The Role of Functional Foot Orthoses on Calcaneal and Tibial Kinematics: A Clinical Perspective Using 3-Dimensional Motion Analysis

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Abstract
In-shoe orthoses are used in the treatment and prevention of lower limb injuries in particular patellofemoral pain associated with subtalar joint pronation. The aim of this thesis was to investigate the precise effects of in-shoe orthotic on the Calcaneus and Tibia. Two-dimension frontal plane kinematics may be used in the clinical setting to determine static and dynamic kinematics of the lower limb and foot. The findings of this thesis highlight several limitations of their use. The adoption of 3-D high speed analyses into the clinical setting is proposed. An investigation into dynamic 3-D trials using a rigid model of a tibia resulted in an overall length measurement accuracy of 0.5 mm. This was considered clinically significant and reached statistical significance (p<0.001). Five healthy volunteers were selected for a repeated measures analysis. Transverse tibial rotation was analysed as subjects walked on a treadmill at a fixed speed of 1.1 m/s (4.1 km/h). Investigations were conducted at the same time each day across a three week period. Statistical analysis was performed using an intraclass correlation coefficient. The findings demonstrated moderate to high repeatability (ICC r=0.786). This was considered clinically acceptable. Thirty two competitive runners with a past history of unilateral knee pain volunteered for participation in the study on rearfoot and tibial kinematics. All subjects had in-shoe orthoses of the same type and design and reported no pain at the time the study was conducted. A unique cluster mount and tibial wand was designed and used to determine 3-D calcaneal frontal plane movement through modified footwear and transverse tibial motion. Statistical analysis was performed using Microsoft Excel™ and SPSS 13.5™. Calcaneal eversion at heel strike was increased in shod and orthoses (p<0.05). Internal tibial rotation was reduced in orthoses with custom orthotics reaching statistical significance over shod and generic orthotics trials (p<0.05). Analysis of subject variability resulted in increased calcaneal eversion and internal tibial rotation variability increases of 0.8° to 12.0° (p<0.05) at heel strike and midstance. The range of variability similarly increased between 1.0° at heel strike and 3.0° at midstance (p<0.05). The unique findings of this thesis provide clinical based evidence the functional role of foot orthoses in the management of lower limb pathomechanics may be more than previously thought. In-shoe orthoses exert angular kinematic changes in rearfoot and tibial motions and increases in rearfoot and tibial variability in walking.
Authors Declaration

I Joseph James Kelly, declare the work contained in the formation of this thesis originate from my own clinical practice. The design, construction and manufacture of bespoke apparatus used in the investigations are the work of my own hands. There were no financial or other commercial interests or help. Mentoring support was provided by my supervisor Dr. A. J. Harrison. Physical Education and Sports Science, Faculty of Health Sciences, University of Limerick.

______________________________
Signed

______________________________
Date
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C: Fabrication Processes
D: Calcaneal & Tibial Angle at Heel strike
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L: 3-D Statistics Chapters 7-8
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N: Ethical Approval and Subject Consent
O: Foot Posture Index
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Q: Images of Pilot Cluster Designs and Data Curves
Chapter 1 THESIS INTRODUCTION

INTRODUCTION
There is evidence that in-shoe orthoses may reduce or prevent movement related injuries, though knowledge about the specific functioning an orthotic provides is limited (Nigg et al., 1999), and there is little knowledge of the differing effects of various orthotic techniques commonly used in podiatry. It has been proposed that abnormal coordination of subtalar joint (STJ) pronation and supination (SUP) combined with flexion and extension of the knee joint may be a major cause of overuse injuries in the lower limb, (Hadley et al., 1999).

This thesis aims to investigate the effect of in-shoe orthoses on dynamic three dimensional (3-D) motions of the rearfoot and tibia. Current examination and treatment of patients with lower limb injury may focus on static assessment and intervention primarily in two dimensional (2-D) frontal and sagittal planes. The outcome of these interventions may be dependent on the patients report following exercise. In clinical practice transverse tibial rotation may not be measured using traditional static assessment, handheld measuring devices or 2-D kinematic techniques. Yet an assumption may be in-shoe devices exert an effect on tibial motion in transverse plane motion and STJ motion with subsequent anecdotal reports from patients of increased, reduced or absence of lower limb pain. STJ inversion eversion range of motion measured at the chairside cannot be objectively translated into dynamic motion nor can the clinician be confident the measurement represents the rearfoot motion within the shod foot. In providing an in-shoe orthotic with a rearfoot wegde observed in the frontal plane primary 2-D measurement of calcaneal angular change may be calculated. One must assume a known rearfoot wedge provided at a static examination exerts an effect at a dynamic level and the effect is not lost withing the soft insole linning within a particular running shoe. Similarly the effect of the medial arch profile of the orthotic device and differing densities used in manufacture may not lend themselves to conventional clinical assessment. Subsequent aims of this thesis were to investigate 2-D and 3-D kinematics of STJ and tibia following the introduction wedged footwear and inshoe orthotic devices. The rearfoot and tibial coupling actions in clinical practice are considered in this work as synonymous with transverse plane knee motion. The investigations were conducted in a podiatry practice were subjects were sampled
from the clinic patient database. All had been in attendance for assessment, treatment and or advice on lower limb injury or prevention. Hence the findings are aimed at clinicians and those involved in the rehabilitation processes involving in-shoe orthotic devices.

Review of the literature concerning foot orthoses reveals that objective evidence to support their prescription is limited. There are many reports of successful outcomes with reported study design and subject limitations. Quantitative studies have produced contradictory results, because of the lack of knowledge regarding the relationship between foot pathology, orthotic treatment and foot biomechanics (Nicolopoulos et al., 1999). Abnormal foot mechanics, in particular excessive STJ pronation increases internal rotation of the tibia. This is termed “tibial coupling” and describes a coupling effect between rearfoot inversion and eversion, and tibial internal and external rotation. It has been the main focus for much of the last decade of research. The muscles and ligaments around the ankle and knee are believed to act as a restraining mechanism preventing abnormal amounts of motion at the joints which may be detrimental to the articular surfaces and the overall functional position of the knee or rearfoot (Besser et al., 1999). Controlling abnormal STJ pronation has been associated with a lower rate of knee injuries reportedly eliminating talar tilt (Sasaki and Yasuda 1987). In summary, there is no clear consensus amongst researchers whether relief of knee pain is due to a reduction in internal tibial rotation, a reduction in ground reaction forces, a shift in force vector or altered alignment of the patellofemoral mechanics in the frontal plane. The exact contribution an external device such as an in-shoe orthoses still remains unclear (Ball and Afheldt 2002; Heiderscheit et al., 2000; Kilmartin and Wallace 1994; Landorf and Keenan 2000; Pratt 2000).

Runners continue to be amongst the most commonly injured athletes (Hreljac et al., 2000). It has been estimated that 27-70% of all runners would sustain an injury in any given year and the knee is the most commonly affected 43-48% (Clement et al., 1981; McIntyre et al., 1991; Taunton et al., 2002). With these injuries evident from patients attending the clinic for treatment this thesis evolved almost organically from the clinical ‘hands on’ nature of the treatment processes and patient management plans. Practicing as a Podiatrist employing techniques accepted in the evidence base
established to date, there were questions concerning the exact contributions of orthoses. These questions were,

1. Do static measurement of the rearfoot and wedgeing translate into dynamic gait?
2. If a rearfoot wedge exerts an effect in dynamic gait does that effect influence transverse tibial motion?
3. How variable is human gait during walking and how reliable are motion analysis systems in the clinical setting?
4. Dominance, individualism, variability in foot morphology and function, do in-shoe orthoses influence lower limb variability within subjects and between groups following intervention?
5. Custom versus preformed orthoses which are better?

1.1 Thesis structure
- Chapter two provides a brief review of anatomy of the ankle, foot and rearfoot complex. The chapter progresses into a retrospective review of the literature to date to establish a baseline of the clinical evidence and aide the reader in the progression of this work through the chapters providing insight into the various studies conducted, the reason for and their design.
- Chapter three details a 2D study investigating dynamic frontal plane rearfoot angles moving forward from current static assessment within the clinic in a group of runners with wedged outsoles. In clinical practice static assessment is performed under the assumption this will predict and carry over into dynamic foot function when the patient leaves the clinical setting.

The functional in-shoe orthotic can be described as having three main components, a wedge component, an arch component and a rigidity or density component. Research often reports combined effect outcomes of all the components. No specific detail concerning individual components may be provided. To determine the exact effect of a wedge aside from the arch component or rigidity of the orthotic shell, this chapter investigated known angular wedges adhered to the outsole of subject’s footwear. Data examined static frontal plane heel valgus and varus angles relative to the posterior calf (Rearfoot angles) and progressed to dynamic motion. The findings while
Chapter I Thesis Introduction

statistically significant had inherent limitation and it was concluded future work would require three dimensional analysis methods.

• Chapter four details a 3D study to establish the validity of 3-D motion analysis within the clinical setting. Having acquired a high speed 3-D system a concern was establishing the validity of the equipment for use within the clinic. Two main concerns as a clinician were, 1.) Data collected would impact on patient treatment. 2.) Can the system measure accurately what it is supposed to measure and could one be confident in the calculated data? To answer these questions a metallic rod was fabricated and analysed as it swung through a custom made calibrated space. The findings of this study allow the user to have knowledge of the system and associated error within their practice so they may judge the findings of subsequent works more objectively.

• Chapter five defines technical conventions adopted by the international society of biomechanics necessary for describing kinematics for human motion and animation analyses.

• Chapter six details a 3-D study to establish repeated measures reliability in a group of “normal” subjects assessed over a three week period. The findings of the study place in perspective the study design setup, relative to the research findings for reliability and repeatability studies. The expansion of the thesis to include this chapter on repeatability was further justified within the clinical context, as use of recorded data via computerisation impacts on patient health. As a clinician one must be confident the data generated is from within the subject or the intervention device and not from the measurement instrument.

• Chapter seven details an investigation focused on 3-D rearfoot and tibial kinematics in a group of competitive runners with a history of lower limb overuse injury six months previous to participation who remain asymptomatic. An insight in the functional aspects of custom made and generic preformed orthoses on rearfoot pronation and tibial internal/external rotation angles and range of motion during stance are provided. More robust than previous studies reported, this chapter presents data generated from custom milled subject specific orthotic devices, eliminating previous handcrafted orthotic fabrication techniques which may reduce internal validity in previous studies. Also custom made cluster mounting devices were designed and fabricated to determine
calcaneal (Calc) movement within the shoe. Progressing further as a subplot subjects were given a preformed generic device. The results present objective data to support use of one device over another.

- Chapter eight details the primary contribution of this thesis and the unique 3-D kinematic findings with respect to the most current theory proposed on dynamic gait variability within subjects. Retrospective literature review demonstrates a common theme of angular and range of motion investigation within subjects which may have been adopted under limited evidence. Recent technique and research points towards multiple sensory input factors in combination with structural and functional processes determining dynamic gait progression. The findings of this thesis propose lower limb variability is inherent within each subject and has disregard for gait being purely cyclical during motion. Using averaged standard deviation data the investigation provides evidence of reduced variability in injury and increased variability with the use of in-shoe orthoses in subjects.

- Chapter nine draws together the conclusions, findings and uniqueness of this thesis and contributions to the evidence base. Suggestions are provided to facilitate the future direction of investigations.
1.2 Thesis abbreviations

Table 1-1 Thesis Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>2-D</td>
<td>Two dimensional</td>
</tr>
<tr>
<td>3-D</td>
<td>Three dimensional</td>
</tr>
<tr>
<td>Calc</td>
<td>Calcaneus, Calcanuem</td>
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<tr>
<td>CI</td>
<td>Confidence interval</td>
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<tr>
<td>ICC</td>
<td>Intraclass correlation coefficient</td>
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<tr>
<td>MRI</td>
<td>Magnetic resonance imaging</td>
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<tr>
<td>MTJ</td>
<td>Midtarsal joint</td>
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<tr>
<td>NCSP</td>
<td>Neutral calcaneal stance position</td>
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<tr>
<td>O/A</td>
<td>Osteoarthritis</td>
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<tr>
<td>PRON</td>
<td>Pronation</td>
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<tr>
<td>QA</td>
<td>Quadriceps angle</td>
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<tr>
<td>RCSP</td>
<td>Relaxed calcaneal stance position</td>
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<tr>
<td>RFA</td>
<td>Rearfoot angle</td>
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<tr>
<td>SD</td>
<td>Standard deviation</td>
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<tr>
<td>STJ</td>
<td>Subtalar joint</td>
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<tr>
<td>STJN</td>
<td>Subtalar joint neutral</td>
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<tr>
<td>SUP</td>
<td>Supination</td>
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<tr>
<td>TM</td>
<td>Treadmill</td>
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<tr>
<td>IR</td>
<td>Infra red</td>
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<tr>
<td>Freq</td>
<td>frequency</td>
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### 1.3 Operational definitions
(Adapted from Richards 2008)

<table>
<thead>
<tr>
<th>Term</th>
<th>Definition</th>
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<tbody>
<tr>
<td><strong>Variability</strong></td>
<td>The extent to which a group of scores varies or is spread out. This is qualified by the descriptive statistic size of the standard deviation.</td>
</tr>
<tr>
<td><strong>Standard Deviation</strong></td>
<td>A measure of dispersion or variability of a group of scores.</td>
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<tr>
<td><strong>Statistical significance</strong></td>
<td>The application of a statistical test, to demonstrate that the obtained result is probably not due to chance but is 'real'.</td>
</tr>
<tr>
<td><strong>Clinical significance</strong></td>
<td>A conclusion an intervention has an effect of practical meaning to patients and the health care provider.</td>
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<tr>
<td><strong>Reliability</strong></td>
<td>The extent to which results are consistent over time and if the results of a test or measurement is reproducible.</td>
</tr>
<tr>
<td><strong>Repeatability</strong></td>
<td>The variation in measurement taken by an instrument on the same item under the same conditions (Bland-Altmann 1986).</td>
</tr>
<tr>
<td><strong>Rearfoot / Subtalar joint</strong></td>
<td>Calcaneus and talus synonymous with the subtalar joint.</td>
</tr>
<tr>
<td><strong>Midfoot / Midtarsal joint</strong></td>
<td>The Navicular, Cuboid and the Cuneiforms.</td>
</tr>
<tr>
<td><strong>The Knee / Tibia</strong></td>
<td>The tibia is synonymous with the knee, the tibia extending from the ankle to the knee joint.</td>
</tr>
<tr>
<td><strong>Heel Strike</strong></td>
<td>The beginning of the stance phase. It is defined as the initial contact of the heel with the ground terminating at foot flat.</td>
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<tr>
<td><strong>Midstance</strong></td>
<td>The period of stance between foot flat and heel lift.</td>
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<tr>
<td><strong>Maximum pronation</strong></td>
<td>The period of stance when the tibia and femur are vertically above the midpoint of the foot, in the sagittal plane.</td>
</tr>
<tr>
<td><strong>Toe off</strong></td>
<td>The point at which the stance phase ends and the swing phase begins.</td>
</tr>
<tr>
<td><strong>Tibial angle</strong></td>
<td>Internal tibial rotation, a motion occurring in the transverse plane.</td>
</tr>
<tr>
<td><strong>Calc angle</strong></td>
<td>Frontal plane inversion and eversion of the calcaneus.</td>
</tr>
<tr>
<td><strong>Accuracy</strong></td>
<td>Accuracy is the degree of conformity of a calculated or measured quantity to some other reference value.</td>
</tr>
<tr>
<td><strong>Precision</strong></td>
<td>Is the mutual agreement among a series of individual measurements or values or results.</td>
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7
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<tr>
<th><strong>Validity</strong></th>
<th>The extent to which a test measures what it is supposed to measure (Joppe 2000).</th>
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<tbody>
<tr>
<td><strong>Noise</strong></td>
<td>Random fluctuations that compromise the deterministic relation between input and output at different levels of analysis in the sensorimotor system.</td>
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Chapter 2 THESIS BACKGROUND LITERATURE

Anatomy of the rearfoot complex
The calcaneus and the talus are the largest of the tarsal bones, responsible for transmitting the stresses of our body weight through the ankle joint mortice to the calcaneus via the STJ. In weightbearing the calcaneus supports the talus with an anterior elevation, or pitch of about 15° from the transverse plane. The elevation ensures the posterior tuberosity of calcaneus is the first point of contact with the ground at heelstrike. The talus forms the centrepiece of the rearfoot complex, linking the three articular entities, the talocrural (ankle joint), talocalcaneal joint (subtalar) and the talonavicular (midtarsal) joint. The talus is securely positioned within the ankle mortice formed by the tibia and fibula (figure 2.1).

![Figure 2.1 Cross section of the rearfoot complex, note the talus located centrally](image)

2.1 Ankle Joint
The ankle joint has an axis approximately 8.0° to the transverse plane and 20-30° to the frontal plane (Michaud 1993). This axis allows for a small amount of talar abduction with ankle dorsiflexion. The ankle joint consists of three bones the tibia, fibula and the talus. The tibial articular surface consists of the tibial plafond (Pilon) and the medial malleolus. Bodyweight is transferred through the tibial plafond to the talus. The plafond is mildly concave and continuous with the medial malleolus. The lateral
malleolus projects further distally than the medial malleolus. Medially, its triangular surface articulates with the talus.

The talus superiorly is covered by cartilage and articulates with the tibial plafond. The surface is continuous medially and laterally covering the talar body and articulating on both sides with the medial and lateral malleoli (figure 2.2).

Figure 2.2 Plain radiograph of the ankle joint mortice with medial and lateral malleoli

2.1.1 Ligaments

The ankle mortice is tightly fixed to the distal ends of the tibia and fibula with the help of the anterior and posterior tibiotalar ligaments and the interosseous tibiofibular ligament (Deltoid ligament). On the medial aspect of the ankle the deltoid ligament, a triangular ligament originating from the medial malleolus and inserting posteriorly on the talus and the calcaneus also anteriorly to the ligaments of the talonavicular joint. On the plantar aspect of the foot, the deltoid ligament blends into the plantar fascia which extends from the calcaneus to provide support for the longitudinal arch. The superficial layer helps to prevent valgus displacement and the deep layer lateral translation. On the posterior aspect of the ankle is a complex consisting of two main ligaments, the posterior tibiotalar and posterior tibiofibular ligaments. The lateral (fibular) collateral ligaments are divided into three groups the posterior talofibular, calcaneofibular and anterior talofibular ligaments. An important structure that maintains ankle stability, the syndesmotic ligaments complex comprises the anterior tibiofibular, posterior tibiotalar, transverse tibiofibular and interosseous ligaments.
Some stability between the distal tibia and fibula is also provided by interosseous membrane.

2.1.2 Tendons
Located at the medial side of the ankle, lie the tendons of the main muscles for inversion, tibialis posterior, flexor hallucis longus and flexor digitorum longus. Posterior to these muscles lie the tendons of the triceps surae tendocalcaneus, (Achilles tendon) and the plantaris. The former is the main plantarflexor of the foot, peroneus longus and peroneus brevis anteriorly are the tendons of dorsiflexors, tibialis anterior, extensor digitorum longus, extensor hallucis longus and peroneus tertius.

Biomechanically load is centered on the midline of the ankle and is divided into two vectors the first is orientated posteriorly towards the calcaneus and the second anteriorly through the neck of the talus, navicular cunieoforms and metatarsals.

Normal range of motion of the ankle joint is between 20.0° of dorsiflexion and 20-30° of plantarflexion. Some researchers have described differing values this includes a maximum range of motion of 62.0° (Hicks 1953). However this is questionable when a person wears a high heeled running shoe (Lieberman et al., 2010).

2.1.3 Musculature
Four groups of muscles act across the ankle joint and produce the motions of,

1. **Dorsiflexion, adduction and supination**: Tibialis anterior and extensor hallucis longus.
2. **Dorsiflexion, abductors and pronation**: Peroneus tertius and extensor digitorum longus
3. **Plantarflexion, abduction and pronation**: Peroneus tertius and peroneus longus
4. **Plantarflexion, adduction and supination**: Tibialis posterior, Flexor digitorum longus, Flexor hallucis longus and triceps surae gastrocnemius and soleus.

2.2 Subtalar Joint
The STJ has been described as having three primary functions, 1) to act as a mobile adapter through STJ pronation allowing to the foot to adapt to the surface being traversed. 2) To act as a torque converter changing transverse plane motion into sagittal plane linear motion. 3) To act as a shock absorber at the moment of heel strike through STJ pronation. The subtalar joint (STJ) is formed by the three facets on the
inferior surface of the talus and the corresponding facets on the superior surface of the calcaneus. Pronation and Supination motion at the STJ is described as triplanar, comprising the motions of abduction, eversion and dorsiflexion (Pronation) and Inversion, adduction and plantarflexion (Supination). The STJ axis lies approximately 42.0° to the transverse plane and 16° to the sagittal plane though this is questionable.

Early works described wooden models to demonstrate the orientation and function of the STJ axis (figure 2.3).

Manter (1941) proposed the STJ is a multifaceted joint with convex articulations and only in a true hinge joint can a single motion range be found. The orientation of the STJ axis allows for significant amounts of eversion and inversion in the frontal plane and abduction, adduction in the transverse plane. Variations in axis alignment are well documented and have been identified as factors affecting tri-planar motions of
pronation and supination, (Manter 1941; Wright et al., 1998; Root et al., 1966; Isman and Inman 1968; Kirby 2001; Van Langelaan 1983; Lundberg and Svensson 1993; Ensberg 1987). The axis of the STJ is clinically significant in that a high axis is thought to be responsible for injury to structures proximal to the STJ and a low axis for injury distal to the STJ. However, clinically convenient to employ a fixed axis for the STJ range of motion, analysis and imaging techniques in particular cadaver studies involving implanted metallic beads have indicated motion occurs around an axis whose position and orientation change during joint motion (Lundberg and Svensson 1993) (figure 2.4).

