston@uclan.ac.uk**

Mail j.burston@uclan.ac.uk tel. 01226 794922

Appendix 2.

Table 1 Calcaneal segment with respect to tibial segment coronal plane range 0-50% stance phase

Pairwise Comparisons ^a							
(I) Orth	(J) Orth	Mean Difference (I-J)	Std. Error	df	Sig. ^c	95% Confidence Interval for Difference ^c	
						Lower Bound	Upper Bound
No orth	3/4 orth	.820	.496	174	.100	-.158	1.799
	FL orth	1.582*	.496	174	.002	.604	2.560
3/4 orth	No orth	-.820	.496	174	.100	-1.799	.158
	FL orth	.762	.496	174	.126	-.216	1.740
FL orth	No orth	-1.582*	.496	174	.002	-2.560	-.604
	3/4 orth	-.762	.496	174	.126	-1.740	.216

Based on estimated marginal means

Table 2 Metatarsal segment with respect to the calcaneal segment transverse plane range

Pairwise Comparisons ^a							
(I) Orth	(J) Orth	Mean Difference (I-J)	Std. Error	df	Sig. ^b	95% Confidence Interval for Difference ^b	
						Lower Bound	Upper Bound
No orth	3/4 orth	.456	.325	174	.163	-.187	1.098
	FL orth	.212	.325	174	.515	-.430	.855
3/4 orth	No orth	-.456	.325	174	.163	-1.098	.187
	FL orth	-.243	.325	174	.456	-.885	.399
FL orth	No orth	-.212	.325	174	.515	-.855	.430
	3/4 orth	.243	.325	174	.456	-.399	.885

Based on estimated marginal means

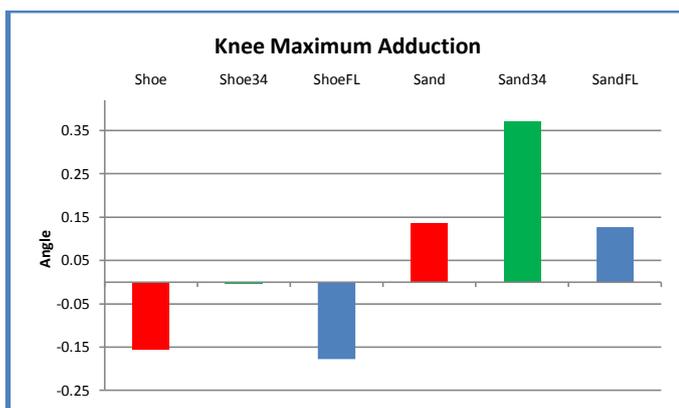


Fig. A2.1. Maximum mean knee Abduction angle bar chart (data spread 0.5°)

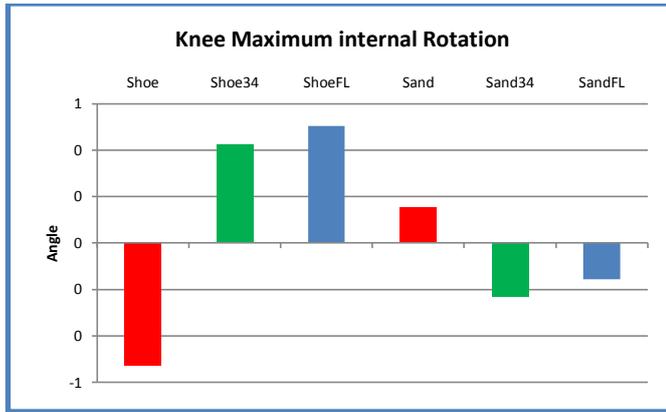


Fig. A2.2. Bar chart showing mean maximum internal rotation (positive values denote an external position) the small values demonstrate the wide variation seen between subjects

Table 3 Knee flexion angle at toe off

Pairwise Comparisons ^a							
(I) Orth	(J) Orth	Mean Difference (I-J)	Std. Error	df	Sig. ^c	95% Confidence Interval for Difference ^c	
						Lower Bound	Upper Bound
No orth	3/4 orth	2.465 [*]	1.067	174	.022	.359	4.571
	FL orth	1.862	1.067	174	.083	-.245	3.968
3/4 orth	No orth	-2.465 [*]	1.067	174	.022	-4.571	-.359
	FL orth	-.603	1.067	174	.573	-2.710	1.503
FL orth	No orth	-1.862	1.067	174	.083	-3.968	.245
	3/4 orth	.603	1.067	174	.573	-1.503	2.710