Figure 2.4 The multiaxial STJ adapted from Kirby, K. (2001) Subtalar Joint Axis location and Rotational Equilibrium Theory of Foot Function. J. Am. Podiatr. Med. Assoc 91 (9)

Lundberg and Svensson (1993) stated any degree of sliding or screw-like motion occurring at the STJ was so small that it was undetectable concluding the STJ was essentially a hinge joint.

Accepted in the literature are the links between STJ pronation and supination and tibial internal and external rotation. The combined movements are termed ‘coupling’ in closed kinetic chain pronation the tibia internally rotates and in supination the tibia
externally rotates. Inman (1969) stated the more cavoid a foot, the greater the inclination of the STJ axis in the sagittal plane and therefore the greater the external and internal tibial rotation is following rearfoot inversion and inversion. This was confirmed by other investigations. (Nawoczenski et al., 1995; Nawoczenski et al., 1998; Nigg et al., 1992)

However aside from in cadaver studies that cannot be linked with dynamic closed chain foot function (Hamil et al., 1989; Mathieson et al., 1999). Non invasive investigations and research has pursued Inman’s research proposal and theory. Kim et al., (1995) reported arch height did not influence eversion and tibial internal tibial coupling ratios in barefoot or shod conditions. Stacoff et al., (2000ab) investigated coupling ratios in vivo with intracortical bone pins to avoid previously reported biases of soft tissue movement between surface mounted marker designs (Reinschmidt et al., 1997). Stacoff et al., (2000a) concluded coupling ratios for calcaneal eversion and tibial internal rotation barefoot at (0.66°) and shod running (0.58°). Further research has reported greater coupling ratios between eversion and tibial internal rotation compared to the ratio between calcaneal inversion and tibial external rotation, Stacoff et al., (2000b).

2.2.1 Neutral position

In order to determine a baseline for collection of data in evaluation of the joints of the body, it is necessary to have a joint reference point about which motion can be measured.

“Reliability of measurement is the hallmark of any scientific endeavour”.

Rothstein (1993)

Root et al., (1977) first proposed the concept of the neutral position of the foot which has been adopted by the majority of clinicians involved with podiatric biomechanics. The neutral theory relies on two principles,

1. The STJ is in neutral (STJN) when the foot is not pronated or supinated and the talus demonstrates bony congruency. This congruent position is midway between maximal rearfoot pronation and rearfoot supination.
2. In subtalar joint neutral position the rearfoot is in perfect alignment. The talus has no malicious forces acting on it. Any force is directed though the head and neck of the talus. There is a harmonic balance between the bony skeletal support structure of the foot and the mechanical stabilisation forces exerted by the musculature, ligaments and tendons acting across the foot and lower limb. In the STJN position the foot has no sagittal, frontal or transverse plane deviation.

In support of this, a clinical technique was developed using the theory that the calcaneus inverts with supination twice as many degrees as it everts with pronation (Root et al., 1977). By calculating the total range of motion of the STJ and dividing this into three equal parts, STJN can be defined as the first part moving from inversion to eversion (figure 2.5).

![Figure 2.5](image)

**Figure 2.5 Location of STJ neutral by palpation using the proposed by Root et al., (1977).**

In practice variation in foot structure from the ideal STJ neutral position in patients presenting with injury is common. Generally the lower limb must be aligned in a particular way if it is to function optimally. The knee axes should be perpendicular to the line of progression allowing flexion and extension in the sagittal plane, (Root et al., 1977). The transverse plane neutral position of the hip joints must not be internally or externally rotated. The leg should have undergone developmental tibial torsion of 5-15° of external rotation. The foot should have a neutral alignment, i.e. a vertical calcaneus and directly under the ankle and forefoot directly anterior to the rearfoot, (Second metatarsal long axis in line with the tibia). On examination congenital or acquired variances are often seen. It has been proposed and accepted that small or subtle variation from the ideal neutral foot is the norm and such individuals may go
through life with no lower limb injury. The greater the variance or deviation from the neutral subtalar joint position the greater the chance of injury. Similarly the greater the stress or demand placed on the structure or structures governing function of the lower limb the more likely one would expect to observe injury. McPoil and Cornwall (1992) stated pain at a particular site depends on the compensatory mechanism or series of compensations, the pain being the site of the weakest compensation link in the kinetic chain. For example, an abducted foot places the STJ in the line of progression and the ankle joint is externally rotated, therefore the resultant progression of gait while walking stresses the STJ in pronation substituting for the lost ankle joint dorsiflexion. A STJN position sees the calcaneus in a vertical position relative to a bisection of the lower leg. At this most central point, the research to date has adopted the proposal of the greatest osseous and structural advantage (Root et al., 1977).

2.3.2 Neutral calcaneal stance position and relaxed calcaneal stance position
Static clinical measurements of the rearfoot are used to determine the dynamic motion of the rearfoot during walking. The technique of employing rearfoot static measurements for dynamic measurement has been questioned (Sajko and Pierrynowski 2005). The use of the arch index, navicular drop and drift are useful predictors of static analysis but not dynamic motion (Cornwall and McPhoil 1994). In Podiatric practice the neutral calcaneal stance position (NCSP) figure 2.6 and the relaxed calcaneal stance position (RCSP) figure 2.7 are major components of the examination of the lower extremity (Wooden 1996).
Described as rearfoot angles (RFA) the RCSP and NCSP are used as a guide for determining treatment and its efficacy (Donnatelli et al., 1999; Genova and Gross, 2000; Sobel et al., 1999). Dynamically the neutral foot during the gait cycle strikes the ground with the calcaneus inverted and immediately after heel strike the foot pronates and adapts to the surface (Root et al., 1977). Closed kinetic chain pronation is a triplanar motion involving abduction eversion and dorsiflexion of the talus. The talus in the pronated position is coupled with internal rotation of the tibia. Pronation persists until 25% of the gait cycle reaching 4-6.0°, persistent pronation beyond 25% of the gait cycle is considered abnormal (Root et al., 1977). After 25% of the gait cycle the foot begins to resupinate. The calcaneus moves from eversion to inversion and the lower leg from internal rotation to external rotation. This function of pronation and supination must
occur in sequence and on time (Root et al., 1977). Researchers have reported the static weightbearing position of the calcaneus during RCSP may predict the capability of the rearfoot to compensate during functional activities (Hunt et al., 2000). If RCSP demonstrates an abnormality for the clinician examining a patient it has been suggested the use of this reference posture value for further dynamic angular calculation may mask the severity of the deviation during walking. The variation between subjects and methods of determining what qualifies for intervention or preventative treatment are subjective. The podiatric clinician depends on their clinical judgement and skills of observation and examination within the majority of clinical settings, placing confidence in the theory of rearfoot alignment and the vertical calcaneal bisection relative to the tibia forming the optimum foundation for the entire lower extremity. A bias or deviation of the calcaneus away from the sagittal plane has been considered as a causative factor in lower limb injury (Nicolopoulos et al., 1999). An assumption of addressing anomaly in the structural alignment of the calc is that it will carry over to dynamic gait. Common theories propose the structural imbalance observed results in forces generated during gait being increased or not readily absorbed (Esalami et al., 2007). Muscles must work harder to maintain an optimum trajectory of foot mechanics during the closed kinetic chain cycle of gait (Nester 1998).

Figure 2.7 The relaxed calcaneal stance position (RCSP) demonstrating calcaneal eversion.
2.3 Pronation.

Pronation has been defined as a,

“Triplanar motion collectively combining the movements of dorsiflexion, abduction and eversion” (Root et al., 1977)

Root et al, (1977) classified over pronation as at least four of the following,

- More than 6° between RCSP and the NCSP.
- Medial bulging of the talar head or midtarsal break, quantified using the navicular drift technique (Menz 1995).
- Lowering of medial longitudinal arch, quantified using the navicular drop technique (Mueller et al., 1993).
- Helbings sign (medial bowing of the Achilles tendon)
- Abduction of the forefoot at the midtarsal joint (MTJ), concavity of the lateral border of foot.
- An apropulsive gait.

Classification of the foot based on the morphology or relative rearfoot position in the clinical setting is common. In a clinical environment STJ range of motion is usually measured by assessing the range of inversion and eversion. The STJN position has been described as the maximum point of congruity and has been criticised for interobserver reliability (Menz 1995; Menz and Keenan 1997). Validity of previous findings and influences of ankle movement coupled with STJ range of motion and subject position are questionable (Ball and Johnson 1993). Required is a baseline and quantification method for the clinician at the chairside examination aside from the complexities of magnetic resonance imaging (MRI) X-ray and 3-D gait analysis.
Despite the conflicting findings of studies three important parameters have been defined during the gait cycle.

1. Maximum pronation angle or the number of degrees the calcaneus moves from varus to valgus (Clarke et al., 1984; Rogers and Leveau 1982; Nigg et al., 1987).

2. The time during the gait cycle for the maximum pronation event to occur (Bates et al., 1979; Smith et al., 1986).

3. The total period of rearfoot pronation during the support phase (Bates et al., 1979; Novick and Kelley 1990).

Limitations are acknowledged in the literature and have been categorised into three areas.

- **Soft Tissue** movement does not reflect underlying osseous structures (Maslen and Ackland 1994).

- **Skin Marking** with felt pen is subject to error therefore using this as a baseline and then moving the calcaneus to another extreme in the range of motion available to make another reference line relative to the first would only compound this (Menz 1995).

- **Boney Congruency** the clinician must be confident with identifying the through palpation of the depressions created anteriorly to the ankle joint either side of the head of the talus. In conjunction with identifying the fifth metatarsal head and applying enough force to dorsiflex and cause the mid tarsal joint to lock and not introducing a forefoot anomaly.

More recent work has questioned STJ neutral rejecting the 2:1 inversion to eversion ratio reporting this as variable between subjects (Miller and Maguire 2000).

The consensus on reliability studies agree in part, but accept discrepancies exist. Despite limitations manufacturers of in-shoe devices and clinicians persist in employing these techniques. It has been suggested despite the fundamental limitations and flaws the prowess of the podiatric clinician involved in orthotic prescription continues to prescribe such devices. The reason for the continued use of these techniques is the lack
of another alternative landmark or method (Nester 2006). The anecdotal reports of successes with in-shoe orthoses from patients and clinicians result in the scientific paradigm being rejected. Clinicians striving to be objective in the clinical setting are nowadays employing motion analysis systems.

Access to 2-D gait analysis and treadmill (TM) analysis lends itself to the clinical environment easily. It could be said in realistic terms 3-D gait analyses along with the financial constraints and depth of the analysis are more institutional orientated.

2.3.1 Accepted causes of abnormal pronation
Rearfoot varus, forefoot varus, tibial varum and limb length inequality are congenital deformities and are causes of over pronation, if one factor alone is evident. It has been proposed if more than one factor influences over pronation then these are cumulative (Shuster 1973). In rearfoot varus the calcaneus is inverted towards a sagittal plane. The compensation mechanism necessary to bring the medial border of the calcaneus and foot to the ground is STJ pronation, which occurs during early stance phase (Root et al., 1997).

Similarly a tibial varus deformity is a condition whereby the angle of the distal tibia deviates towards the sagittal plane with a resulting foot that orientates along the lateral border and adopts an inverted position. Compensatory STJ pronation and MTJ pronation may be necessary to evert the calcaneus and bring the forefoot to the plantigrade position. The pronation of the STJ and MTJ may exceed 6.0° of accepted normal pronation range of motion (Luo et al., 1997). The range of motion of the STJ is highly variable and no consensus exists regarding absolute measurement criteria or equipment used to determine accurate measurement of STJ motion. In addition there are no clear reference values of normal STJ motion (Beimers et al., 2008). Clinically STJ range of motion refers to frontal plane motion only and may range between 6-12° for normal gait (Sobel et al., 1999). In non weightbearing STJ inversion ranges from 5-50° with a mean value of 20°. STJ eversion ranges from 5-26° with a mean value of 10° (Engsberg, 1987; Huson, 2000; Inman, 1976; Isman, 1969; Leardini, 2001; Manter 1941).

The inverted forefoot or forefoot varus deformity is evident when the STJ is in a neutral position and the forefoot remains inverted to the sagittal plane. Compensatory pronation is necessary to reduce the forefoot medial aspect and first metatarsal head
to the ground. The pronation involved causes excessive medial rolling of the foot during midstance and has been linked with knee injury in the literature (Heiderscheit et al., 2002). As pronation continues through the midstance period of gait until the late stage of propulsion and toe off prevents the transverse (midtarsal) joints from locking to form a rigid lever for propulsion. In limb length inequality the foot on the longer limb may roll into compensatory pronation in an attempt to minimise the length discrepancy.

Despite frequent clinical use of RFA’s. The direct relationship to abnormal rearfoot motions during locomotion and the relationship to lower extremity injury are sparse (Cornwall and McPoil 2004). They concluded it is not the rearfoot position whether inverted or everted, as ultimately the pronation amounts are even. The differences are only the starting point, and the end point is the same. They propose the research has a dilemma of tunnel vision preoccupied with calcaneal position and range of motion. Subsequently a subject may make their necessary biomechanical compensations to minimise, eliminate or maximise an advantageous position and orientation. Essentially all individuals may pronate the same amount but the differences that may be observed are a result of the starting point of the rearfoot angle at heel strike rather than the end point value of motion (Cornwall and McPoil 2004). Beyond this excessive pronation may become an issue of tolerance of the extremes of motion contained by the tissues and as such may be described as a tissue stress model (McPoil and Hunt 1995). Accepted in the literature rearfoot pronation induces obligatory internal tibial rotation with subsequent soft tissue tensile stresses, lateral articular compressive stress at the STJ and knee joints (Tiberio 1988).

2.3.2 Abnormal rearfoot motion

Rearfoot motion is believed to indicate the amount of STJ pronation occurring within a foot. In normal circumstances, the STJ must pass from a position of 2.0° supination just prior to heel strike to 4.0° of pronation at midstance (Root et al., 1977). An excessive RFA or excessive period of pronation and abnormality of velocity of STJ pronation is thought to be associated with a range of running injuries (Taunton et al., 1988).

Bates et al, (1979) took six runners with a history of lower limb injury and orthoses worn for at least one year. The subjects were asked to run on a treadmill while their
rearfoot angles were recorded. Orthoses significantly reduced the period and the amount of maximum pronation by aligning the rearfoot relative to the running surface. In a similar study Smith et al. (1986) with 11 subjects with no leg pain found that soft and semi-rigid orthoses reduced rearfoot eversion by 1.0° while the subjects ran at 7-minute pace on a treadmill. The rate or velocity of rearfoot movement was more significantly affected being reduced by 15%. Kelly et al. (1992) found a significant decrease in calcaneal eversion when orthoses were used by 21 subjects of mean age 30 years with a minimum of 5.0° calcaneal eversion in stance. The reduction in the calcaneal eversion measured with 3-D kinematic analysis, was thought to result in a reduction of the movement generated by the opposing supinating muscles, hence the success of orthotic therapy in dealing with overuse muscle syndromes.

Common to all studies reviewed, was the assumption that rearfoot movement can be determined by placing markers on the shoe. For this to be accepted, it must be assumed that the foot does not move within the shoe and the angle the rearfoot makes relative to the supporting surface represents inversion and eversion only. The super structure above the foot must have no influence (Kilmartin and Wallace 1994). In a study of eleven men and nineteen women no obvious abnormality McPoil and Cornwall (1992) found that the STJ neutral position was never actually attained during the midstance phase of walking. This finding has important implications for understanding of the way in which in-shoe biomechanical orthoses affect foot and leg function. Until now, the objective of orthotic therapy has been to place the foot in a position that may never be attained by normal feet (McPoil and Cornwall 1992).

2.3.3 Limitations of rearfoot motion angles

Despite widespread adoption in podiatric fields the NCSP and RCSP have their limitations (Payne and Dannanberg 1997). A limitation arises when comparison between the two positions are made from only information relative to the frontal plane of the rearfoot is observed (Mueller et al.,1993). Cornwall and McPoil (1995)suggest sagittal plane motion or transverse plane motion may be a better indicator of abnormal pronation. A major assumption is that transverse tibial rotation can be used as an indirect measure of rearfoot pronation, there is a possibility that internal rotation could be due to pronation of the forefoot during the propulsive stage of gait (Tweed 2008). Research has focused on the stance phase of foot to ground interaction with the
exclusion of the swing phase of gait in orthotic studies. Root et al., (1966) nurtured the paradigm of NCSP and RCSP. Closed kinetic chain involves weight-bearing which is the most obvious factor in determining if an orthotic achieves the clinical goal of positioning and orientating a malaligned foot in a hypothetical advantageous vertical calcaneal position. In weight-bearing the calcaneal rearfoot angle is everted 37% more than in non weight-bearing (Lattanza et al., 1988). The observed increase or decrease in the rearfoot angle has provided a clinically useful method of frontal plane measurement for evaluation of static and dynamic motion, and in evaluation of orthotic effects and outcome. Static and dynamic observational assessments are made of foot position and RFA’s relative to the NCSP. The NCSP is employed for a referenced position during weight-bearing. The technique involves examiner palpation of talar congruence and has acknowledged limitations. In the RCSP the weight-bearing loaded foot demonstrates the compensations adopted by the lower limb which are necessary to bring the foot plantigrade with the ground. If the orthoses is correct the RFA may approximate the NCSP with bias away from the compensations of the RCSP however this has been questioned within the literature (Sobel and Levitz 1997; McPoil and Cornwall 1996; Elveru et al., 1988b). The RCSP has been shown as more stable as the foot is in a pronated position, which may increase the tension on the plantar fascia, intrinsic muscles and plantar longitudinal ligaments. The result is proposed as enhancing the effect of the windlass mechanism making the foot more rigid for active propulsion at toe off.

### 2.4 Proposed links between pronation and the knee

Excessive pronation has been linked with knee pain (Cornwall and McPoil 2004). Pronation has been identified as a causative factor in pathomechanics of the knee having an effect on the quadriceps angle, increasing the anterior displacement of the proximal tibia, increasing the quantity of ground forces reaching the knee and increasing loading of the intrinsic structures of the knee. If the foot pronates beyond the 4-6° and beyond the 25% of the gait cycle the tibia undergoes excessive internal rotation. At the same time the femur under influence of the plexus in rotation lateral forces pull the patella laterally and the effect on patella tracking increases compressive forces (Insall et al., 1983; Hvid et al., 1981).
The medial compartment is the most common site of osteoarthritis (OA). It has been reported all three compartments of the knee may be affected. There are several proposed pathways to degenerative arthritis of the knee. Repetitive impactions if generated during walking may raise varus moments at the knee, creating a large compressive force in the medial compartment, contact forces within the knee when walking are greater in the medial lateral compartment than in the lateral compartment. And a varus alignment of the knee is associated with the progression of medial compartment osteoarthritis (Sasaki and Yasuda 1987).

2.4.1 Anterior displacement of the proximal tibia

If the calcaneus is in an everted position during the stance phase, anterior displacement of the proximal tibia occurs with the internal rotation of the tibia (McPoil and Cornwall 1996). The anterior displacement causes knee flexion which in turn causes increased quadriceps activation (Hunt et al., 2000). Increased quadriceps activation may lead to glycogen depletion within the muscle, fatigue, inflammation of the muscle tendon unit (i.e. patella tendonitis) and a traumatic joint injury secondary to failure of the musculo-tendonous support system (Bender 1964). Increased force on the patella facets may gradually erode bone with resulting pain (Hughston 1984). Anterior displacement of the proximal tibia and increased quadriceps activation transmit traction loads to the anterior cruciate ligament (Frankel et al., 1971). Clinical and laboratory data indicate that these traction loads are not tolerated well by this ligament (Butler et al., 2007). Excessive pressures between the lateral facet of the patella and the lateral femoral condyles, and the iliotibial band have similarly been linked with STJ pronation (Greenfield 1990). If the calcaneus is maximally everted as a resultant compensation for a varus deformity of the tibia, rearfoot or forefoot it is then incapable of further eversion which is a main component of shock absorption of ground forces. The inability to absorb ground reaction forces has been linked with pain in the knee and hip joints (Valoshin and Wosk 1982).

Patellofemoral pain is one of the most common knee disorders seen in orthopaedic practice (Powers et al., 2004). Current literature suggests patellofemoral pain is multifactorial (Shepsis and Watson 2005), and related to patellar malalignment (Wilson et al., 2009). Imaging techniques using Magnetic Resonance imaging, Computed Tomography, X-ray, Motion analysis and Goniometry are employed for assessment of
the knee. With the exception of goniometry none of these methods lend themselves easily to the chairside examination. Exercises for strengthening the quadriceps may be used to prevent muscular imbalance between the vastus medialis which pulls the patella inwards and the vastus lateralis which pulls the patella outwards causing knee pain (O Shea 1985). In frontal plane structural deviation of the knee in genu valgum, genu varum may influence muscle function. It remains unclear whether subjects with different knee alignment show the same muscle function pattern (Sogabe et al., 2009).

In-shoe orthoses may reduce rearfoot movement and exert an effect on knee function but the clinical significance of excessive rearfoot movement has yet to be proven. Edington and Cavanagh (1990) studied knee movement in treadmill runners who were fitted with a 10.0° varus and 18.0° valgus, and neutral shoe midsoles. A small clinically insignificant change in the amount or the timing of knee flexion extension was detected. They concluded this was consistent with work by Lafortune (1987) who used intracortical pins to measure lateral movement of the patella in subjects wearing 10.0° varus 10.0° valgus and neutral midsoles. The greatest change in patellar position was in the first 1.6 mm. Similar work was carried out by Bates et al. (1979) who recorded maximum knee flexion and patella internal rotation in six runners who ran on a treadmill while barefoot, shod and in shoes with orthoses. No significant differences were found between any of the groups tested. Englewood and Pierynowski (1993) used kinematic analysis to determine the effects of foot orthoses on transverse, frontal and sagittal bone movement of the knee in subjects with patellofemoral pain syndrome. Movement in the transverse and frontal plane were significantly reduced at heel strike and the midstance phase of walking. No significant effect was observed in the sagittal plane. The orthoses had no influence on knee movement during the push off phase of walking, though frontal plane movement of the knee was reduced in the swing phase. In running subjects the only significant effect of orthoses was noted during the push off phase when the orthoses reduced both frontal and transverse plane movement of the knee.

In-shoe orthoses were found by D'Amico and Rubin (1986) to have a significant influence on the quadriceps angle (Q) figure 2.8. The angle was determined by measuring the centre part of the patella. A line was drawn from the anterior superior iliac spine through the posterior tibial tubercle distally figure 2.8. A high Q angle is
thought to be a significant cause of patella malalignment leading to prepatellar pain. Normal Q angles range from 8° to 18° in males and 12° to 16° in females (Hughston 1984). A significant reduction of the Q angle was found to occur where orthoses were positioned beneath the feet of twenty one standing subjects. The Q angle was reduced from 17.6° to 11.6°.

\[ \text{Figure 2.8 Clinical measurement of the quadriceps angle (Q-angle)} \]

Lefebvre and Boucher (1989) measured the effect of medial wedges of the foot on the Q-angle of the knee with 2-D motion video analysis. While the Q-angle was significantly reduced in static trials there was no significant effect in the dynamic motion. Doubling the thickness of the medial wedge under the foot did not produce any further reduction of the Q-angle. They concluded orthoses significantly reduced the Q-angle of the knee in static trials but there effect was not significant in dynamic trials. Studies using laterally placed insoles have demonstrated significant findings (Sasaki and Yasuda 1987; Wolfe and Brueckmann, 1991; Tohyama et al., 1991). Though there are documented difficulties with in-shoe and out outsole modifications, the greatest being compliance. Sasaki and Yasuda (1987) reported 55% of subjects no longer wearing devices after five year due to difficulties fitting devices into shoes, general discomfort or worsening pain. Apparent in the literature is an absence of documentation concerning in-shoe devices exacerbating a subject’s complaint or causing discomfort. A possible explanation for
this may be due to the subject’s ability to remove the device from their respective footwear.

2.5 Summary of literature
An orthotic may be defined as a biomedical device that alters limb function. The positive clinical effects of functional foot orthoses have been reported in various outcome studies (Donnatelli et al., 1988; Blake and Denton 1985; Moranos and Hodge 1993). Studies using 2-D optical systems to assess the effect of foot orthoses on frontal plane and rearfoot mechanics have reported mixed results. Genova and Gross (2000) suggested orthoses significantly alter rearfoot kinematics whereas other researchers suggest that orthoses do not significantly alter rearfoot motion. Orthoses are of multiple designs and material densities. Some investigators use custom made orthoses (Bates et al., 1979; Ferber et al., 2005) and others preformed (Redmond et al., 2009). The choice of materials is many and varied, for example Englewood and Pierrynowski (1993) used soft orthoses. Questioning the efficacy of soft devices (McCulloch et al., 1993; Nawoczenski et al., 1995) used semi-rigid shells and rigid devices (Bates et al., 1979; McCulloch et al., 1993). As a result individual structural and functional factors of subjects may not have always been taken into account (Donoghue et al., 2008).

Generally custom made devices may be constructed over a neutral cast with no shell modification and therefore may be considered a quasi functional or accommodative mould. Preformed devices may be adjusted in-house to medial longitudinal arch profiles with various additional clip on wedges that may exert a greater functional effect. As a result the all encompassing term in-shoe orthotic has become blurred. How much customisation is necessary (Menz 2009)? A recent Cochrane review concluded there is good evidence for custom devices in several patient conditions (Hawke et al., 2008) but there is a lack consensus with respect to prescribing guidelines. Knowledge about the specific functioning an orthotic provides is limited, (Nigg et al., 1999). It has been proposed that abnormal coordination of subtalar joint (STJ) pronation and supination (SUP) combined with flexion and extension of the knee joint may be a major cause of overuse injuries in the lower limb (Hadley et al., 1999). “Tibial coupling” and describes a coupling effect between rearfoot inversion and eversion, and tibial internal and external rotation. The muscles and ligaments around the ankle and knee are believed to act as a restraining mechanism preventing abnormal motion at the
knee joint which may be detrimental to the articular surfaces and the overall functional position of the knee or rearfoot, (Bessor et al., 1999). There is no clear consensus amongst researchers whether relief of knee pain is due to a reduction in internal tibial rotation, a reduction in ground reaction forces, a shift in force vector or altered alignment of the patellofemoral mechanics in the frontal plane. The exact contribution an external device such as an in-shoe orthoses still remains unclear (Ball and Afheldt 2002; Heiderscheit et al., 2002; Kilmartin and Wallace 1994; Landorf and Keenan 2000; Pratt 2000).

Clinical significance, statistical significance and practical significance are useful measures but are often absent. Clinical significance has been defined as,

“The extent to which a therapy moves someone outside the range of dysfunctional population or within the range of the functional population” (Jacobson and Truax 1991).

The proposed two components of this index of change, the status of the subject after the orthotic therapy and how much change has occurred during the course of the orthotic therapy. With respect to the amount of change measured in degrees (°) for the STJ, ankle and knee joints clinical ranges of joint motion are defined in the literature but may be considered relative to holistic data concerning the patient. The primary reason, differences are common in a population and small amounts within or between individuals may be unlikely to be clinically significant because they reflect normal variation. Small differences may lack a practical significance for example a single degree of transverse rotation of a hip joint or a leg length discrepancy of 3.0 mm.

Clinical significance may also be considered when interpreting a difference in outcome measures that are statistically significant and unlikely to have occurred by chance. However statistical significance used in hypothesis testing does not provide information about magnitude of effect, practical significance or clinical significance (Haase et al., 1989). As may be the case where the reader is left to accept a statistically significant value and cannot make as one can in a clinical situation a link between the outcome of a specific intervention and the individuals more general functioning (Sattler 2008; Kaufman and Lichenberger 2005). In terms of clinical treatment, clinical significance answers the question is a treatment effective enough to cause the patient to be normal.
To attempt to quantify for the purpose of this thesis clinically significant describes what is clinically measureable using a 3-D motion analysis system within our laboratory and that is 0.5 mm in length measurement and in angular change 0.5°. However these values are used only to provide for statistical analysis and hypothesis testing. The principle driving force in acceptance of values a small as these are verbal reports from the individuals of improvements in their ailment. Boney articulations have been investigated using imaging techniques that place a grid reference across the plateau of articular surface and may be measured in terms of millimetre and square millimetres (Whitney et al., 2009). While this has a role in development of technology for advances in the future in patient care and computerised application the usefulness in a clinical setting is questionable.

This thesis aims to investigate the effect of custom made computer designed and manufactured in-shoe orthoses and standard preformed devices on dynamic motion of the rearfoot and tibial kinematics considered as synonymous with transverse plane knee motion.
Chapter 3 MEASUREMENT OF REARFOOT ANGLES IN SUBJECTS WEARING WEDGED FOOTWEAR: A 2-D STUDY

INTRODUCTION
Clinical observation of foot alignment is common and aids the clinician in classification of the foot into pronated neutral or supinated types. An important factor in classification and formulation of treatment intervention is the measurement of the calcaneal (calc) deviation around the vertical calc position. If the calc is wedged into a more advantageous position at heel strike, restoring tibio-talar alignment in the frontal plane, there is an assumption the events of the stance phase will proceed along a more beneficial pathway of progression. Rearfoot eversion has been linked with pronation related injury and impacts the function of adjacent joints (Carter et al., 2002). The literature suggests STJ pronation and knee flexion during the stance phase of gait is coupled with internal tibial rotation. During the latter stages of stance the STJ supinates as the knee joint extends and the tibia externally rotates. If resupination of the STJ is delayed in excessive rearfoot eversion, then extension of the knee may be affected and an antagonistic coupling stress may arise (Hadley et al., 1999). Given the repetitive nature of running the research proposes this as a possible cause of injury (McClay and Manal 1997; Bates et al., 1979; James 1990; Hamill et al., 1999). The introduction of an in-shoe orthotic device that alters the rearfoot angle at heel strike and ultimately midstance and has been accepted universally (Braithwaite et al., 2004). Research during the last decade has concentrated on kinematic analysis of the tibia and rearfoot. Several studies have investigated in-shoe orthoses and running. The conclusions include reductions in maximum rearfoot eversion, maximum rearfoot eversion velocity, internal ankle inversion moments, impact peaks, maximal vertical loading rate and maximum tibial internal rotation angle.

In-shoe devices vary from study to study and are questionable in terms of design and material used in manufacture. The literature is less clear about the specific effect of a single wedge less any medial arch support. Some researchers use machined wedges (Sasaki and Yasuda 1987) others modify the outsole of footwear (Clarke et al., 1983; Stacoff et al., 1988; Nawoczenski et al., 1995; Butler et al., 2007). Researchers have used accredited laboratories to manufacture the devices (MacLean et al., 2006) and some use their own in-house services (Rubin 2005; Hannaford 1986).
The materials used also vary. Some researchers have used polypropylene (McPoil and Cornwall 1992; Tomaro 1993; Sloss 2002) and others ethyl vinyl acetate (EVA) (Lafortune et al., 1994) or a carbon fibre composite (Ferber et al., 2005). Some study designs have given the same preformed device to all subjects which may compound a subject’s pain or compensation mechanisms, or introduce discomfort and as a result skew data for the trial subject (Mundermann 2003). The past research indicates much contradiction between studies because of methodological variations in study designs, inclusion / exclusion criteria, subjects used, healthy or symptomatic. We have been grounded in the proposals of Root (1977), and current research is questioning using modern techniques, what constitutes a normal gait or stride. With increasing reports of anecdotal successes with different orthotic types ranging from the very soft to ultra rigid the research has began to look at tissue stress models, joint coupling, and tarsal gears as alternative explanations of injury and prevention.

3.1 Rationale for Hypothesis #1

The rationale for this study was to investigate the most fundamental approach to clinical intervention, the role of a simple wedge on the rearfoot during the stance phase of walking gait on which the origins of the Root theory were proposed. The aim of this study was to determine relative rearfoot kinematics between the calcaneus, talus and tibia at the ankle and subtalar joints during gait following the introduction of wedged strips to the outsole of subject’s footwear. Specific aims and objectives were,

1. To determine the effect of wedged footwear on the 2-D kinematics of the rearfoot during walking.

2. Assessment of 2-D kinematic techniques of frontal plane rearfoot angle measurement in dynamic gait.

It was hypothesised wedged footwear would change frontal plane (rearfoot inversion / eversion) angles at heel strike and midstance phases of gait during walking.
3.2 Methods

3.2.1 Formal hypotheses

The hypotheses generated were,

\( H_a \) Rearfoot 4° medially wedged footwear changes the mean rearfoot angle at heel strike when compared to shod only conditions.

\( H_o \) Rearfoot 4° medially wedged footwear does not change the mean rearfoot angle at heel strike when compared to shod only conditions.

\( H_a \) Rearfoot 4° medially wedged footwear changes the mean rearfoot angle at midstance when compared to shod only conditions at midstance.

\( H_o \) Rearfoot 4° medially wedged footwear does not change the mean rearfoot angle at midstance when compared to shod only conditions at midstance.

\( H_a \) Rearfoot 6° medially wedged footwear changes the mean rearfoot angle at heel strike when compared to shod only conditions.

\( H_o \) Rearfoot 6° medially wedged footwear does not change the mean rearfoot angle at heel strike when compared to shod only conditions.

\( H_a \) Rearfoot 6° medially wedged footwear changes the mean rearfoot angle at midstance when compared to shod only conditions at midstance.

\( H_o \) Rearfoot 6° medially wedged footwear does not change the mean rearfoot angle at midstance when compared to shod only conditions at midstance.

\( H_a \) Full length 4° medially wedged footwear changes the mean rearfoot angle at heel strike when compared to shod only conditions at heel strike.

\( H_o \) Full length 4° medially wedged footwear does not change the mean rearfoot angle at heel strike when compared to shod only conditions at heel strike.

\( H_a \) Full length 4° medially wedged footwear changes the mean rearfoot angle at midstance when compared to shod only conditions at midstance.
Chapter Three Two-Dimensional Rearfoot Angles

$H_0$: Full length 4° medially wedged footwear do not change the mean rearfoot angle at midstance when compared to shod only conditions at midstance.

3.2.2 Subject sampling method
The method of recruiting subjects for inclusion in this research was convenience sampling. This may be described as a simple and cost effective method for a study of this size. The convenience of this sampling method was justified over the considerable time expense and planning of other sampling methods for a pilot study.

3.2.3 Sampling error
The time parameters for data collection and the extent of subject involvement, investigator participation and manual digitisation processes, it was decided a group of subjects may be justified by increased trials per subject in each condition. The investigator was aware a small sample size may result in a reduction in the power of the statistical analysis however research in this area similarly demonstrates small numbers of subjects with fewer trials per subject. Finally overuse injuries are observed in the younger athletic population therefore were sampled for inclusion in this research.

3.2.4 Description of sample
Nine male subjects were selected from a group of recreational runners. All subjects voluntarily underwent a lower extremity Podiatric biomechanical examination and agreed to participate in the investigation. Subjects were healthy volunteers and were in attendance at a podiatry clinic for general lower limb complaints. These complaints were shin splints, achilles tendonitis, plantar fasciitis and patellofemoral pain. All subjects were given an explanation of the research prior to participation. All subjects signed an informed consent form prior to participation in the study. (Appendix N) A full description of the subject sample used is provided (table 3.1). Clinical characteristics of subjects and demographic data provide a picture of the sample used. This will assist further investigations in similar areas of research so one can judge the sample to their studies or institutions.
Table 3-1 Demographics of subjects.

<table>
<thead>
<tr>
<th>n=9</th>
<th>Male/Female</th>
<th>Age (Yrs)</th>
<th>Height (m)</th>
<th>Weight (kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean</td>
<td>male</td>
<td>24.1 ± 7.2</td>
<td>1.7 ± .09</td>
<td>80.6 ± 13.0</td>
</tr>
</tbody>
</table>

3.2.5 Inclusion and exclusion criteria

All subjects underwent a podiatric medical history and detailed examination of the lower extremities carried out by a Department of Health registered Podiatrist prior to participation in this investigation (table 3.2). Subjects were selected using the following criteria.

Table 3-2 Inclusion and exclusion criteria for study

<table>
<thead>
<tr>
<th>Inclusion criteria</th>
<th>Exclusion criteria</th>
</tr>
</thead>
<tbody>
<tr>
<td>Over 18 years of age</td>
<td>Under 18 years of age</td>
</tr>
<tr>
<td>Passed medical examination</td>
<td>Failed medical examination</td>
</tr>
<tr>
<td>Navicular drop ≥10 mm</td>
<td>Navicular drop &lt; 10 mm</td>
</tr>
<tr>
<td>No acute injury of lower limbs</td>
<td>Acute injury of lower limb</td>
</tr>
<tr>
<td>Familiar with treadmill walking</td>
<td>Unfamiliar with treadmill walking</td>
</tr>
<tr>
<td>Nil medication</td>
<td>Taking medication</td>
</tr>
<tr>
<td>Signed consent</td>
<td>No signed consent</td>
</tr>
</tbody>
</table>

3.2.6 Dependent and independent variables

A quantitative experimental design was chosen for this research study. A pre and post test design was employed. Measurements were taken of the independent variable in this case the rearfoot angle in RCSP before any intervention. Then measurement of the dependent variable change in the rearfoot angle after the intervention of the wedged trial conditions. This method of investigation allowed the measurement of change for individual cases and provided a baseline barefoot walking reference universally adopted in the literature. Previously reported the measurement process pre-intervention can induce a change in the measured outcome termed, “practice effect”. Therefore all subjects were blinded to the wedged footwear interventions by separating the subject from the laboratory where the footwear and equipment was prepared. Subjects were allowed acclimatise to the surroundings and the equipment.
In summary the pre and post-test design involved manipulation of the calc angle using a known angular wedge and measurement of the outcome or amount change of the calc angle, using 2-D motion measurement. The control within the design maximised the internal and external validity of the investigation giving strength to inferences drawn. The following dependent variables for each subject performing ten walking trials in each of the treatment conditions were recorded.

- **Heel strike angle (°)** The rearfoot angle at frame ten of the contact phase.
- **Maximum pronation angle (°)** The rearfoot angle at frame 100 of the contact phase angle of the foot during ground contact.

The heel strike angle recorded at frame ten accounted for impact forces, fat pad and footwear compression. Providing validity to the angle attained at this point in the gait cycle. The maximum pronation angle frame one hundred represented a vertical tibia with maximum foot loading providing validity to the angle attained.

The subjects were their own baseline, walking barefoot to establish a dynamic gait cycle reference pre-intervention, from which any change induced from the wedge could be determined. The experiment consisted of five experimental trial intervention levels (Table 3.3).

**Table 3-3 Clinical groupings (n=9) demonstrating the five experimental conditions**

<table>
<thead>
<tr>
<th></th>
<th>Baseline</th>
<th>Trial 3</th>
<th>Trial 4</th>
<th>Trial 5</th>
<th>Trial 6</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experiment</td>
<td>Barefoot walking</td>
<td>Shod only</td>
<td>4° rearfoot wedge</td>
<td>4° full length wedge</td>
<td>6° rearfoot wedge</td>
</tr>
</tbody>
</table>

### 3.2.7 Location of the subtalar joint neutral position

The location of the STJN position was identified in weight-bearing and non weight-bearing position by palpation of talar congruence on the medial and lateral sides of tibiotalar joint using the thumb and forefinger following inversion and eversion of the foot and ankle. The technique had been previously documented by Wernick and Langer (1971).
3.2.8 Location of the relaxed calcaneal stance position

The RCSP was obtained from each subject by bisecting the posterior surface of the calcaneus. The subject was requested to position themselves in their relaxed calcaneal stance position by instructing them to march on the spot bringing the femur parallel to the floor with the knee flexed to 90°. Subjects were then requested to stop, and standstill. The angle of the calcaneal bisection was recorded using a fluid filled goniometer (Figure 3.1).

All angles recorded were calculated relative to the NCSP of the subject on the treadmill. The investigator was competent at locating the NCSP for each subject on the treadmill having been in practice for twenty years within a biomechanical setting and conducting examination of the foot daily. The rearfoot angle was measured as according to methods adopted previously in the literature (Clarke et al., 1983; Eddington et al., 1990) (Figure 3.2).
3.2.9. Navicular drop

The vertical height of the navicular of the right foot of each subject was measured in NCSP and RCSP using a vertical height gauge. Following measurement of RCSP the examiner identified NCSP by the palpation method described previously (Soutas-Little 1996; Wernick and Langer 197). Subjects with a navicular drop ten millimetres or more measured between the NCSP and the RCSP were included in the study.

3.3 Instrumentation

The instrumentation used in this investigation was standardised apparatus for lower limb gait analysis and evaluation. A full description of instrumentation is provided for further research.

- Inclinometer
- Reflective tape 3M™
- Standard universal goniometer 600mm
- Standard measuring tape
- Treadmill model Proform 725ex
- Two Digital Sony DV cameras model
- Peak Performance software version
- Vision mixer Panasonic AVE 55
- 500 watt Lights with barn doors
- Ethyl vinyl acetate (EVA) shore 60 wedged strip 4°
- Ethyl vinyl acetate (EVA) shore 60 wedged strip 6°
3.3.1 Environmental setup

Two digital video cameras (DV) were mounted on tripods 600 mm above the ground. Camera one was positioned at the right side of the treadmill and camera two at the rear. A light source was positioned behind each camera (Figure 3.3).

![Figure 3.3 The laboratory setup for data collection.](image)

3.3.2 Camera setting

The calibration of the digital cameras and vision mixer was conducted after the cameras had been placed relative to the treadmill. Cameras were setup and calibrated by the same investigator and remained at the same setting and position throughout data collection. The following procedure was followed with the cameras powered on and in record mode.

1. The shutter duration was set at 1/500 of a second.
2. Focus was set to manual.
3. The calibration weight was placed on the centre of the treadmill.
4. Focus was zoomed fully in, then fully out to the required frame.
5. The camera remained in this position and setting throughout the recording.
6. Lighting was positioned parallel to the line of the camera.
7. All reflective surfaces in the field of the camera were masked.
8. The aperture was partially closed to dim the background light and introduce a contrast with the skin markers.
9. All reflective surfaces on the subject’s footwear were also masked.
3.3.3 Camera calibration

1. With the cameras powered on and in recording mode the calibration weight 0.25 kg was suspended above the centre of the treadmill.
2. The calibration weight was dropped from a vertical height of one metre. Recording began two seconds before the weight was dropped and continued for two seconds after the weight impacted with the treadmill surface.
3. Audio was powered on to provide more objectivity for identification of the single frame recorded at the moment of impact of the calibration weight.
4. The synchronization process was then checked on the viewing monitor to ensure both cameras were recording the same event at exactly the same time.

3.3.4 Treadmill calibration

The treadmill speed was calibrated at the time of manufacture to the speed of the treadmill belt. Manufacturers specify calibration is not required. Treadmill calibration is not specified in the literature and was therefore accepted as true.

3.4 Subject skin marker placement

Adhesive reflective skin markers 10 × 10 mm, were placed over bony anatomical landmarks identified by observation and palpation. Prior to application of reflective markers the skin was swabbed with 70% isopropyl alcohol swabs to remove residual skin oil and aid skin marker adhesion. The location of the skin markers were described by Clarke et al., (1983) and are used throughout the literature. The locations of skin marker placement were listed and shown in (Figure 3.4).

1. The lateral head of the fibula.
2. The lateral malleolus.
3. The lateral border of the calcaneus.
4. The lateral aspect of the fifth metatarsal head.
5. The midpoint of the gastrocnemius muscle belly between the popliteal fossa and the ground.
6. The lower aspect of the gastrocnemius muscle between the medial and lateral malleoli.
7. The superior aspect of the calcaneus.
8. The inferior aspect of the calcaneus.
3.4.1 Footwear marker placement

Retroflective markers were placed on the right shoe of each subject at the following locations (Figure 3.5).