Based on estimated marginal means

Table 4 Maximum knee adduction moment

Pairwise Comparisons ^a							
(I) Orth	(J) Orth	Mean Difference (I-J)	Std. Error	df	Sig. ^c	95% Confidence Interval for Difference ^c	
						Lower Bound	Upper Bound
No orth	3/4 orth	-.027	.018	174	.129	-.062	.008
	FL orth	-.041 [*]	.018	174	.022	-.076	-.006
3/4 orth	No orth	.027	.018	174	.129	-.008	.062
	FL orth	-.014	.018	174	.434	-.049	.021
FL orth	No orth	.041 [*]	.018	174	.022	.006	.076
	3/4 orth	.014	.018	174	.434	-.021	.049

Based on estimated marginal means

Table 5 Coronal plane knee moment range

Pairwise Comparisons ^a							
(I) Orth	(J) Orth	Mean Difference (I-J)	Std. Error	df	Sig. ^c	95% Confidence Interval for Difference ^c	
						Lower Bound	Upper Bound
No orth	3/4 orth	-.036	.019	174	.065	-.074	.002
	FL orth	-.053*	.019	174	.007	-.091	-.015
3/4 orth	No orth	.036	.019	174	.065	-.002	.074
	FL orth	-.017	.019	174	.376	-.055	.021
FL orth	No orth	.053*	.019	174	.007	.015	.091
	3/4 orth	.017	.019	174	.376	-.021	.055

Based on estimated marginal means

Table 6 Mean sagittal plane motion of the calcaneal segment

Pairwise Comparisons ^a							
(I) Orth	(J) Orth	Mean Difference (I-J)	Std. Error	df	Sig. ^b	95% Confidence Interval for Difference ^b	
						Lower Bound	Upper Bound
None	34orth	-2.614	1.582	83	.102	-5.759	.532
	FLorth	-2.132	1.596	83	.185	-5.306	1.042
34orth	None	2.614	1.582	83	.102	-.532	5.759
	FLorth	.481	1.596	83	.764	-2.692	3.655
FLorth	None	2.132	1.596	83	.185	-1.042	5.306
	34orth	-.481	1.596	83	.764	-3.655	2.692

Based on estimated marginal means

Table 7 Maximum internal moment of the knee

Pairwise Comparisons ^a							
(I) Orth	(J) Orth	Mean Difference (I-J)	Std. Error	df	Sig. ^b	95% Confidence Interval for Difference ^b	
						Lower Bound	Upper Bound
None	34orth	-.007	.019	83	.700	-.045	.030
	FLorth	-.030	.019	83	.121	-.067	.008
34orth	None	.007	.019	83	.700	-.030	.045
	FLorth	-.022	.019	83	.240	-.060	.015
FLorth	None	.030	.019	83	.121	-.008	.067
	34orth	.022	.019	83	.240	-.015	.060

Based on estimated marginal means

Table 8 Metatarsal segment ROM late stance

Pairwise Comparisons^a

(I) Orth	(J) Orth	Mean Difference (I- J)	Std. Error	df	Sig. ^e	95% Confidence Interval for Difference ^e	
						Lower Bound	Upper Bound
None	3/4	.529 ^{b,c}	.344	128	.126	-.151	1.209
	FL	.746 ^{b,c,*}	.345	128	.032	.064	1.428
3/4	None	-.529 ^{b,c}	.344	128	.126	-1.209	.151
	FL	.217 ^{b,c}	.345	128	.530	-.465	.899
FL	None	-.746 ^{b,c,*}	.345	128	.032	-1.428	-.064
	3/4	-.217 ^{b,c}	.345	128	.530	-.899	.465

Based on estimated marginal means

Table 9 Metatarsal segment ROM early stance

Pairwise Comparisons^a

(I) Orth	(J) Orth	Mean Difference (I- J)	Std. Error	df	Sig. ^d	95% Confidence Interval for Difference ^d	
						Lower Bound	Upper Bound
None	3/4	.175 ^{b,c}	.120	128	.147	-.062	.412
	FL	.223 ^{b,c}	.120	128	.066	-.015	.460
3/4	None	-.175 ^{b,c}	.120	128	.147	-.412	.062
	FL	.048 ^{b,c}	.120	128	.693	-.190	.285
FL	None	-.223 ^{b,c}	.120	128	.066	-.460	.015
	3/4	-.048 ^{b,c}	.120	128	.693	-.285	.190

Based on estimated marginal means

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