1. The area over the fifth metatarsal head.
2. The inferior lateral calcaneus above the outsole on the heel counter.
3. The superior lateral calcaneus above the outsole on the heel counter and under the lateral malleoli.
4. The posterior heel-counter inferior aspect.
5. The superior heel-counter superior aspect.
3.4.2 Placement of wedges on footwear

Prior to recording, footwear for each subject was removed from the room by the examiner. A known angular wedge was placed on the outsole of the left and right shoe beginning with a 4° medial rearfoot wedge (Figure 3.6). Each subject was assigned an identification number and recorded on the data sheet. This procedure was repeated for each subject in the investigation.

![Figure 3.6 Wedge placement adhered to the outsole of footwear (Left 4° rearfoot wedge, Right 4° full length wedge).](image)

3.5 Data collection

All subjects were allowed a five minute walk barefoot on the treadmill at a self selected pace to allow for acclimatisation to the surroundings and treadmill surface. During this period recordings were not made.

Subjects were instructed to stand barefoot on the treadmill in RCSP while the cameras were operational for five seconds. The right foot was then placed in the NCSP using the palpation method (Soutas-Little 1996). This was recorded for a further five seconds. This position served as a baseline reference for analysis of the rearfoot angle.

The subject was instructed to adjust the treadmill speed to a self selected pace. After one minute, the recording of rearfoot angles was powered on and continued for fifteen heel strikes of the right heel of each subject, in each trial condition. The procedure was repeated for each subject as they donned their respective footwear relative to the trial assignment. Ten digitized trials were recorded for each subject in each condition (5 levels). A subject mean for each of these variables was computed. The subject means were then analysed as a group mean. The statistical analysis was performed using subject means for each of the dependent variables. A repeated measures analysis of
variance (ANOVA) was performed using SPSS for Windows version 13 with p<0.05 level of significance.

3.6 RESULTS
The independent variable was the orthotic wedge intervention. Within subject factor measures are provided for mean heel strike angles and group mean data for the rearfoot angle at heel strike is provided (Table 3.4).

3.6.1 Rearfoot angle at heel strike
Within-Subjects Factors

Table 3-4  Within subject factor description and measure of mean angle at heel strike with 95% confidence intervals * Denotes significance at 0.05 level

<table>
<thead>
<tr>
<th>Wedge</th>
<th>Dependent Variable</th>
<th>n=9</th>
<th>Mean (°)</th>
<th>SD (°)</th>
<th>95% CI (°) Lower bound</th>
<th>95% CI (°) Upper bound</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Barefoot</td>
<td>9</td>
<td>-3.1</td>
<td>4.6</td>
<td>-6.7</td>
<td>0.3</td>
</tr>
<tr>
<td>2</td>
<td>Shod</td>
<td>9</td>
<td>-0.9</td>
<td>6.8</td>
<td>-6.2</td>
<td>4.2</td>
</tr>
<tr>
<td>3</td>
<td>Rft4deg</td>
<td>9</td>
<td>-0.5</td>
<td>4.9</td>
<td>-4.3</td>
<td>3.2</td>
</tr>
<tr>
<td>4</td>
<td>Full4deg</td>
<td>9</td>
<td>-2.7</td>
<td>5.1</td>
<td>-6.6</td>
<td>1.2</td>
</tr>
<tr>
<td>5</td>
<td>Rft6deg</td>
<td>9</td>
<td>-1.1</td>
<td>6.1</td>
<td>-5.8</td>
<td>3.5</td>
</tr>
</tbody>
</table>

All null hypotheses were retained at heel strike and alternative hypotheses rejected at the 0.05 level of significance for all wedge conditions (Appendix A). Wedge within-subject effects were computed. Mauchly’s test of sphericity, assumption of homogeneity of variance was violated. Sphericity assumed p=0.813. Corrected values for wedge within-subject effect at heel strike were Greenhouse-geisser p=0.662.

Tests for between-subject effects were computed at F=4.079 p=0.078

Post hoc pairwise comparisons to identify were differences lay were not significant as were bonferroni post hoc comparisons.
3.6.3 Rearfoot mean angles at heel strike.

Boxplots of rearfoot angles at heel strike are presented in graph 3.1.

![Mean Rearfoot Angle At Heel Strike With 95% Error Bars (n=9)](image)

Graph 3-1 Mean rearfoot angles at heel strike with 95% confidence intervals (p>0.05).

3.6.4 Maximum pronation angle at midstance

Mauchly’s test of sphericity was violated. Post hoc corrections were Greenhouse-geisser p=0.511.

Tests of between-subject effects were F=.159 p=0.700

3.6.5 Maximum pronation angle descriptive statistics at midstance

Group mean and standard deviation data for the maximum pronation angle at midstance are provided in (Table 3.5).

Table 3.6 lists means and 95% confidence intervals of the means for all wedge interventions on rearfoot angle at midstance. Post hoc pairwise comparisons were calculated to see where the differences lay. At midstance 4° rearfoot and 4° full length wedges were significantly different than the shod only condition, with changes of 4.0° p=0.014 and 5.0° p=0.005) respectively. 4° full length wedge and 6° rearfoot wedges introduced rearfoot angle change of 2.0° degrees p=0.002 (graph 3.2).
A summary of hypotheses accepted and rejected is provided (Table 3.7).

Table 3-5 Descriptive statistics mean maximum pronation angle at midstance

<table>
<thead>
<tr>
<th></th>
<th>N=9</th>
<th>Mean (°)</th>
<th>SD (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Barefoot</td>
<td>9</td>
<td>-0.7</td>
<td>5.2</td>
</tr>
<tr>
<td>Shoe only</td>
<td>9</td>
<td>2.5</td>
<td>4.2</td>
</tr>
<tr>
<td>4deg rearfoot wedge</td>
<td>9</td>
<td>-1.5</td>
<td>5.7</td>
</tr>
<tr>
<td>4deg full length wedge</td>
<td>9</td>
<td>-2.4</td>
<td>5.6</td>
</tr>
<tr>
<td>6deg rearfoot wedge</td>
<td>9</td>
<td>-0.4</td>
<td>5.4</td>
</tr>
</tbody>
</table>

Table 3-6 Maximum pronation angles at midstance and 95% confidence intervals
* Denotes significance at 0.05 level

<table>
<thead>
<tr>
<th>wedge</th>
<th>Mean (°)</th>
<th>Standard error (°) (SEM)</th>
<th>(°) Lower bound</th>
<th>(°) Upper bound</th>
</tr>
</thead>
<tbody>
<tr>
<td>bf</td>
<td>-0.7</td>
<td>1.7</td>
<td>-4.7</td>
<td>3.2</td>
</tr>
<tr>
<td>so</td>
<td>2.5</td>
<td>1.4</td>
<td>-0.7</td>
<td>5.7</td>
</tr>
<tr>
<td>Rf4</td>
<td>-1.5*</td>
<td>1.9</td>
<td>-5.9</td>
<td>2.9</td>
</tr>
<tr>
<td>Fl4</td>
<td>-2.4*</td>
<td>1.8</td>
<td>-6.7</td>
<td>1.8</td>
</tr>
<tr>
<td>Rf6</td>
<td>-0.4*</td>
<td>1.8</td>
<td>-4.5</td>
<td>3.7</td>
</tr>
</tbody>
</table>
Graph 3-2 maximum pronation angles at midstance with 95% confidence intervals

Table 3-7 Hypotheses accepted and rejected at the 0.05 level of significance

<table>
<thead>
<tr>
<th>Heel strike</th>
<th>Accept or Reject</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$H_a$ 4° rearfoot medially wedged footwear changes the mean rearfoot angle</td>
<td>Reject</td>
<td>P=0.899</td>
</tr>
<tr>
<td>at heel strike when compared to shod only conditions.</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$H_a$ 4° full length medially wedged footwear changes the mean</td>
<td>Reject</td>
<td>P=0.630</td>
</tr>
<tr>
<td>rearfoot angle at heel strike when compared to shod only conditions at heel</td>
<td></td>
<td></td>
</tr>
<tr>
<td>strike.</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$H_a$ 6° rearfoot medially wedged footwear changes the mean rearfoot angle</td>
<td>Reject</td>
<td>P=0.972</td>
</tr>
<tr>
<td>at heel strike when compared to shod only conditions at heelstrike.</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Midstance</th>
<th>Accept or Reject</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$H_a$ 4° rearfoot medially wedged footwear changes the mean rearfoot angle</td>
<td>Accept</td>
<td>P=0.014</td>
</tr>
<tr>
<td>at midstance when compared to shod only conditions.</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$H_a$ 4° full length medially wedged footwear changes the mean</td>
<td>Accept</td>
<td>P=0.005</td>
</tr>
<tr>
<td>rearfoot angle at heel strike when compared to shod only conditions at heel</td>
<td></td>
<td></td>
</tr>
<tr>
<td>strike.</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$H_a$ 6° rearfoot medially wedged footwear changes the mean rearfoot angle</td>
<td>Reject</td>
<td>P=0.057</td>
</tr>
<tr>
<td>at midstance when compared to shod only conditions at midstance.</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$H_a$ 6° rearfoot medially wedged footwear changes the mean rearfoot angle</td>
<td>accept</td>
<td>P=0.002</td>
</tr>
<tr>
<td>at midstance when compared to 4° medially wedged footwear at midstance</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
3.7 DISCUSSION
The aim of this study was to determine relative rearfoot kinematics between the calcaneus, talus and tibia at the ankle and subtalar joints during gait following the introduction of wedged strips to the outsole of subject’s footwear. The findings support in part previous reports achieving a reduction of 3-4.0° in maximum eversion with the use of medial wedges (McCulloch et al., 1993; Nigg et al., 1998).

At heel strike barefoot values were reduced from -3.1° to -0.9° in shod trials. Addition of a 4.0° medial rearfoot wedge further reduced the shod value to -0.5° though these did not reach statistical significance (Table 3.7). On review a reduction of 0.4° in a clinical practice context is not measurable or clinically useful.

At midstance a 4.0° rearfoot wedge reduced the rearfoot angle 0.8° over shod only trials p=0.014 (Table 3.7). Increasing the outsole wedge to a 4.0° full length wedge provided a further 0.9° reduction over the rearfoot wedge that was surprising given the global inverted observation of the subject shod with the longer 4.0° length of the wedge. It was anticipated the mean angle difference would have been greater than 1.0°. A possible explanation for this may be forced dorsiflexion of the first ray absorbing the medial thickness of the wedge overall with little increase in the rearfoot angle. The rearfoot eversion angles at midstance were the only significant changes caused by medial wedged outsoles. Though it is acknowledged in the literature there are other factors that influence the success of insole treatment, the small angular change may be due to the lack of a medial arch support. A medial arch support may help to prevent medial rolling of the foot on the wedge. A lack of support or pressure directed against the sustentaculum tali and medial tarsus may be less effective at limiting STJ pronation and have resulted in the small values seen between trials. Payne et al. (2003) assessed changes in the frontal plane calcaneal angle in 18 subjects (36 feet) in static weight-bearing trials recording data as subjects stood on six different types of orthoses only three devices changed the calcaneal angle in the frontal plane however all increased navicular height measurements. A heel wedge produced a significant change in the rearfoot angle which is similar to the findings of this study. Payne et al., (2003) reported devices that are designed to apply greater movement to invert the rearfoot than on the longitudinal arch area account for the lack of change in the rearfoot angle, which would be in agreement with our findings but results of their
study must be interpreted with caution as the frontal plane rearfoot angle was taken at static midstance. One cannot assume static data may be representative of dynamic function (Hunt et al., 2000; McPoil and Cornwall 1996).

The design of this pilot study was to break down the functional wedge component of the orthotic, removing the arch to determine the exact contributions each piece may have. With no arch filler it did not take into account the role of moulded devices on the Windlass mechanism and the 3-D structure of the foot. Sir Isaac Newtons (cited in Hamill and Knutzen 2003) third law stated for every action there is an equal and opposite reaction. If the medial longitudinal arch is supported inadequately and the subjects bodyweight or range of motion exceeds the support given by the device, in principle there may be no change in the subjects symptoms, pain or maximum rearfoot angle. This would appear in keeping with this study and has been previously reported with investigations using accomodative and soft shore devices. (Braithwaite et al., 2004) This may be a possible reason for the lack of statistical significance at heel strike, and an explanation why the 6.0° rearfoot wedge was not significant over shod only conditions when the previous 4.0° wedged trial was significant at midstance (Table 3.7).

Supporting the longitudinal medial arch has been suggested as assisting dynamic foot pronation and supination from all foot segments. A medial arch filler may increase the angular values attained in studies utilising contoured or moulded orthotics. The objective of this study was to determine the effect of a simple machined outsole wedge on the rearfoot angle and surprisingly the values attained between conditions were small raising a theory of a “combined effect”. The multiple components of an orthotic device may provide the greatest mechanical advantage. Subjects with abducted feet may roll medially in the absence of a resistive medial arch support consequently the rearfoot angle values may be increased and statistical significance not reached.

Cornwall and McPoil (1999) investigated kinematic coupling mechanisms within the STJ and talonavicular components of the tarsus concluding foot motions occur in coordination throughout the gait cycle. Research has associated each foot type with overuse injury (Simkin 1989). Although it has been reported pronated foot types with low medial longitudinal arches are not more prone to injury rather there appears to be an inherent inbuilt compensation mechanism that provides protection against injury.
This is in agreement with Cowan et al., (1993) but until recently conflicted with most schools of thought.

3.7.1 Limitations of 2-D motion analysis

This study has several limitations, calcaneal inversion and eversion was recorded using two reference markers placed on the midline of the calcaneus for barefoot trials and on the outside of the shoe for walking trials. A difficulty arose when studying rearfoot movement exhibited by each subject’s running shoe because the midline of the calcaneus was estimated from the two markers placed vertically on the rear of the shoe. Marker placements did not allow for identification of specific joints within the foot. The method adopted the foot as a single rigid segment model. Despite an acknowledged possibility of heel movement within the shoe, it was adopted in this research to employ this method of data acquisition as this is a common method employed at a clinical level. Previous researchers have cut an aperture in the heel centre of footwear which has been linked with altered rearfoot control (Ferber et al., 2004). Clinically this method could not be utilised in an everyday outpatient setting. In support of this study design heel and shoe movements are well correlated (Bates et al., 1983; Nigg et al., 1992).

Footwear was not standardised and common to running shoes are gel type shock absorbers manufactured into the heel. The overall effect of the wedge may have been lost to compression of the heel over the wedge at heel strike when body weight was combined with the ground reaction force.

Few previous studies have independently measured the abduction angle in their research. Braithwaite et al., (2004) investigated this angle and concluded that the abduction angle is not influenced by anti pronatory treatments. This conforms to existing, understanding of the pronation components and the abduction angle is not a determinant of STJ pronation (McCulloch et al., 1993). In hindsight the abducted foot places the STJ longitudinal axis and line of progression more medial to the talo navicular joint and has been linked with degenerative disease and pain Kirby (2001). Projection errors may occur during recording where the knee flexed during motion and obscured by the opposite limb. The camera rate was at 50 frames per second (fps). A possible reason for the lack of significance observed may be due to the poor system sensitivity.
of the small movement of the markers relative to the camera frontal plane set up in the abducted heel strike and midstance events, or the during the visual determination of the heel strike event by the investigator during video editing processes.

Individuals with pathological foot deformities were not measured. Subjects were asymptomatic but included if navicular drop was 10 mm or greater. Foot types exhibiting greater rearfoot angle valgus values may have required greater angular wedges to exert any effect. The wedges used in this study were commercially made and were not subject specific to any excessive deviations of the foot. In the clinical situation practice would be to increase or decrease the wedge value. The RCSP was manipulated to place the foot in NCSP, this was then subtracted from the recorded foot motion. In this way the motion of the foot was expressed relative to each subject’s NCSP angle. This method is common amongst researchers to normalise data to more clinically significant measures of rearfoot STJ range and angle of motion (Donoghue 2008). The principal of employing a digital system to measure objectively and reliably rearfoot position may be undermined as intratester bias is introduced and may reduce internal validity.

3.8 CONCLUSION
A reduction in rearfoot eversion can be directly linked to a reduction in STJ pronation (Blake and Ferguson 1993; Landorf and Keenan 2000; Stacoff et al., 2000abc). This hypothesis was accepted at midstance in this study of wedged footwear. Use of 2-D motion analysis returned a rearfoot angle measurement at heel strike that was not statistically significant. The results may have been more informative if custom moulded devices had been used and a range of motion calculated rather than a single discrete data point angle. At a clinical level single joint orientation values may never be attained again if the same study was repeated. Methodological rigor must include standardised footwear, a larger subject sample and more robust method for inclusion and classification of foot types. Identification of heel strikers, forefoot strikers, excessive pronators and supinators would be desirable for exclusion or inclusion purposes. Determination of motion analysis system reliability and validity must be established. Future work may investigate the motion of the calcaneus within the shoe using 3-D motion coupled with transverse plane tibial rotation. Tibial rotation may be a better
indicator of foot function though it is not as practical in clinical situations because of the technical practicalities of calibration, multiple camera system setup and cost.

3-D motion analysis may provide a more holistic insight into the contribution of in-shoe devices on the rearfoot heel angle and maximum pronation values during walking in subjects who abduct their feet. In investigating complex dynamic gait variables it was concluded 2-D analysis would fall short. The relative ease of use of 2-D motion over 3-D methodology may be a factor in the continual use in mainstream clinical practice. The clinical acceptance of 2-D outcomes may retard future progress in our understanding of lower limb function and pathology. The results of this study suggest 3-D motion analysis may enhance objective assessment of the foot and orthotic interaction in dynamic gait with use of a greater sampling frequency and modified laboratory setup to maximise objective data acquisition for improved patient and research outcomes.
Chapter 4 VALIDATION OF THE SIMI 3-D MOTION ANALYSIS SYSTEM USING A RIGID TIBIAL MODEL AND BESPOKE CALIBRATION FRAME.

INTRODUCTION
Important sources of error in motion analysis systems have been identified (Stagni et al., 2009; Collins and Adamczyk et al., 2009; Collins and Ghoussayni et al., 2009; Hagemeister et al., 2005). Some of these errors have been linked to marker placement and replacement across trials, by the investigator or between investigators. Soft tissue artefact (STA) has been reported as the most invalidating source of error in human motion analysis (Alexander and Andriacchi 2001). The assumption of a skin surface marker representing underlying bone without formal account of soft tissue movement and marker displacement error being estimated and reported may leave results questionable. How much error and what amount may be tolerated in analysis must be determined and reported. Sources of error from within the camera setup include radial torsion due to imperfect lens shape, decentering or misalignment of lens components relative to the optical axis and thin prism distortions, which are generally due to improper lens and camera assembly (Furnee 1997).

The task of calibration is to obtain the best possible results based on the system at hand (Chiara et al., 2005). Improving on the design of the 2-D pilot study (Chapter 3) that used two 50 Hz cameras the current investigation in 3-D required at a minimum, two cameras placed between 60° and 120° or orthogonal to the point of interest (Woltring 1980). It was decided to employ four high speed cameras capturing 210 frames per second (fps). Justification was the increased number of cameras, capture speed and their position improves identification of skin marker tracking processes and the accuracy of the data produced. Increasing the number of cameras may result in camera redundancy, which describes a situation where two cameras must see a marker point to calculate 3-D coordinate data (Woltring 1980). If more than two cameras see the point the resultant calculation processes are improved. Also there may be less lost markers or dropouts in cluster designs and segment models which may be used in future investigations of knee joint kinematics. A common source of Camera error in lower-limb analysis occurs when markers are obscured by swinging or rotating limbs. This was evident from recorded video data obtained during the 2-D pilot study. Inter-marker distance must be definite with no overlap to prevent merging of markers during
tracking processes. The method of placing collinear markers on the rearfoot and
posterior heel of the shoe in the pilot study demonstrated practical limitations for
objective identification of critical events of the gait cycle through marker merging. In
barefoot trials the pilot study demonstrated deformation of the fibro fatty padding of
the heel at the heel strike event. This finding may impact on the inter marker distance
and orientation within the calibrated space for each gait cycle recorded.

Instrument errors have been classed as two types, instrument systemic errors (ISE) and
instrumental random errors (IRE) (Huiskes et al., 1985). ISE arise from calibration errors
inaccuracies and distortion (Woltring 1980; Huiskes et al., 1985). The magnitude of
error is dependent on the size of the calibrated volume and the position a marker
assumes within it (Gazzani 1993). IRE may be due to electric noise, marker flickering i.e.
the impression with which marker images are converted into points in the digitising
process itself which transforms marker image coordinates into their numerical values
(Furnee 1997; Della Croce 2000). Marker image and shape distortions may result from
motion velocity, or partially obscured markers merging with each other (Furnee 1997;
Pedotti and Ferrigno 1995). Blurring or the apparent streaking of a rapidly moving
marker occurs when a camera captures an image at speed, that image does not
represent a single instant in time. The image may look smeared along the direction of
relative motion.

4.1 Calibration
3-D coordinates are constructed from 2-D coordinates. Intrinsic data are generated
within the camera and consists of focal length, centre of the image in relation to the
lens and the distortion parameters of the lens. Extrinsic data relates to the position and
orientation of the camera to the other cameras and the global coordinate system (GCS).
Combined the intrinsic and extrinsic data are the calibration of the motion analysis
system. Setting up the SIMI motion™ analysis within the clinical setting required a
static global frame of reference or calibration frame (Figure 4.1). This frame defined
the capture area into three planes relative to an origin. The calibration frame was
recorded and marker coordinates determined. Once the calibration procedures for
each camera were completed the image coordinate data was stored for transformation
into 3-D coordinates.
There are several calibration 3-D reference objects. Most common are cube shape designs (Figure 4.1).

**Figure 4.1** Calibration frames left and right.

The accuracy of data produced from a motion analysis system is dependent on the accuracy of the calibration procedure. The greater number of coordinates on a calibration frame the greater the accuracy. This may range from eight to thirty on static frames (Woltring 1980).

**Accuracy:** Describes the deviation of the measured value from the true one (Abdel-Aziz and Kara 1971).

**Precision:** Describes the repeatability of measurements taken under identical circumstances (Windolf et al., 2008).

For laboratory based calibration work ± 0.1 mm is acceptable in all three planes (Richards 2008). The consensus amongst researchers is increased accuracy is achieved when,( Windolf et al., 2008; Woltring 1980; Furnee 1997; Huiskes et al., 1985; Ehara 1987; Everaert et al., 1999; Richards 2008).

1. Coordinate markers are large in number and evenly distributed in the calibration volume.
2. The greater the distance recorded markers are from the calibration frame, the more inaccurate the results become.
3. The calibration system should fill as much of the picture as possible.
4. Additional cameras increase the accuracy since errors are better offset due to a point being processed through multiple camera views.

The SIMI motion™ system was supplied with a large twelve point calibration frame to define the parameters of the spatial volume. To record whole body motion of a person
of dimension 2 m x 2 m x 2 m is required assuming a cube type is used. In large calibration areas a reduction in accuracy has been reported and there is an optimal marker size in motion analysis setups (Windolf et al., 2008). Utilisation of a large capture volume for lower limb analysis may compromise the data generated, therefore a custom made calibration frame was manufacture to reduce the volume. Considered as optimal the calibrated capture volume must be large enough to encompass the motion of the leg but not too large as to result in large areas of unused volume. The frame was approximately the length of the treadmill walkway, the width of the treadmill and the height of the subject sample mean pelvic iliac crest. Researchers have investigated Camera setup marker size and spatial volume (Windolf et al., 2008) concluding larger marker projection increases the number of pixels and enhances resolution. With this in mind 23 mm diameter markers were used in the calibration frame manufacture and segment models.

Comparison of motion systems has been investigated (Ehara et al., 1997; Everaert et al., 1999; Richards 2008). These comparisons include analysis of accuracy of static and or dynamic points, distance and or angles. Generally error ranges between 1.0-5.0 mm for linear displacements and 1.0-5.0° in angular displacements calculated when tracking the tibia segment across stance (Houck et al., 2004). Peak errors as high as 8° have been reported, using surface markers to track transverse plane orientation of the tibial segment during swing (Houlden et al., 1997). However in clinical assessments the focus is specific to closed kinetic chain events. These values where considered clinically significant and would jeopardise adoption of motion analysis into clinical decision making processes if errors were of this magnitude. Measurements taken in the clinical setting are not useful if clinicians cannot be confident in the results of those measurements (Barker et al., 2006). A term that describes level of confidence is trustworthiness (Rothstein and Echternach 1993). In terms of clinical significance an error of 5 mm in linear displacement would have little bearing if one was interested in the swing phase which may exceed 1000 mm. However in calcaneal inversion / eversion measurement an error of 5° represents a clinically significant error. Valid and reliable data is possible with computerised motion analysis (Evans et al., 1997). To determine the error associated with the SIMI motion system within our clinical setting, so that a decision concerning quantification of error could be established a rigid body
experiment was conducted and a review of current motion analysis systems provided (Table 4.1 (a) and 4.2(b)).

Table 4-1  (a) Comparison of motion analysis system specifications
Adapted from Bronner (2010)

<table>
<thead>
<tr>
<th>Company</th>
<th>System</th>
<th># markers</th>
<th>Lighting</th>
<th># cameras</th>
<th>Calibration</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ariel</td>
<td>Apas</td>
<td>256</td>
<td>Flood</td>
<td>Up to 6</td>
<td>Frame</td>
</tr>
<tr>
<td>BTS</td>
<td>Eliteplus</td>
<td>100</td>
<td>IR</td>
<td>Up to 16</td>
<td>Wand</td>
</tr>
<tr>
<td>Motion Analysis</td>
<td>HiRes</td>
<td>400</td>
<td>Red/IR</td>
<td>Up to 16</td>
<td>Cube/Wand</td>
</tr>
<tr>
<td>Peak Performance</td>
<td>Motus</td>
<td>500</td>
<td>Red/IR</td>
<td>6-12</td>
<td>Wand</td>
</tr>
<tr>
<td>Qualisis</td>
<td>Proreflex</td>
<td>150</td>
<td>IR</td>
<td>Up to 16</td>
<td>Frame/Wand</td>
</tr>
<tr>
<td>Oxford</td>
<td>Vicon</td>
<td>150</td>
<td>Red/IR</td>
<td>Up to 24</td>
<td>Frame/Wand</td>
</tr>
</tbody>
</table>

Table 4-2  (b) Comparison of motion analysis system specifications
Adapted from Bronner (2010)

<table>
<thead>
<tr>
<th>Company</th>
<th>Sampling Frequency</th>
<th>Accuracy (mm)</th>
<th>RMS error (mm)</th>
<th>RMS error (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ariel</td>
<td>60-400</td>
<td>1.0</td>
<td>1.51</td>
<td>2.11</td>
</tr>
<tr>
<td>BTS</td>
<td>50-120</td>
<td>1.0</td>
<td>4.46</td>
<td>4.29</td>
</tr>
<tr>
<td>Motion Analysis</td>
<td>60-240</td>
<td>0.1</td>
<td>1.49</td>
<td>1.76</td>
</tr>
<tr>
<td>Peak Performance</td>
<td>50-2000</td>
<td>0.6</td>
<td>1.77</td>
<td>3.77</td>
</tr>
<tr>
<td>Qualisys</td>
<td>1-1000</td>
<td>0.6</td>
<td>2.21</td>
<td>4.5</td>
</tr>
<tr>
<td>Oxford</td>
<td>60-1000</td>
<td>----</td>
<td>1.29</td>
<td>1.42</td>
</tr>
</tbody>
</table>

Determination of error associated with individual markers allows for identification of excessively noisy markers for subsequent adjustment or elimination, to facilitate measurements of higher accuracy. When applied to coordinate markers on calibration frames, tracking wands it may facilitate calibration (Anderson 2009). Overall justification for a custom made calibration frame over the genuine manufacturers frame and pyramid type designs was to reduce the capture volume significantly to a clinically useable size within the confines of our working environment and to increase
the validity of data generated. For example if one was interested in flexion and extension of a forefinger a 2 m x 2 m 2 m, calibration frame would be considered excessive and jeopardise the findings.

In conclusion measurement error can affect the result of motion analysis significantly. Instrument errors may be compensated and adjusted for within the laboratory. External setups, camera placement and calibration volume may be tuned for optimal gain. Internal configurations of the camera software may minimise optical distortion, image processing and data output noise through filtering and smoothing.

Failure to control these issues reflects in the prime objective of data collection. For this reason estimates of accuracy and precision of specific systems should become regular practise (Chiara et al., 2005).

4.2.1 Purpose
The aim of this study was to measure 3-D length of a validated rigid rod within a calibrated volume.

4.2.2 Specific aims
1. To determine the motion system error associated with dynamic 3-D motion analysis and a bespoke calibration frame.
2. To determine the optimal system setup within the laboratory setting for 3-D kinematic data collection of the lower limb.

4.3 Hypothesis #2
There is no mean length difference between the validated steel tibial model and the three-dimensional calculated length across trials.

4.3.1 Rationale for hypothesis #2
For an instrument to be useful within a clinical environment the user must be confident with the measure outcome. The present study design was constructed to determine the validity of 3-D motion length measurement within a custom made calibration frame. The custom made calibration frame was fabricated to encompass the immediate area of the treadmill and the approximate anterior superior iliac crest level of the subject sample in the pilot study. Previous research has recommended individual laboratories identify the amount of error within the respective laboratory (Windolf et al., 2008) in order to be better able to distinguish the outcome of intervention between
subject variability and instrument error. This is particularly relevant for outcome of tests conducted with in-shoe orthotics and without orthotics. The aim of this study was to establish the validity of motion analysis measurement of 3-D length in lower limb assessment with respect to our clinical laboratory. The objective measure was a 3-D length measurement of a known validated steel rod. It was hypothesised there is no difference between the validated mean length and the calculated mean length using the SIMI 3-D motion™ system.

4.3.2 Formal Hypotheses

\[ H_0 \] There is no difference between the validated rigid steel tibial model and the 3-D calculated length across dynamic trials.

\[ H_a \] \( < 380 \text{ mm} \)

\[ H_a \] \( > 380 \text{ mm} \)

\[ H_a \] Not equal to validated length

4.4 Methods

4.4.1 Apparatus

No human subjects were used in this study. A metal rigid body model was selected for all validation trials. The metal rod was produced from tensile steel and had been subject to three measurements of length by three independent examiners using a metre rule. The model was kept in the laboratory room at all times for the duration of the study to maintain temperature equilibrium and minimise metal expansion and contraction.

4.5 Materials

SIMI Motion 7.5 3D System. (2/3-D kinematics C3Dexport and DV2)

4 x Basler High Speed Camera. Camera Basler Pilot piA640-210gc. Industrial camera, 210fps at full resolution of 648x488pixels, 1/3” colour CCD sensor, progressive scan, lens adapter C-mount, GigE interface (Gigabit Ethernet, 1000MBits/s). Synchronisation via external trigger signal. Power supply via external power adapter.

4 x Pentax Lenses. ½” 8.0-48.0mm F1.2-16C manual zoom, manual focus, manual iris lenses.
Synchronisation Box. (External trigger of high speed cameras)

4 x Trigger Cables.

4 x CAT 6 High Performance Cables (10 metre)

4 x Tripods with 3-way PTZ quick release head.

4 x Basler Camera Tripod Mounting Adapter.

4 x Halogen light system. Tripod mounted 500w

1 x 1000 mm Aluminium rule.

SIMI Workstation High-Speed HS4. Workstation for 4 high-speed cameras, Intel Core 2 Duo, 2GB RAM, DVD-RW drive, individual hardware configuration. Microsoft Windows XP Professional.

4.5.1 Bespoke Calibration Cube.

The 3-D motion system was supplied with a twelve point calibration cube of dimension 2 m x 2 m x 2 m. Our laboratory specialism is specific to lower limb biomechanics. Use of a calibrated volume of this size would not allow the cameras to see the upper coordinate markers of the frame. The overall dimension of the laboratory was too small with the zoom out camera feature exhausted. A second problem arose using a small segment within a large volume. Greater accuracy is achieved when an object occupies as much of the calibrated space as possible (Windolf et al., 2008).

It was therefore necessary to construct a bespoke calibration cube to encompass a lesser spatial volume. This was made to approximate an upper axis height similar to a human pelvic iliac crest level. The laboratory room length was insufficient for an over-ground walk-way therefore the bespoke calibration cube was designed to contain the treadmill length and width (Figure 4.2). (A full description is provided in Appendix C)
4.5.2 The Rigid Model

The rigid tibial body segment was constructed from a length of 3 mm threaded tensile steel rod. This was placed in a bench vice and cut using a hand held hacksaw to 380 ± 0.01 mm in length. The length was checked against a one metre rule (Figure 4.3).

Twenty three millimetre retroreflective spheres were fixed to the opposite ends of the tibial rod. A piece of black string 1000 mm long was attached to the tibial rod 100 mm off centre using black adhesive tape. The remaining rod was covered with black tape until only the markers were exposed (Figure 4.4). (A full description is provided in Appendix C).
4.6 Laboratory setup
The bespoke calibration frame was positioned on the floor of the laboratory. Each corner of the frame was identified and marked on the laboratory floor using black adhesive tape. The tape remained on the floor throughout all data collection trials for the duration of the study. A spirit level was used to adjust to level the calibration frame and the treadmill belt. The calibration frame was removed temporarily and the treadmill belt was located between the areas marked with black tape. The hinged treadmill was raised to a closed upright position, and the calibration frame was returned to the original marked floor position (Figure 4.5).

Figure 4.5 Laboratory setup and marked location of the calibration frame.

Four 210 Hz high speed cameras were mounted on tripods and positioned at the right side of the treadmill in an umbrella placement format. A light source was positioned beside each camera parallel to each camera (Figure 4.6).
4.6.1 Camera setup
Each camera aperture was closed fully and slowly opened to produce a contrast with the background and markers.

4.6.2 Camera calibration
Calibration was carried out according to Simi-motion manufacturer recommendations. The origin of the calibration frame was positioned at the distal left hand corner side of the treadmill belt (Figure 4.7) (left).
A computer generated image of the GCS and origin at the intersection of XYZ axes was provided in (Figure 4.7) (right).

![Computer Image](image)

**Figure 4.7 The computer image of the calibration frame (Left) and the origin of the GCS at the intersection of the XYZ axes (Right).**

The axes of the calibration system were defined as,

- **X-axis** positioned perpendicular to the Y and Z axes pointing to the right representing movement in a medio-lateral direction.

- **Y-axis** positioned perpendicular to the X and Z axes pointing in the direction of movement representing an antero-posterior direction.

- **Z-axis** positioned perpendicular to the other two axes pointing upwards representing movement in a dorsal-proximal direction (Figure 4.7) (above right)

The calibration coordinates were recorded using exactly the same procedure and sequence for each camera. A direct linear translation (DLT-11) method was used (Table 4.3).
Table 4-3 Calibration coordinates

<table>
<thead>
<tr>
<th>X (m)</th>
<th>Y (m)</th>
<th>Z (m)</th>
<th>Point</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
<td>origin</td>
</tr>
<tr>
<td>0.000</td>
<td>0.542</td>
<td>0.000</td>
<td>P2</td>
</tr>
<tr>
<td>0.000</td>
<td>1.085</td>
<td>0.000</td>
<td>P3</td>
</tr>
<tr>
<td>0.545</td>
<td>1.085</td>
<td>0.000</td>
<td>P4</td>
</tr>
<tr>
<td>0.545</td>
<td>0.542</td>
<td>0.000</td>
<td>P5</td>
</tr>
<tr>
<td>0.545</td>
<td>0.000</td>
<td>0.771</td>
<td>P6</td>
</tr>
<tr>
<td>0.000</td>
<td>0.542</td>
<td>0.771</td>
<td>P7</td>
</tr>
<tr>
<td>0.000</td>
<td>1.085</td>
<td>0.771</td>
<td>P8</td>
</tr>
<tr>
<td>0.545</td>
<td>0.542</td>
<td>0.771</td>
<td>P9</td>
</tr>
<tr>
<td>0.545</td>
<td>0.542</td>
<td>0.771</td>
<td>P10</td>
</tr>
<tr>
<td>0.545</td>
<td>0.000</td>
<td>0.771</td>
<td>P11</td>
</tr>
</tbody>
</table>

4.6.3 Capture of 3-D calibration system
A single recorded video frame was selected at random from captured data of the calibration frame for assignment of the calibrated points.

4.6.4 Creation of a spatial model
A spatial model for the rigid tensile tibial rod was created within the SIMI-motion software. This consisted of a connection line between the two markers defined as points P1 and P3 respectively (Figure 4.8).

Figure 4.8 A spatial model of the tibial rod identified by retroflective markers P1 and P3.
4.6.5 Data collection
The tensile rod was representative of a modelled human tibia bone. The tibial rod was suspended on a string in the calibrated volume. The investigator swung the tibial rod in an anterior-posterior direction until the speed and angle of oscillation were considered in the investigators opinion similar to tibial motion velocity during walking. All testing data collection was carried out on the same day and in a single session.

Ten repetitive swing cycles were recorded at 210 frames per second (fps). Of the recorded cycles, two complete consecutive cycles were selected for analysis. A single cycle consisted of the tibial rod at a stationary instantaneous point most proximal, to the most distal point of swing (Figure 4.9).

![Figure 4.9 The tibial spatial model (A single dynamic tibial rod cycle traced in black)](image)

4.6.7 Phases of stance determination using colour shading
SIMI-motion 3-D™ was used to define the phasic cycles of forward and backward swing individually. The phases were defined in time and colour to differentiate measurement data. Green shaded areas represented forward swing and purple shaded areas backward swing (Graph 4.1).
Graph 4-1  Green colouring represented the forward swing phase and purple the backward swing phase.

4.6.8 3-D Motion analysis of the tibial rod
Tibial rod motion was tracked automatically through ten cycles until all points were defined.

4.6.9 Calculation of scaled 3-D coordinates
All cameras were selected for calculation of 3-D coordinates using normal raw data created during video capture.

4.6.9.1 Calculation of Angles/Distances
2/3-D distance coordinate data were created for points P1 and P3.

Angle/time data was created for P1 and P3 relative to the XYZ planes to determine if smoothing algorithms were necessary(Table 4.4).

Table 4-4 Angle time definitions

<table>
<thead>
<tr>
<th>Points</th>
<th>Angle</th>
</tr>
</thead>
<tbody>
<tr>
<td>P1-P3</td>
<td>Angle with the XY plane</td>
</tr>
<tr>
<td>P1-P3</td>
<td>Angle with the YZ plane</td>
</tr>
<tr>
<td>P1-P3</td>
<td>Angle with the XZ plane</td>
</tr>
</tbody>
</table>
4.6.9.2 Static still image measurement
Static 3-D still image measurements were recorded in millimetres at maximum proximal swing, mid swing and maximum distal swing points during each cycle.

4.6.9.3 Export data for analysis
Standard raw data for all cycles were normalised to time and exported as 2 / 3-D inter-marker distance. Angle/time series data and frequency data were exported to Microsoft Office Excel™ for further analysis. Descriptive statistics, mean values, standard deviations and variance were calculated.

4.7 RESULTS
The purpose of this study was to assess the precision and accuracy of the Simi 3-D motion™ analysis system within a clinical environment for lower limb treatment. A human tibia was modelled using a rigid body defined with two retroflective marker points.

The rigid model was manufacture to a length of 380 mm ± 0.1 mm. 3-D length of the tibia model was plotted for the complete video capture cycle, and was calculated over time. Four time series curves represented forward and backward movement of the tibial rod during swing phase (Graph 4.2).

Graph 4-2 mean length, maxima and minima of tibial model across trials.

Descriptive statistics, mean length and standard deviation across complete trials are given in (Table 4.5).
Table 4-5 Mean tibial rod dynamic length

<table>
<thead>
<tr>
<th>Phases</th>
<th>Rigid Tibia model manufactured length 380 ±0.1 (mm)</th>
<th>Rigid tibia model 3-D calculated length (mm)</th>
<th>Rigid tibia model 3-D length Standard deviation (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward</td>
<td>-</td>
<td>380.5</td>
<td>7.7 x 10^{-4}</td>
</tr>
<tr>
<td>Backward</td>
<td>-</td>
<td>380.6</td>
<td>7.1 x 10^{-4}</td>
</tr>
</tbody>
</table>

The results demonstrate no significant difference between the validated length and the mean 3-D length calculated across trials using the SIMI motion™ analysis system within the custom calibration frame.

Further analysis of the data included sub-division of the complete cycle into separate swing phases, of forward and backward 3-D motion. Table 4.6 demonstrates the results table for phase analysis following capture of video data.

Table 4-6 Cycle duration percentage and time data

<table>
<thead>
<tr>
<th>Name</th>
<th>Start (sec)</th>
<th>End (sec)</th>
<th>Start in Cycle</th>
<th>End in Cycle</th>
<th>Duration</th>
<th>Start/Cycle</th>
<th>End/Cycle</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward Swing</td>
<td>0.148</td>
<td>0.962</td>
<td>0.000</td>
<td>0.814</td>
<td>0.814</td>
<td>0.0%</td>
<td>50.0%</td>
</tr>
<tr>
<td>Backward Swing</td>
<td>0.962</td>
<td>1.776</td>
<td>0.814</td>
<td>1.628</td>
<td>0.814</td>
<td>50.0%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Forward Swing</td>
<td>1.776</td>
<td>2.571</td>
<td>0.000</td>
<td>0.795</td>
<td>0.795</td>
<td>------</td>
<td>------</td>
</tr>
<tr>
<td>Backward Swing</td>
<td>2.571</td>
<td>3.300</td>
<td>0.795</td>
<td>1.524</td>
<td>0.729</td>
<td>------</td>
<td>------</td>
</tr>
</tbody>
</table>

The mean tibial length for forward swing phase was 380.5 mm and 380.6 mm for backward swing. The Standard plus and minus deviation difference values for forward swing phases were 2.9 x 10^{-4} mm (Graph 4.3) and backward swing 5.5 x 10^{-4} mm (Graph 4.4) respectively.
Graph 4-3 Mean 3-D tibial length forward swing phases with plus and minus deviation.

Graph 4-4 Mean 3-D tibial length result for backward swing phases with plus and minus deviation.
Graph 4-5 Mean tibial model length across trials with SD error bars

Following a mean value finding the same as the validated model (Graph 4.5), the SD data for two forward swing phases and two backward swing phases were plotted over a normalised time (Graph 4.6).

Graph 4-6 Standard deviation of the forward and backward tibial model motion.
Motion capture of tibial model length deviation variance are summarised in table 4.7.

<table>
<thead>
<tr>
<th>Phase</th>
<th>Mean variability of SD (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward</td>
<td>$1.4 \times 10^{-4}$</td>
</tr>
<tr>
<td>Backward</td>
<td>$2.8 \times 10^{-4}$</td>
</tr>
</tbody>
</table>

Table 4-7 Mean variability of validated tibial model length across trials

Still frame image measurement figures 4.10 (a) and 4.10 (b) were selected for analysis at proximal, mid-swing and distal swing and were consistent with the length of the tibial model (Table 4.8). Analysis within the motion analysis software was to whole figures only.

Table 4-8 Still image measurement 3-D distance (mm)

<table>
<thead>
<tr>
<th>Proximal swing (mm)</th>
<th>Mid-swing (mm)</th>
<th>Distal swing (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>381.</td>
<td>381.</td>
<td>381.</td>
</tr>
<tr>
<td>381.</td>
<td>381.</td>
<td>381.</td>
</tr>
</tbody>
</table>

Figure 4.10 (a) 3-D distance calculation start point (b) 3-D distance calculation end point.

Resultant values of frequency across data rows range for forward and backward swing phases are given (Table 4.9 (a) and 4.10 (b). The frequency ranged from 379.1 mm to 380.1 mm
Table 4-9  (a) forward swing phases frequency data

<table>
<thead>
<tr>
<th>Swing 1 Forward Frequency (mm)</th>
<th>Swing 2 Forward Frequency (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>379.1</td>
<td>379.2</td>
</tr>
<tr>
<td>379.1</td>
<td>379.2</td>
</tr>
<tr>
<td>379.1</td>
<td>379.5</td>
</tr>
<tr>
<td>379.2</td>
<td>379.5</td>
</tr>
</tbody>
</table>

Table 4-10  (b) backward swing phase frequency data

<table>
<thead>
<tr>
<th>Swing 1 Backward Frequency (mm)</th>
<th>Swing 2 Backward Frequency (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>379.7</td>
<td>379.8</td>
</tr>
<tr>
<td>379.7</td>
<td>379.4</td>
</tr>
<tr>
<td>379.9</td>
<td>379.6</td>
</tr>
<tr>
<td>379.8</td>
<td>380.1</td>
</tr>
</tbody>
</table>

4.8 DISCUSSION

The findings of this study confirm the SIMI 3-D motion™ system is a clinically accurate measurement tool. The null hypothesis was accepted. There was no difference between the mean 3-D length of the tibial rod using either measurement method. This finding has a direct impact on healthcare. Clinical specialists utilising computerised motion analysis can be confident with the resultant measurement of 3-D length.

Trustworthiness is a term used to describe a level of confidence of a measure that is related to its degree of reliability and validity (D’Auria 2007).

The actual length of the validated tibial model (380 mm ± 0.1) was compared to the calculated 380 mm mean 3-D length of the model by the motion system. Within the software a provision is made for static frame by frame measurement. Analysis of instantaneous length was equal to the validated length.

A clinically accurate measurement is important. The structural alignment of the skeletal system has been linked with injury originating from malalignment. Osteoarthritis (OA) of the knee has been reported in frontal plane deviation of the femur and tibia. In clinical tibial varum, tibial valgus, genu valgum and genu varum loading of the tibial plateau is thought to be a significant factor in articular degeneration (Hunt et al., 2007). Pelvic obliquity and limb length discrepancy has also been suggested as a causative factor in lumbar spine and hip disease (Ashford and Shippen 2003). In the foot
malalignment of the rearfoot complex may exacerbate subtalar degenerative changes, tarsal coalition and toe deformity.

The tibial model constructed with known validity had a mean length of 380 mm ± 0.1 mm across forward and backward trials. The total variance of mean length did not exceed 1.6 mm across trials. Analysis of the average standard deviation variability did not exceed $2.8 \times 10^{-4}$ mm. These findings are supported by previous investigators. Some studies have examined accuracy and precision using motorised marker setup models, robotic arms and cylindrical weights fitted with markers to control extraneous variables. In studies with human subjects markers adhered to the tempromandibular joint have demonstrated similar accuracy of ± 1.5 mm (Baltali et al., 2008).

This study conducted within a clinical outpatient clinic may help with baseline evaluation pre and post operative outcomes and audit. There are several limitations the most obvious being the absence of a human subject. The focus of this paper was instrumental error which was to guide the clinician in error amount with motion analysis. If this error has been quantified within the setting then treatment of subjects can be better evaluated. The use of a human subject may introduce more variables and may mask instrumental error with artefact and variability. The use of marker size has been investigated (Windolf et al., 2008). The markers in this study were 23 mm in diameter. This size was used because in a clinical setting the emphasis is on time, appointment duration, cost and equipment ease of use. Several pilot studies were carried out with marker sizes ranging from 5 mm up to 23 mm. Initial trial runs proved unsuccessful with marker loss, markers not being seen by the cameras with constraints of the laboratory size restricting the setup. (Appendix Q)

4.9 CONCLUSION
The results of this investigation show that the SIMI Motion 3-D™ system is a clinically accurate measurement tool. Clinicians can be confident in dynamic 3-D length measurement. Our hypothesis was retained. There is no mean difference between a validated length and a calculated length using the Simi Motion™ system. Recommendation is based on,

1. A custom made calibration frame.
2. The object of interest filled as much as possible the calibrated volume.
3. All retroflective markers were seen by four cameras at all times.
4. 23 mm diameter retroflective markers.

In lower limb assessment a deviation of maximum 1.6 mm achieved in this investigation was acceptable. In overground walking variation in surface, camber and consistency of surface e.g. sand would see even greater requirement for variability, compensation and adaption from absolute values. It has been reported 10 %-15 % of the adult population have limb length discrepancies (Rush and Steiner 1946). Clinical treatment is often aimed at addressing the shortening by a conservative 50% demonstrating an acceptable error margin but this has been questioned (Michaud 1997), who proposed an additional 33% as necessary because the effect of a wedge is to work as a postero-anterior wedge which does not act the same as as full length raise. Taking into account compensation differences, footwear heel height and ankle equinus, Dominguez et al., (2006) reported the value of a heel wedge type raise should be increased only 23 %. Clearly the variance supports a 1.6 mm deviation in this study.

Future work may determine transverse plane angle error using multi point marker setup designs given reliability and repeatability studies have reported greater variability in transverse plane than sagittal plane motion and a standardised testing procedure being used across all laboratories to determine in-house error relative to other laboratories.
Chapter 5 DEFINITIONS OF ANATOMICAL CONVENTIONS

INTRODUCTION
Progressing from the previous chapters technical conventions are required to define the kinematics of human gait and spatial models used in this study. All conventions are based on current international standards from the International Society of Biomechanics and are modified for use in this research from reports of the Anatomical reference frame convention. Joint system conventions are adapted from the VAKHUM project report (2002).

5.1 Anatomical coordinate systems general direction of axes
The directions of the three Cartesian axes are mutually perpendicular. One axis is vertical and the directions of the remaining two horizontal axes are not usually contentious (Wu and Cavanagh 1995; Cappozzo et al., 1995).

- **Z-axis** the z-axis is vertical parallel to the field of gravity and points upwards.
- **X-axis** the x-axis is perpendicular to the z-axis and y-axis and is pointing to the right direction.
- **Y-axis** the y-axis is pointing in the anterior direction (direction of progression).

5.1.1 Anatomical Planes

- **Transverse plane:** This plane is perpendicular to the z-axis.
- **Sagittal plane:** This plane is perpendicular to the x-axis and parallel to gravity.
- **Frontal plane:** This plane is mutually perpendicular to both transverse and sagittal (Figure 5.1).
Figure 5.1 Anatomical coordinate axes and planes.

5.2 Femur Segment

Femur anatomical landmarks (Modified from Vakhum project 2002) (Figure 5.2).

**ft** - Lower third of Femur

**le** - Lateral epicondyle

**me** - Medial epicondyle

Figure 5.2 Femur anatomical landmarks.
5.2.1 Femur Anatomical Plane

Femur quasi-frontal plane

This plane is divided by me, le and ft

Femur quasi sagittal plane

This plane is orthogonal to the femur quasi frontal plane and contains both O_t (midway between me, le) and ft.

Femur quasi transverse plane

This plane is mutually perpendicular to both quasi frontal and quasi sagittal plane of the femur.

5.2.2 Femur Anatomical Frame

O_t The O_t point defines the origin of the anatomical frame of the thigh segment x_t,y_t,z_t this point is midway between me and le.

Y_t This axis is mutually perpendicular to the x_t and z_t axis and is pointing to the anterior.

X_t This axis is the femur quasi frontal plane and is perpendicular to the y_t axis with the positive direction pointing right.

5.3 Tibial / Fibula Segment

Tibial / fibula anatomical landmarks (modified from the Vakhum project 2002) (Figure 5.3).

Hf apex of head of Fibula (or me / le epicondyle of femur )

Two retroreflective sphere at end of tibial wand 50 mm.

Lm 10 mm above apex of lateral malleolus.

Mm 10 mm above apex of medial malleolus.
Figure 5.3 The tibial / fibula segment.

5.3.1 Tibial/Fibula anatomical plane

Tibial / fibula quasi-frontal plane-this is defined by hf lateral malleolus (Im) and midpoint Os between Im and mm (tibial wand 50 mm).

Tibial / fibula quasi-sagittal plane-this plane is orthogonal to the tibial/fibula quasi frontal plane and contains Os and hf and tw (parallel tw).

Tibial / fibula quasi-transverse plane- this plane is mutually perpendicular to the tibial/ fibula quasi-sagittal plane.

5.3.2 Tibial/Fibula Anatomical Frame

O_s  the O_s point defines the origin of the anatomical frame of the shank segment x_s,y_s,z_s and is located at the midpoint of the line joining Im and mm.

Ys-axis  this axis is mutually perpendicular to the xs-axis and the z-axis and is pointing to the anterior.

Zs-axis  this axis is lying in the tibial/fibula quasi frontal plane and is perpendicular to the y-axis with the positive direction pointing right.
5.4 Foot Segment

5.4.1 Foot Anatomical Frame

Foot anatomical landmarks were modified from the Vakhum project (2002) (Figure 5.4).

- **Ca**\textsubscript{7}  Posterior Superior Calcaneum.
- **Ca**\textsubscript{8}  Posterior Inferior Calcaneum.
- **Sm**  Dorsal aspect of the second metatarsal head.

![Figure 5.4 Foot anatomical frame landmark locations](image)

5.4.2 Foot Anatomical Plane

- **Foot quasi-transverse plane**- this plane is defined by Ca1, Ca2 and Sm.

- **Foot quasi-sagittal plane**- this plane is orthogonal to the foot quasi-transverse plane and contains both Ca1 Ca2 and Sm.

- **Foot quasi-frontal plane**- this plane is mutually perpendicular to the foot quasi transverse and quasi sagittal plane.
5.4.3 Foot Anatomical Frame

- $O_f$: the $Of$ point defines the origin of the anatomical frame of the foot segment $x_fy_fz_f$; this point is midway between $Ca_1$ and $Ca_2$ and $lm$ and $mm$.
- $Y_f$: this axis is mutually perpendicular to the $x_f$-axis and the $z_f$-axis and is pointing to the anterior.
- $Z_f$: this axis is defined by the intersection between the foot quasi-frontal and quasi-sagittal plane with positive direction upwards.
- $X_f$: this axis is lying in the foot quasi-transverse plane and is perpendicular to the $y_f$-axis with positive direction to the right.

5.5 Joint systems

In order to interpret data for lower limb motion, joint coordinate systems (JCS) are defined. The JCS positions are dependent upon anatomical landmark positions. The present study modelled the lower limb as three segments, which are considered as rigid bodies.

1. The Femur.
2. The Tibia and Fibula.
3. Foot (including the talus, calcaneus, navicular, cuboid, cuneiforms, metatarsals and phalanges).

A reference frame was fixed in each of these segments. The relative motion of these segments was defined by models of the Hip, Knee and ankle joints (Hilal et al., 1998).

5.5.1 Joint coordinate system

The JCS reported by Grood and Suntay (1983) has the advantage of being easily described in clinical terms and is independent of the order in which the rotational transformations are used (D’lima et al., 2000). The JCS corresponds to conventions using Euler angles in the following order Flexion, adduction-abduction and internal-external rotation of the moving segment coordinate system with respect to the fixed segment coordinate system.
5.6 Knee Joint Coordinate system  
The origin of this system is at the O₆ point

The flexion-extension is defined around the X₆-axis. Defining internal-external rotation around the Z₆-axis and abduction-adduction occurs around the floating axis, mutually perpendicular to the X₆-axis and Z₆-axis (Figure 5.5).

![Figure 5.5 Knee joint coordinate system adopted from Grood and Suntay (1983)](image)

Medio-lateral translation is measured along the X₆-axis, proximal-distal translation along the Z₆-axis and antero-posterior translation along mutually perpendicular floating axis.

5.7 Ankle Joint Coordinate system  
The origin of this system is at the O₇ point (Figure 5.6).

Plantar flexion-dorsiflexion is defined around the X₇-axis. 
Internal-external rotation about Y₇-axis. 
Adduction-abduction around the floating axis, which is mutually perpendicular to the X₇-axis and Z₇-axis.

Euler reconciled XYZ. 
Medio-lateral translation is measured along the X₇-axis proximal-distal translation along the mutually perpendicular floating axis.
• Flexion-extension is defined around the X_t-axis.
• Internal-external rotation around the Z_s-axis.
• Abduction-adduction around the floating axis mutually perpendicular to the X_t-axis and the Z-axis (reconciled with euler XYZ axes).
• Medio-lateral translation is measured along the X_t-axis.
• Proximal-distal translation along the Z_s-axis.
• Antero-posterior translation along mutually perpendicular floating axis.

Figure 5.6 Ankle joint coordinate system
Chapter 6 REPEATABILITY OF 3-D TIBIAL KINEMATICS DURING TREADMILL WALKING IN HEALTHY SUBJECTS

INTRODUCTION
There is inherent variability in human gait. In chapter three 2-D gait measurement demonstrated rearfoot angle data that conflicted with previous research findings at heel strike. The calcaneal angle at heel strike calculated in chapter three saw a barefoot calcaneal angle of 3.1° that reduced to 0.9° in shod trials. This was considered clinically significant. The addition of a 4.0° rearfoot wedge reduced the rearfoot calcaneal angle to 0.5° a reduction of 0.4°. This small amount was considered insignificant as it is not clinically measureable in an outpatient setting. Previous research has been focused on the effect of the in-shoe orthotic being realised at heel strike with the STJ being most proximal and the joints distal to the STJ following a sequence of events determined by the calcaneus and STJ at this point of ground contact (Root et al., 1971). The rejection of the hypothesis, a medial rearfoot wedge of 4.0° may change the calcaneal angle at heel strike raises questions concerning previous research based on this 2-D assessment. This may be due to difficulty in translating static wedged calcaneal angles or the outcome of dynamic rearfoot calcaneal angle within shod subjects, and the assumption of markers placed on the heel of the shoe representing any movement of the foot within the shoe. Furthermore the adoption of rigid single segment foot models, have no respect for the complexities of the forefoot multifunctional theories that are currently emerging with the use of 3-D motion analysis techniques and clinicians must be ready to adopt to change as new information becomes apparent and may provide better insight into the functional role of in-shoe orthoses.

Utilising 3-D motion analysis, chapter four described accuracy and validity of the SIMI motion system™ using a rigid model. This study moves forward to investigate the repeatability of transverse plane tibial rotation. The findings are important and required as gait variables collected are linked directly with treatment intervention.

Repeatability is the variation in measurements taken by a single person or instrument on the same item under the same conditions (Bland and Altman 1986). Previous research has reported kinematic data is reliable and repeatable within the same day (Kadaba et al., 1989; Diss 2001; Ferber et al., 2002; Queen et al., 2006). Generally sagittal plane kinematics may be more repeatable than frontal or transverse plane
kinematics (Mills et al., 2007). Poor between day repeatability has been identified arising from placement of markers and the speed of the subject in motion. Kadaba et al., (1989) proposed greater within day repeatability for sagittal plane angles resulted from a higher degree of neuromuscular control in the sagittal plane motions and suggested the lack of between day repeatability was due to repositioning of thigh and shank wands affecting the calculated joint angles. Other investigations conducted on gait variability often view it as a reflection of inconsistency in the central nervous muscular control systems (Hausdorf 2005). In studies of fall risks in older people variability is associated with the ability to maintain a repeatable walking cycle. The aim of this study was assess the repeatability of 3-D motion analysis of lower limb tibial motion in a group of healthy runners over a three week period to determine the magnitude of gait variability, so future work could determine independent values above and beyond average values. Specific aims were,

1. To determine the repeatability and subject variability associated with dynamic 3-D motion analysis of transverse tibial rotation.
2. To determine mean tibial angular data walking differences within subject trials and between subject trials over a three week period.

6.1 Rationale for hypothesis #3
Use of computerised technologies for determination of ntervention or assessment of pre-operative and post-operative treatment, rehabilitation or injury prevention programmes depend on the tool being reliable and valid. In-shoe orthoses are provided to patients to prevent, treat or enhance motion performance. The assumption is following initial prescription of an in-shoe device an objective measure provided by a 3-D gait system may demonstrate the effect of the device when measurements are repeated. In addition it may help identify the mechanisms for lower limb injury through examination of static and dynamic gait. The results must originate from the subject or the intervention and not the equipment. The reliability and validity was established with a rigid tibial model of validated length chapter four were the rigid tibial model of known length was calculated using the laboratory motion analysis setup and compared to a known validated validated measure . Reliability was established through successive measures of the rigid model through dynamic motion trials. The present study is focused on subject repeatability of tibial transverse plane angles in healthy subjects
over time. The hypothesis was healthy subjects would demonstrate a moderate reliability Intraclass correlation coefficient between 0.4 and 0.74 (Bland Altman 1986). of mean tibial transverse plane (internal/external) rotation in standardised barefoot treadmill trials measured with SIMI 3-D motion™ analysis.

6.2 Methods
6.2.1 Subject sample
Five subjects volunteered for participation in this study. All were healthy with no history of injury and did not wear orthotics. All subjects were familiar with treadmill walking and were classified as pronated foot types following assessment using the Foot Posture index (FPI) described by Redmond et al., (2006) (Appendix Q.) The FPI is a measure of standing foot posture and is a diagnostic clinical tool aimed at quantifying the degree to which a foot can be in a pronated supinated or neutral position. Table 6.1 provided the demographic data for the sample. The rationale for recruitment of an opportunistic sample of healthy subjects was determination of reliability and repeatability of 3-D tibial kinematics in the context of extrapolation to normal healthy individuals. Inclusion and exclusion criteria are listed in (Figure 6.1).

6.2.2 Ethical consideration
All subjects were given a subject information sheet and signed a consent form (Appendix N) approved by the ethics review boards of the University of Limerick.

Table 6-1 Subject demographics

<table>
<thead>
<tr>
<th>N=5</th>
<th>Sex (m/f)</th>
<th>Age (Yrs)</th>
<th>Height (mm)</th>
<th>Weight (Kg)</th>
<th>*FPI</th>
<th>**RCSP (°)</th>
<th>***NCSP (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean</td>
<td>2m/3f</td>
<td>33.2</td>
<td>1720</td>
<td>81.2</td>
<td>8.4</td>
<td>44.4</td>
<td>40.8</td>
</tr>
<tr>
<td>SD</td>
<td>-----------</td>
<td>5.4</td>
<td>.960</td>
<td>15.5</td>
<td>2.2</td>
<td>3.6</td>
<td>4.5</td>
</tr>
</tbody>
</table>

*Foot posture index **Relaxed calcaneal stance position ***Neutral calcaneal stance position.
Figure 6.1 Inclusion / Exclusion criteria

<table>
<thead>
<tr>
<th>Inclusion</th>
<th>Exclusion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy with no medical or surgical history.</td>
<td>A medical or surgical history.</td>
</tr>
<tr>
<td>Can walk on treadmill for duration of treadmill trials.</td>
<td>Unfamiliar with treadmill walking</td>
</tr>
<tr>
<td>Can attend for conducted trial sessions over three weeks.</td>
<td>Cannot attend for conducted trial sessions over three weeks.</td>
</tr>
<tr>
<td>Not wearing orthoses.</td>
<td>Wearing in-shoe orthoses.</td>
</tr>
<tr>
<td>No lower limb injury or history of lower limb injury.</td>
<td>A lower limb injury or history of lower limb injury.</td>
</tr>
<tr>
<td>Foot posture index score of 5 or greater</td>
<td>Foot posture index score of 5 or less</td>
</tr>
<tr>
<td>Over 18 years of age.</td>
<td>Under 18 years of age.</td>
</tr>
<tr>
<td>Signed consent.</td>
<td>No signed consent</td>
</tr>
<tr>
<td>Foot posture index</td>
<td>Foot posture index</td>
</tr>
</tbody>
</table>

6.3 Instrumentation

SIMI Motion 7.5 3D System. (2/3-D kinematics C3D export and DV2)

4 x Basler High Speed Camera. Camera Basler Pilot piA640-210gc. Industrial camera, 210fps at full resolution of 648x488pixels, 1/3” colour CCD sensor, progressive scan, lens adapter C-mount, GigE interphase (Gigabit Ethernet,1000MBits/s). Synchronisation via external trigger signal. Power supply via external power adapter.

4 x Pentax Lenses. ½” 8.0-48.0mm F1.2-16C manual zoom, manual focus, manual iris lenses.

Synchronisation Box. (External trigger of high speed cameras)

4 x Trigger Cables.

4 x CAT 6 High Performance Cables (10 metre)

4 x Tripods with 3-way PTZ quick release head.

4 x Basler Camera Tripod Mounting Adapter.

4 x Halogen light system. Tripod mounted 500w

1 x SIMI Workstation High-Speed HS4. Workstation for 4 high-speed cameras, Intel Core 2 Duo, 2GB RAM, DVD-RW drive, individual hardware configuration. Microsoft Windows XP Professional.
1 x Proform 725ex Treadmill.

1 x Bespoke Calibration Cube. The calibration cube was used previously in chapter 4. A detailed description is provided in appendix C.

1 x Tibial wand. A detailed description is provided in appendix C.

6 x Retroreflective markers

6.4 Procedures
All subjects were barefoot during recording. Subjects were requested to stand in a relaxed calcaneal stance position. Retroreflective markers were adhered to palpated anatomical landmarks. Skin marker placement was carried out by the same investigator throughout the study. Anatomical landmarks on the right leg of each subject were the fibular head, lateral malleoli, posterior calcaneus, inferior aspect one centimetre above the ground and the dorsal surface of the foot two centimetres proximal to the second metatarsal head. A tibial wand was positioned at mid-tibia height over the lateral compartment between the sagittal and frontal planes (Figure 6.2).

Figure 6.2 The anatomical skin marker setup.

6.4.1 Laboratory setup
A bespoke calibration was positioned on the floor of the laboratory. Each corner of the frame was identified and marked on the laboratory floor using black adhesive tape. The
tape remained on the floor throughout all data collection trials for the duration of the study. Using a spirit level the frame was levelled using the adjustable legs.

The calibration frame was removed temporarily and the treadmill belt was located between the areas marked with black tape. The hinged treadmill was raised to a closed upright position and the calibration frame was returned to the original marked floor position. The coordinates of the frame were then ready for digitisation (Figure 6.3).

![Figure 6.3 Location of the calibration frame, note the hinged treadmill to the right of picture and floor markings.](image)

Four 210 Hz high speed cameras were mounted on tripods and positioned at the right side of the treadmill in an umbrella placement format. A light source was positioned beside each camera parallel to each camera (Figure 6.4).
6.4.2 Camera setup
Each camera aperture was closed fully and slowly opened to produce a contrast with the background and markers.

6.4.3 Camera calibration
Calibration was carried out according to Simi-motion™ manufacturer recommendations. The origin of the calibration frame was positioned at the proximal left hand side of the treadmill belt (Figures 6.5 (a) and (b)).
6.4.4 Capture of calibration image.

Cameras were turned on and the calibration frame was recorded continually for three
seconds. Approximately 600 frames were recorded. A single recorded video frame
was then selected at random from captured data of the calibration frame for
assignment of the calibrated points.

The calibrated camera optimisation and principle point data were recorded for each
camera. Mean and standard deviation were accepted if below mean 0.30 and SD 0.12
mm.

The GCS axes orientations were defined in the following directions,

**X-axis** positioned perpendicular to the Y and Z axes pointing to the right representing
movement in a medio-lateral direction.

**Y-axis** positioned perpendicular to the X and Z axes pointing in the direction of
movement representing an antero-posterior direction.

**Z-axis** positioned perpendicular to the other two axes pointing upwards representing
movement in a dorsal-proximal direction (Figure 6.5b).

The calibration coordinates were recorded using exactly the same procedure and
sequence for each camera. A Direct linear translation (DLT-11) method was used (Table
6.2).
### Table 6-2 Calibration coordinates of the bespoke calibration frame

<table>
<thead>
<tr>
<th>X (m)</th>
<th>Y (m)</th>
<th>Z (m)</th>
<th>Point</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
<td>origin</td>
</tr>
<tr>
<td>0.000</td>
<td>0.542</td>
<td>0.000</td>
<td>P2</td>
</tr>
<tr>
<td>0.000</td>
<td>1.085</td>
<td>0.000</td>
<td>P3</td>
</tr>
<tr>
<td>0.545</td>
<td>1.085</td>
<td>0.000</td>
<td>P4</td>
</tr>
<tr>
<td>0.545</td>
<td>0.542</td>
<td>0.000</td>
<td>P5</td>
</tr>
<tr>
<td>0.545</td>
<td>0.000</td>
<td>0.000</td>
<td>P6</td>
</tr>
<tr>
<td>0.000</td>
<td>0.000</td>
<td>0.771</td>
<td>P7</td>
</tr>
<tr>
<td>0.000</td>
<td>0.542</td>
<td>0.771</td>
<td>P8</td>
</tr>
<tr>
<td>0.000</td>
<td>1.085</td>
<td>0.771</td>
<td>P9</td>
</tr>
<tr>
<td>0.545</td>
<td>0.542</td>
<td>0.771</td>
<td>P10</td>
</tr>
<tr>
<td>0.545</td>
<td>0.000</td>
<td>0.771</td>
<td>P11</td>
</tr>
<tr>
<td>0.545</td>
<td>0.000</td>
<td>0.771</td>
<td>P12</td>
</tr>
</tbody>
</table>

### 6.4.5 Creation of a spatial model

The skin markers were linked by connecting lines within the software to provide a segment spatial model representative of the lower limb (Figures 6.6 (a) and 6.6 (b)).

![Figure 6.6 (a) The stick model superimposed on the recorded image.](image-url)
6.6 (b) The spatial model tibial wand and GCS.

6.5 Data Collection
A test-retest design was selected. Subjects were requested to attend the laboratory on the same day at the same time for three consecutive weeks. All passive skin marker placements were carried out by the same investigator for the duration of the study. All markers were removed and reapplied for the second and third weekly data sessions. It was assumed the investigator was competent at identifying boney anatomical landmarks for application of skin surface markers given the investigator has examined some one thousand subjects in their clinic. The laboratory was re-calibrated between weekly trials.

Subjects were allowed a five minute acclimatisation period prior to data collection. The subjects were recorded for fifteen consecutive heel strikes ending with toe off phase of stance. Calculations were performed on ten consecutive trials for each subject across the three weeks (Table 6.3).

<table>
<thead>
<tr>
<th>Table 6-3 Data trials recorded weekly for three consecutive weeks</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Subject</strong></td>
</tr>
<tr>
<td>-------------</td>
</tr>
<tr>
<td>Subject 1</td>
</tr>
<tr>
<td>Subject 2</td>
</tr>
<tr>
<td>Subject 3</td>
</tr>
<tr>
<td>Subject 4</td>
</tr>
<tr>
<td>Subject 5</td>
</tr>
</tbody>
</table>
6.5.1 Gait analysis

Five passive reflective markers were placed on the anatomical landmarks by the investigator, an experienced Podiatrist with fifteen years practice in lower limb biomechanical assessment. The subject was instructed to stand on the treadmill. The investigator turned on the treadmill and adjusted the speed to 3.1 km/h (1.1m/sec). Subjects were instructed to walk looking toward the line of progression until instructed to stop.

Four high speed cameras with a sampling frequency of 210 fps recorded the dynamic 3-D coordinate trajectories of the skin markers. The collected data was analysed within the SIMI 3-D motion™ software.

6.5.2 Phases

SIMI-motion™ 3-D was used to define the phasic cycles of stance and swing individually. The phases were defined in time and colour to differentiate phasic measurement data. Green shaded areas represented the stance phase and purple shaded areas the swing phase. The events of heel strike and toe off were identified using frame by frame inspection of the time series graph (Graph 6.1).

Graph 6-1  Green colouring represented stance phase, purple colour swing phase.

6.5.3 3-D motion analysis of the tibial segment

Tibial motion was tracked automatically through ten cycles until all points were defined.

6.5.4 Calculation of scaled 3-D coordinates

All cameras were selected for calculation of 3-D coordinates using normal raw data created during video capture.
6.5.5 Calculation of Angles/Distances

2-D / 3-D distance coordinate data were created for the fibular head, tibial wand lateral malleoli and posterior calcaneus.

Angle/time data was created for the tibial wand relative to the YZ planes (Table 6.4).

Table 6-4  Angle and time data calculation point and plane

<table>
<thead>
<tr>
<th>Angle/time data calculations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Points</td>
</tr>
<tr>
<td>Motion</td>
</tr>
<tr>
<td>Tibial Rotation</td>
</tr>
<tr>
<td>Internal / External</td>
</tr>
<tr>
<td>P6-P7</td>
</tr>
<tr>
<td>Angle with the YZ plane</td>
</tr>
<tr>
<td>Rearfoot Angle Calc.</td>
</tr>
<tr>
<td>Inversion / Eversion</td>
</tr>
<tr>
<td>P8-P10</td>
</tr>
<tr>
<td>Angle with the XY plane</td>
</tr>
</tbody>
</table>

6.6 Export of data for analysis

Standard raw data for all cycles were normalised to time and exported as the mean values and standard deviations of the ten recorded trials for each subject. All data was analysed using Microsoft Office Excel™ and SPSS™ 13.5 for Windows™. Significance was set at 0.05 level. Data was presented as mean values and SD. Microsoft Office Excel™ was used for graphical analysis. SPSS version 13.5™ was used for calculation of descriptive statistics, mean values, standard deviations and Intraclass Correlation Coefficients (ICC). Further analysis included the exclusion of specific events, heel strike (HS) and toe off (TO).

The main measures of repeatability are within-subject random variation and retest correlation. Within-subject random variation can be expressed as a coefficient of variation (CV) or standard error of the measurement (SEM) (Hopkins 2000). Both represent error in measurement.

With-subject variation was calculated as the mean tibial angle value across three weeks for each subject and their respective standard deviation. Between weeks group variation was similarly calculated for all subjects on week one, week two and week three. The group standard deviation was calculated in the same way.

The CV is particularly useful for representing the repeatability of performance tests (Yavuzer 2008). The variation represented by typical error may be due to ISE or IRE effect. The CV is unit-less and unaffected by absolute values and has been used in several studies (Steinwender et al., 2000; Thorpe et al., 2005).
In this study the CV was defined as the ratio of the standard deviation (SD) to the mean value. This was calculated as a measure of within-session and between-session (Bland and Altman 1986). Retest (correlation of time) of lower limb kinematics was assessed and intraclass correlation coefficient (ICC) and confidence interval of the ICC, using one-way analysis of variance.

A high correlation indicates subjects are unaffected by the weekly time interval between trial test days and the laboratory setup. Accepted measures of high correlation in the literature are given in (Table 6.5).

<table>
<thead>
<tr>
<th>Value</th>
<th>Repeatability</th>
</tr>
</thead>
<tbody>
<tr>
<td>≥0.75</td>
<td>Excellent</td>
</tr>
<tr>
<td>0.4-0.74</td>
<td>Moderate</td>
</tr>
<tr>
<td>≤0.40</td>
<td>Poor</td>
</tr>
</tbody>
</table>

6.7 RESULTS
Nine healthy subjects actively involved in regular sports activities having no history of lower limb injury were analysed over a three week period. 3-D tibial motion was calculated across the stance phase beginning at heel strike and ending at toe off (Graph 6.2). The hypothesis was retained when within subject and between subject ICC values were above 0.4 and 0.74

![Within Subject Mean Tibial Angle And SD Bars Across Three Weeks](image)

Graph 6-2 mean tibial angle values and SD for subject trials including and excluding HS and TO. (HS = heel strike. TO = toe off. FF = foot flat. HL = heel lift.)
6.7.1 Within-Subject Mean Tibial Angle.

6.7.1.1 Heel strike to toe off analysis

Mean tibial angle data, SD and CV for subjects over three weeks from HS to TO are presented (Table 6.6).

The Intraclass correlation coefficient was calculated for between week effect across trials. A one-way random effects model, where people effects are random resulted in a moderate reliability ICC value of .549, (Table 6.7).

<table>
<thead>
<tr>
<th>Table 6-6 Within-subject mean tibial angle and SD. HS-TO</th>
</tr>
</thead>
<tbody>
<tr>
<td>week 1</td>
</tr>
<tr>
<td>(°)</td>
</tr>
<tr>
<td>s1</td>
</tr>
<tr>
<td>s2</td>
</tr>
<tr>
<td>s3</td>
</tr>
<tr>
<td>s4</td>
</tr>
<tr>
<td>s5</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Table 6-7 Intraclass Correlation Coefficient (HS –TO)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intraclass Correlation</td>
</tr>
<tr>
<td>Single Measures</td>
</tr>
<tr>
<td>Average Measures</td>
</tr>
</tbody>
</table>

6.7.2 Foot flat to toe off analysis

6.7.2.1 Between Week/Subject

Between week mean values an SD were calculated. The CV reduced for subjects 2, 4 and 5, (Table 6.8). The intraclass correlation coefficient was calculated again for between week effect across trials. A one-way random effects model with elimination of the heel strike event from trials increased the reliability ICC from moderate 0.549 to an excellent value of 0.786, (Table 6.9).
### Chapter Six Repeatability Of 3-D Tibial Kinematics

#### Table 6-8 Between week mean tibial angle across trials (foot flat-toe off)

<table>
<thead>
<tr>
<th></th>
<th>week 1</th>
<th>week 2</th>
<th>week 3</th>
<th>Mean</th>
<th>SD</th>
<th>CV%</th>
</tr>
</thead>
<tbody>
<tr>
<td>s1</td>
<td>-10.7</td>
<td>-8.5</td>
<td>-3.3</td>
<td>-7.5</td>
<td>3.8</td>
<td>-50.6</td>
</tr>
<tr>
<td>s2</td>
<td>-16.1</td>
<td>-14.1</td>
<td>-17.5</td>
<td>-15.9</td>
<td>1.7</td>
<td>-10.7</td>
</tr>
<tr>
<td>s3</td>
<td>-13.9</td>
<td>15.9</td>
<td>-11</td>
<td>-3</td>
<td>16.4</td>
<td>54.7</td>
</tr>
<tr>
<td>s4</td>
<td>-8.5</td>
<td>-13.9</td>
<td>-11.4</td>
<td>-11.2</td>
<td>2.7</td>
<td>23.9</td>
</tr>
<tr>
<td>s5</td>
<td>-11.2</td>
<td>-10.8</td>
<td>-12.9</td>
<td>-11.6</td>
<td>1.1</td>
<td>9.5</td>
</tr>
</tbody>
</table>

#### Table 6-9 Intraclass Correlation Coefficient (Foot flat –toe off)

<table>
<thead>
<tr>
<th></th>
<th>Intraclass Correlation</th>
<th>95 % Confidence</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lower Bound</td>
<td>Upper Bound</td>
</tr>
<tr>
<td>Single Measures</td>
<td>.550</td>
<td>-.015</td>
</tr>
<tr>
<td>Average Measures</td>
<td>.786</td>
<td>-.044</td>
</tr>
</tbody>
</table>

#### 6.7.3 Foot flat to heel lift analysis

##### 6.7.3.1 Between Week/Subject

A final ICC was calculated for between week effect across trials. A one-way random effects model where people effects are random was used. Eliminating the toe off event similarities could be seen between subjects CV values across the three tables (Tables 6.6, 6.8 and 6.10). The ICC result was a moderate reliability score of 0.648, (Table 6.11).

#### Table 6-10 Between week mean tibial angle across trials (Foot Flat-Heel Lift)

<table>
<thead>
<tr>
<th></th>
<th>week 1</th>
<th>week 2</th>
<th>week 3</th>
<th>mean</th>
<th>sd</th>
<th>CV%</th>
</tr>
</thead>
<tbody>
<tr>
<td>s1</td>
<td>-10.1</td>
<td>-8</td>
<td>-2.2</td>
<td>-6.7</td>
<td>4.0</td>
<td>-60.4</td>
</tr>
<tr>
<td>s2</td>
<td>-14.9</td>
<td>-11.8</td>
<td>-14.9</td>
<td>-13.8</td>
<td>1.7</td>
<td>-12.9</td>
</tr>
<tr>
<td>s3</td>
<td>-11.2</td>
<td>-13.7</td>
<td>-7.2</td>
<td>-10.7</td>
<td>3.2</td>
<td>-30.6</td>
</tr>
<tr>
<td>s4</td>
<td>-5.8</td>
<td>-11.3</td>
<td>-9</td>
<td>-8.7</td>
<td>2.7</td>
<td>-31.7</td>
</tr>
<tr>
<td>s5</td>
<td>-11.7</td>
<td>-10.7</td>
<td>-12.5</td>
<td>-11.6</td>
<td>0.9</td>
<td>-7.7</td>
</tr>
</tbody>
</table>
6.8 DISCUSSION
Adoption of motion analysis into the clinical environment brings with it inherent responsibility. Measurements obtained must be reliable and valid if used to determine treatment intervention or outcome (Chapter 4). Despite tight experimental methodologies there are remaining sources of error and inaccuracy which contribute to variability of measurement data (Growney et al., 1997). Repeatability is defined as, the associations between variables, an investigation where a group of cases are measured on more than one occasion (Polgar 1991). The aim of this study was to determine the repeatability of the SIMI motion™ system in a group of normal subjects over a three week period. The hypothesis was the SIMI motion™ system within the clinical laboratory would demonstrate moderate repeatability (0.4 to 0.74) correlation coefficient.

Variation between different occasions should be due to subject improvement or deterioration depending on any intervention and not due to random error (Hammer and Lindmark 2003). If measurement errors conceal gait deviations meaningful information will be lost. On the other hand if the limitations of the measurement are not known, small deviations may be considered meaningful thereby leading to over interpretation (Yavuzer 2008).

Repeatability is important for comparisons of data between laboratories (Biden et al., 1987; Brand and Croninshield 1981; Stebbins et al., 2006). Researchers are proposing standardisation of motion analyses techniques and reporting this will aid comparisons between centres (Stebbins et al., 2006) The research acknowledges the greatest repeatability is found in the sagittal plane which has the least variability, followed by the frontal and transverse planes, the transverse plane showing the most variability.

Table 6-11 Intra class Correlation Coefficient (Foot Flat – Heel Lift)

<table>
<thead>
<tr>
<th></th>
<th>Intraclass Correlation</th>
<th>95 % Confidence</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lower Bound</td>
<td>Upper Bound</td>
</tr>
<tr>
<td>Single Measures</td>
<td>.380</td>
<td>-.138</td>
</tr>
<tr>
<td>Average Measures</td>
<td>.648</td>
<td>-.572</td>
</tr>
</tbody>
</table>
Chapter Six Repeatability Of 3-D Tibial Kinematics

(Winter 1984; Moseley et al., 1996; keppe et al., 1990; Leardini et al., 2007; Ferber et al., 2002).

The repeatability of kinematic variables was investigated in running at standardised speeds, 3.5 m/s and 4.0 m/s. The results indicated all measured variables had acceptable between day and within trial repeatability with the coefficient of multiple correlation values being greater than 0.78 (Diss 2001). Schwartz et al., (2004) investigated between week and within subject trials in healthy subjects they reported two types of variability, intrinsic and extrinsic. Intrinsic (which occur naturally through within session or subject to subject variability) or extrinsic variability (from experimental errors and are subject to quality improvement measures). They suggested intrinsic errors could not be reduced but need to be measured at a baseline for comparison, whereas extrinsic variation arises from various methodological sources (Schwartz et al., 2004; Yavuser et al., 2008)

In this study the treadmill speed was standardised at 3.1 m/s. Previous investigators have reported the repeatability of spatial and temporal parameters are more variable at slower walking speeds (Bilney et al., 2003; Terrier et al., 2005). This may be a limiting factor in this study. The attempt to compensate in the design of this study was to allow a long period of acclimatisation to treadmill walking prior to recording data. It is acknowledged taking an average of a greater number of trials may improve further the reliability of measurements. Acceptable are six to eight trials (Besser et al., 1999). This design utilised a mean of ten barefoot trials for analysis. Conversely increasing the speed of gait may result in further gait alterations within the calibrated volume around the treadmill (Cavanagh et al., 1984; Mann et al., 1986). In this study running may have resulted in a greater stride length the leg would have extended outside the calibrated area and compromised the usefulness of the data. It was speculated keeping the calibrated volume tighter around the segments of interest helped the ICC result. Analysis of time series graphs demonstrated the greatest variability in the curve occurred at heel strike. When heel strike was removed the correlation coefficient increased from 0.549 to 0.786 a significant increase in ICC value as one may have anticipated. The toe off data was removed from the stance phase. The data trial for foot flat to heel lift was then calculated and an ICC value of (0.648) resulted. This finding lends support to current theory on heel strike variability in healthy subjects. The
marker may have fluctuated at heel strike to produce artefact but the use of a large marker surface are may aid system calculation hence the small difference (0.138).

A significant factor was the clinical area eliminates the possibility of an over-ground walkway necessitating the use of a motorised treadmill. Besides the natural variability in gait velocity of different individuals, at different speeds it was considered acceptable to this investigation given the subject sample was small, and the aim was to determine repeatability as reliability and validity of the SIMI™ motion system was assessed in chapter four. The treadmill may have controlled some of the natural walking variability between weeks. This was accepted as control of these factors exemplifies the effect of the device from the extraneous variables. There are several limitations with this study which must be acknowledged. Yavuser et al, (2008) described potential errors generated from marker placement, data collection and data processing. In agreement with their findings, the process of defining specific events of heel strike and toe off, foot flat and heel lift was difficult. In defence of this study the high speed cameras were recording at 210 fps which aided identification of these events.

6.9 CONCLUSION
The findings of this study were that the SIMI motion™ system resulted in moderate to excellent repeatability y scores for subject tibial rotation angles across three weeks. These findings suggest that with walking subjects tibial transverse plane angles are repeatable. With this in mind it was considered reasonable to base significant clinical decisions based on the data obtained during the gait trial. The ICC score was accepted for further investigation into the dynamic functional aspects of in-shoe devices on lower limb overuse injures within a clinical setting. This provided a baseline reference of gait variability in healthy subjects.

Future research may compare different calibrated volumes and multiple surface markers rather than the single point marker design used in this study.
Chapter 7 TIBIO-CALCANEAL 3-DIMENSIONAL KINEMATICS IN RUNNERS WEARING IN-SHOE ORTHOSES

INTRODUCTION
Research supports the use of in-shoe orthotics in treatment of lower limb injuries in runners (Landorf and Keenan 2000). Excessive or prolonged pronation has been accepted as a major causative factor in overuse injury (Stergio and Bates 1997). Decades of research and undergraduate teaching have adopted a model of an ideal osseous and structural alignment with respect to a vertical calcaneus and subtalar joint complex (Root et al., 1971; Ball and Afheldt 2000; Root et al., 1977). Malalignment of the subtalar joint in particular rearfoot valgus and a low medial longitudinal arch, a fully compensated forefoot varus, compensated rearfoot varus, pes planus or hypermobile foot have been identified as introducing excessive pronation during midstance phase such that the foot fails to resupinate adequately for an active toe off propulsive phase.

Coupling of the STJ and tibia amplifies transverse plane motion in excessive pronation and generates soft tissue stresses and further up the kinetic chain. Research has concluded excessive tibio calcaneal coupling range of motion secondary to excessive pronation is a significant factor in patellofemoral pain (McClay and Manal 1997).

In-shoe orthotics provide a mechanical method of bringing the ground up to the foot, eliminating or reducing STJ and or mid tarsal joint (MTJ) compensations in frontal plane anomalies of the forefoot and rearfoot relationship. This provides the foundation of the accepted hypothetical “neutral foot” that may be described as, a foot with an inherent harmony between boney structural alignment, and musculo-tendonous force. This study seeks to identify the exact contribution in-shoe orthoses may have on frontal and transverse plane rearfoot and tibial motion during walking gait.

7.1 Specific Aims
1. To describe the effect of in-shoe orthotics (specifically EVA and a standard generic polypropylene orthotic) on the 3-D kinematics of the rearfoot and tibia during treadmill walking.
2. To determine the extent of effect between a custom made high density EVA orthotic and a standard generic semi rigid polypropylene orthotic on the heel strike and midstance phases of gait.
7.2 Hypothesis #4
In-shoe orthoses will change the mean frontal plane (calc to ground) and transverse plane (tibial internal / external rotation) motion during heel strike and midstance phases of walking.

7.3 Methods
7.3.1 Subjects/Patients
Sampling/Definition of Population

A sample of 32 patients 30 male and 2 female were chosen for the study (Table 7.1). The patients had been in attendance at a podiatry sports clinic six months previous for in-shoe orthotic therapy. The patients had presented with lower limb overuse injuries. The sample was considered representative of the running population. All subjects were asymptomatic from lower limb complaints following initiation of orthotics six months previously and remained asymptomatic for the duration of the study.

Table 7-1 Patient demographics

<table>
<thead>
<tr>
<th></th>
<th>Male</th>
<th>Female</th>
<th>Age (yrs)</th>
<th>Height mm</th>
<th>Weight kg</th>
<th>**FPI</th>
</tr>
</thead>
<tbody>
<tr>
<td>MEAN</td>
<td>*30</td>
<td>*2</td>
<td>32.3 ± 9.5</td>
<td>1725 ± 84.6</td>
<td>76.3 ± 18.7</td>
<td>8.2 ± 3.6</td>
</tr>
</tbody>
</table>

*Number of subjects. **Foot posture index.

7.3.2 Ethical Consideration
All Patients were fit and healthy. A health status form, advice sheet and consent form was read and signed by all participants. The investigation was examined for protocol adherence to ethical principles involving human subjects and certified by the ethics committee review board of the University of Limerick. (Appendix N) Inclusion exclusion criteria are listed in (Table 7.2).
### Table 7-2 Inclusion/Exclusion Criteria

<table>
<thead>
<tr>
<th>Inclusion</th>
<th>Exclusion</th>
</tr>
</thead>
<tbody>
<tr>
<td>No known neurological or orthopaedic procedures</td>
<td>known neurological or orthopaedic procedures</td>
</tr>
<tr>
<td>Run ≥ 40Km per week</td>
<td>Run ≤ 40 km per week</td>
</tr>
<tr>
<td>Wear custom made EVA orthotics</td>
<td>Not wearing custom made EVA orthotics</td>
</tr>
<tr>
<td>Not under medical care</td>
<td>Under medical care</td>
</tr>
<tr>
<td>A history of lower limb running complaints</td>
<td>No history of lower limb running complaints</td>
</tr>
<tr>
<td>Foot Posture Index &gt; 5</td>
<td>Foot Posture Index &lt; 5</td>
</tr>
<tr>
<td>Not pregnant or planning</td>
<td>Knowing or possibility of pregnancy</td>
</tr>
<tr>
<td>≥ 18 years of age</td>
<td>≤ 18 years of age</td>
</tr>
<tr>
<td>Familiar with treadmill walking</td>
<td>Unfamiliar with treadmill walking</td>
</tr>
<tr>
<td>No learning disability</td>
<td>Learning disability</td>
</tr>
<tr>
<td>Have read and signed information sheet</td>
<td>Have not read and signed information sheet</td>
</tr>
<tr>
<td>Signed informed consent</td>
<td>No signed informed consent</td>
</tr>
<tr>
<td>Available on a study day</td>
<td>Not Available on a study day</td>
</tr>
</tbody>
</table>

### 7.4 Instrumentation

**SiMI Motion® 7.5 3D System.** (2/3-D kinematics C3D export and DV2)

**4 x Basler High Speed Camera.** Camera Basler Pilot piA640-210gc. Industrial camera, 210fps at full resolution of 648x488pixels, 1/3” colour CCD sensor, progressive scan, lens adapter C-mount, GigE interphase (Gigabit Ethernet, 1000MBits/s). Synchronisation via external trigger signal. Power supply via external power adapter.

**4 x Pentax Lenses.** ½” 8.0-48.0mm F1.2-16C manual zoom, manual focus, manual iris lenses.

**Synchronisation Box.** (External trigger of high speed cameras)
4 x Trigger Cables.

4 x CAT 6 High Performance Cables (10 metre)

4 x Tripods with 3-way PTZ quick release head.

4 x Basler Camera Tripod Mounting Adapter.

4 x Halogen light system. Tripod mounted 500w

1 x 1000 mm Aluminium rule.

SIMI Workstation High-Speed HS4. Workstation for 4 high-speed cameras, Intel Core 2 Duo, 2GB RAM, DVD-RW drive, individual hardware configuration. Microsoft Windows® XP Professional.

7.4.1 The Amfit System®

The Amfit® computer aided design and computer aided manufacture (CAD-CAM) fabrication system uses an electronic scanning (Digitising) process and CAD-CAM production technology to carve custom foot orthoses.

The system consists of:

7.4.2 Contact Digitiser

The digitiser uses a measuring process to capture an impression of a foot across an intrinsic flat plane. An array of sensor pins controlled by the investigator and a computer microprocessor elevated the pins to capture a weight-bearing or semi-weight bearing contour of a foot (Figure 7.1).

Figure 7.1 A contact digitiser, note the pins that capture the arch contour

The captured contour was then adjusted using Amfit software® to compensate for the mechanical anomalies within the foot. The saved file was then transferred to the milling machine for manufacture.
7.4.3 Milling/Carving
The milling machine consisted of an electric motor and burr controlled by a computer (Figure 7.2). The carving process operated on an XYZ axes. Data files from the digitiser were loaded onto the hard drive for manufacture.

Figure 7.2 The amfit milling system and computer processor.

7.4.4 Ethyl Vinyl Acetate Blank (EVA)
The in-shoe orthotic was carved from a pre-cut foot shaped block of EVA. The EVA used for all patients was high density rated shore 65 (Figure 7.3).

Figure 7.3 The EVA blank after carving to shape (front) and preformed device (rear)

7.4.5 Bespoke calibration cube.
A bespoke calibration frame was used in the study (Figure 7.4). The dimensions and coordinate point data was described previously (Validation of 3-D motion length measurement using a rigid model chapter four). A description of the manufacture of the calibration frame is provided in Appendix C.
Chapter Seven 3-D Tibio-Calcaneal Kinematics In Runners Wearing Orthoses

Figure 7.4 A bespoke calibration cube designed and manufactured for the investigation.

7.4.6 The Calcaneal Cluster Model

The custom made cluster model was constructed from an EVA block with two non co-linear marker spheres (Figures 7.5 (a) and (b)). An aluminium shaft 40 mm was positioned to allow the cluster to protrude through the posterior heel counter of the subject’s footwear (Figure 7.15b). Mounting of the cluster to the subject’s calcaneus (calc) was by an adhesive semi flexible mounting plate. A detailed description of the calcaneal cluster construction is provided (Appendix C).

Figure 7.5 (a) The calcaneal cluster and skin surface mount. (b) Lateral view

7.4.7 The Tibial Wand

A custom made tibial wand was fabricated. A detailed description of the fabrication procedure was provided (Appendix C). The tibial wand consisted of three components 1) a skin mounting plate 2) a shaft 3) a retroflective sphere (Figures 7.6 (a) and (b)).
Chapter Seven 3-D Tibio-Calcaneal Kinematics in Runners Wearing Orthoses

7.4.8 Treadmill

The treadmill chosen was designed such that it was folded into an upright position to accommodate the calibration frame. On release the belt was then within the calibrated space (Figure 7.7).

Figure 7.7 The treadmill was manufactured such that it could be raised to accommodate the calibration frames by lifting in the direction of arrows.

7.4.9 Orthoses

Two orthotic types were used in this study. A custom milled full length device made of EVA and preformed polypropylene ¾ length device with a 6° low medial arch profile (Figures 7.8 (a) and (b)).
7.4.9.1 Footwear

Footwear was standardised for all subjects. The right shoe was modified to accommodate the calcaneal cluster mount (Figure 7.9). A detailed description of manufacture is provided (Appendix C).

Figure 7.9 Footwear modifications to accommodate the calcaneal cluster.

7.5 Procedures

7.5.1 Subject Marker placement

Subjects were asked to stand upright in a barefoot condition. The investigator a podiatrist with substantial clinical experience palpated bony anatomical landmarks (Table 7.3). Markers were adhered to the respective location (Figures 7.10. (a) and (b)).
Table 7-3 Names and positions of markers used in lower limb model

<table>
<thead>
<tr>
<th>Marker name</th>
<th>Position</th>
<th>Segment</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee1*</td>
<td>Medial femoral condyle</td>
<td>Femur</td>
</tr>
<tr>
<td>Knee2*</td>
<td>Lateral femoral condyle</td>
<td>Femur</td>
</tr>
<tr>
<td>LF</td>
<td>Head of fibula</td>
<td>Tibia</td>
</tr>
<tr>
<td>LMal</td>
<td>Lateral malleolus</td>
<td>Tibia</td>
</tr>
<tr>
<td>MMal*</td>
<td>Medial malleolus</td>
<td>Tibia</td>
</tr>
<tr>
<td>TibWand</td>
<td>Mid antero-lateral tibia</td>
<td>Tibia</td>
</tr>
<tr>
<td>Calc7</td>
<td>Posterior superior calcaneus</td>
<td>Foot</td>
</tr>
<tr>
<td>Calc8</td>
<td>Posterior inferior calcaneus</td>
<td>Foot</td>
</tr>
<tr>
<td>2ndMet</td>
<td>Second metatarsal head</td>
<td>Foot</td>
</tr>
<tr>
<td>Femur</td>
<td>Lateral lower Femur</td>
<td>Femur</td>
</tr>
</tbody>
</table>

*Markers were used in static trial only and removed for dynamic trials.

Figure 7.10 A lateral camera view (a) posterior camera view (b)

7.5.2 Laboratory setup

The bespoke calibration frame was positioned on the floor of the laboratory. Each corner of the frame was identified and marked on the laboratory floor using black adhesive tape. The tape remained on the floor throughout all data collection trials for the duration of the study. A spirit level was used to adjust to level the calibration frame and the treadmill belt.
The calibration frame was removed temporarily and the treadmill belt was located between the areas marked with black tape. The hinged treadmill was raised to a closed upright position and the calibration frame was returned to the original marked floor position (Figure 7.11).

Figure 7.11 Location of the calibration frame with treadmill in raised position.

Four high speed cameras were mounted on tripods and positioned at the right side of the treadmill in an umbrella placement format. A light source was positioned beside each camera parallel to each camera (Figure 7.12).

Figure 7.12 The laboratory setup and cameras placement in umbrella layout.
7.5.3 Camera setup

Each camera aperture was closed fully and slowly opened to produce a contrast with the background and markers.

7.5.4 Camera calibration

Calibration was carried out according to the recommendations of the Simi-motion™ manufacturers. The origin of the calibration frame was positioned at the proximal left hand side of the treadmill belt and was identified in multiple camera views using yellow tape (Figure 7.13).

![Figure 7.13 Calibration frame with origin marked in yellow tape.](image)

The axes of the calibration system were define as,

**X-axis** positioned perpendicular to the Y and Z axes pointing to the right representing movement in a medio-lateral direction.

**Y-axis** positioned perpendicular to the X and Z axes pointing in the direction of movement representing an antero-posterior direction.

**Z-axis** positioned perpendicular to the other two axes pointing upwards representing movement in a dorsal-proximal direction.

The calibration coordinates were recorded using exactly the same procedure and sequence for each camera. A direct linear translation (DLT-11) method was used (Table 7.4).
Table 7-4 Calibration coordinates shown in metres

<table>
<thead>
<tr>
<th>X (m)</th>
<th>Y (m)</th>
<th>Z (m)</th>
<th>Point</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
<td>origin</td>
</tr>
<tr>
<td>0.000</td>
<td>0.542</td>
<td>0.000</td>
<td>P2</td>
</tr>
<tr>
<td>0.000</td>
<td>1.085</td>
<td>0.000</td>
<td>P3</td>
</tr>
<tr>
<td>0.545</td>
<td>1.085</td>
<td>0.000</td>
<td>P4</td>
</tr>
<tr>
<td>0.545</td>
<td>0.542</td>
<td>0.000</td>
<td>P5</td>
</tr>
<tr>
<td>0.545</td>
<td>0.000</td>
<td>0.000</td>
<td>P6</td>
</tr>
<tr>
<td>0.000</td>
<td>0.000</td>
<td>0.771</td>
<td>P7</td>
</tr>
<tr>
<td>0.000</td>
<td>0.542</td>
<td>0.771</td>
<td>P8</td>
</tr>
<tr>
<td>0.000</td>
<td>1.085</td>
<td>0.771</td>
<td>P9</td>
</tr>
<tr>
<td>0.545</td>
<td>0.542</td>
<td>0.771</td>
<td>P10</td>
</tr>
<tr>
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<td>0.542</td>
<td>0.771</td>
<td>P11</td>
</tr>
<tr>
<td>0.545</td>
<td>0.000</td>
<td>0.771</td>
<td>P12</td>
</tr>
</tbody>
</table>

7.5.5 Capture of 3-D calibration system
A single recorded video frame was selected at random from captured data of the calibration frame for assignment of the calibrated points. Calibration was repeated on each day of data collection before subjects arrived for participation.

7.5.6 Creation of a spatial model
A spatial model for the right lower limb was created within the SIMI-motion™ software. This consisted of connection lines linking the retroreflective markers defined as point’s number one to ten (Figures 7.14 (a) and (b)).
7.5.7 Data collection

Three days were allocated at the clinic setting for data collection. Subjects selected a suitable time and day to accommodate their everyday commitments. Each day consisted of the assessment of approximately ten subjects.

A repeated measures design was used to determine mean differences between the subjects as four independent groups. The subjects were exposed to all four conditions during data collection trials (Table 7.5).

Table 7-5 Four independent variable factor levels tested in the experiment

<table>
<thead>
<tr>
<th>Baseline</th>
<th>Baseline/Experiment</th>
<th>Experiment</th>
<th>Experiment</th>
<th>Experiment</th>
</tr>
</thead>
<tbody>
<tr>
<td>Static trial*</td>
<td>Barefoot walk</td>
<td>Shod walk</td>
<td>Custom</td>
<td>Standard</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>orthotic walk</td>
<td>orthotic walk</td>
</tr>
</tbody>
</table>

*Trial was used as baseline to normalise data to neutral stance position baseline.

7.5.7.1 Dependent Variables

For the purpose of this study calcaneal angle and tibial angle at heel strike and midstance were chosen as dependent variables.
Heel strike was defined as;

- The initial point where the heel struck the ground, terminating when the first metatarsal head contacted the ground.

Midstance was defined as;

- The initial point from foot flat through the stance phase, terminating with heel lift on the same foot.

7.5.8 Static trials

Subjects were requested to stand on the treadmill. The investigator palpated the subtalar joint into a neutral position. The SIMI motion™ system was used to record all marker point data in this position. A single frame was used to calculate angular data of the calc cluster and tibial wand.

7.5.9 Dynamic trials

The treadmill speed was set to 3.1 km/h (1.1 m/s) and the subject was allowed a five minute acclimatisation period. Fifteen consecutive steps of the right leg were recorded in each condition of barefoot, shod, custom orthotic and standard orthotic (Table 7.6). Ten consecutive steps were selected from the fifteen recorded for analysis. The calc cluster was detached from the skin mounting plate on the calc so footwear and orthoses could be donned or removed between experimental conditions (Figures 7.15 (a) and (b)).

| Table 7-6 The sequence of motion analysis trial data capture |
|-----------------|---------------|-------------|----------------|----------------|
| Trial 1         | Trial 2       | Trial 3     | Trial 4        | Trial 5        |
| Static Neutral  | Barefoot Only | Shod Only   | Custom Orthotic| Standard Orthotic |
Figure 7.15 (a) Calc cluster mounting plate adhered to the subject (b) Calc cluster on the shod subject allowing footwear and orthoses to be donned or removed between trials.

The procedure was repeated for each subject in the same sequence by the same investigator on each study day, (Table 7.6).

7.6 Phases

SIIMI-motion™ 3-D was used to define the cyclic data generated from walking trials into phases and events. Heel strike, foot flat, mid-stance heel lift and swing phases were identified and marked with individual colours to aide recognition. This was standardised throughout the study (Graph 7.1).

Graph 7-1 Phasic events were colour coded to aide recognition.
7.6.1 3-D motion analysis of the clusters and skin mounted markers
Marker motion was tracked automatically through ten step cycles. A single step cycle began at initial heel strike on the right leg and terminated with heel strike on the same right leg. Skin marker tracking continued until all points were defined.

7.6.2 Calculation of scaled 3-D coordinates
Four cameras were selected for calculation of 3-D coordinates using normal raw data created during video capture.

7.6.3 Calculation of angular data
2-D /3-D coordinate data was created for each point, the ten retroreflective markers. For the purpose of the study, calculation of angular data was created and defined as the angle between a plane and a straight line defined by two points. Angle/time series data was created for points three and four (Tibial wand) and points seven and eight (Calc) relative to the GCS (Table 7.7).

Table 7-7 Angle/time data calculations

<table>
<thead>
<tr>
<th>Marker</th>
<th>Points</th>
<th>Angle</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibial Wand</td>
<td>P3-P4</td>
<td>Angle with the YZ plane</td>
</tr>
<tr>
<td>Calcaneus</td>
<td>P7-P8</td>
<td>Angle with the XY plane</td>
</tr>
</tbody>
</table>

7.6.4 Calculation of data for export.
A calculation template was created to facilitate data analysis prior to export. The template format followed the same sequence as the video recording of trials described previously. During the data export a corrupted file resulted in the loss of a single subject tibial transverse plane data at the heel strike phase only. Given the data collected for the subject unaffected included calcaneal angle at heelstrike and midstance and tibial angle data at midstance the available data was included.
Data envelopes for export consisted of:

- The mean values across ten trials for each subject in each condition.
- The standard deviation across ten trials for each subject in each condition.
- The maximum value minus the minimum value across ten trials for each subject in each condition.
- The plus and minus one standard deviation value.

All standard raw data rows were stretched to the same duration and exported as a text file format angle/time series data, to Microsoft Office Excel® for further analysis.

7.6.5 Within-subject angular mean values
Ten trials were recorded for each subject in each experimental condition. Within-subject mean data was generated as an ensemble mean trial of ten trials in each experimental condition.

7.6.6 Within-subject range of motion
Ten trials were recorded for each subject in each experimental condition. Within-subject range of motion was generated from an ensemble mean trial of ten trials in each experimental condition. The minimum ensemble mean trial value was subtracted from the maximum ensemble mean trial value within each experimental condition to determine the range of motion.

7.6.7 Between-subject angular mean values
Ten trials were recorded for each subject in each experimental condition. A single ensemble mean value was generated across ten trials for each subject group in each experimental condition.

7.6.8 Between-subject range of motion
Ten trials were recorded for each subject in each experimental condition. Between-subject group range of motion was generated from a group ensemble mean trial of ten trials. The minimum ensemble mean trial value was subtracted from the maximum
ensemble mean trial value within each experimental condition to determine the range of motion.

### 7.7 RESULTS

The purpose of this study was to investigate any effect observed in the lower limb kinematics of subjects wearing prescribed custom made in-shoe EVA orthoses and a preformed generic in-shoe device. Presented are main effect graphs for each group and intervention. Also provided are data graphs that did not reach statistical significance but may indicate a significant trend toward clinical significance of custom orthoses versus generic preformed. Negative values represent calcaneal inversion. The results were,

#### 7.7.1 Calc angles at heel strike.

Between group mean calc angles at heel strike were significantly different to barefoot trials (p<0.001) (Graph 7.2).

![Graph 7.2](image)

**Graph 7-2** shod, custom and standard orthoses mean calc angles at heel strike were significantly different than barefoot trials (p<0.001)

Calc angles at heel strike in shod, custom and standard orthotic conditions resulted in a significant difference of 10.3° for each condition when compared to barefoot trial 2.3° mean angles (Table 7.8).
Table 7-8 Mean calc angle at heel strike in each experimental trial

<table>
<thead>
<tr>
<th>(n=32)</th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (*)</td>
<td>2.3 ± 5.0</td>
<td>-9.2 ± 7.2</td>
<td>-10.2 ± 5.6</td>
<td>-10.3 ± 6.6</td>
</tr>
</tbody>
</table>

There were no significant mean calc angle differences at heel strike between shod and custom orthoses (p=0.262), custom and standard orthoses (p=0.976) or shod and standard orthoses (p=0.445).

The hypotheses accepted were,

Shod walking changed the mean calcaneal angle at heel strike when compared to barefoot mean walking calcaneal angles at heel strike. (p<0.001)

Custom orthotics change the mean calcaneal angle at heel strike when compared to barefoot mean walking calcaneal angles at heel strike. (p<0.001)

Standard orthotics change the mean calcaneal angle at heel strike when compared to barefoot mean walking calcaneal angles at heel strike. (p<0.001)
7.7.2 Calc angles at midstance

Between group mean calc angles at midstance were also significantly different in shod, custom and standard orthoses over barefoot trials (p<0.001) (Graph 7.3).

![Graph 7-3](image)

**Graph 7-3**  shod, custom and standard orthoses mean calc angles at midstance were significantly different than barefoot trials. (p<0.001)

There were no significant differences between shod and custom orthoses though these groups had the smallest p-value. (p=0.225) Custom and standard orthoses were (p=0.787) and shod and standard orthoses (p=0.470) at midstance. (Appendices L)

The mean calc angles at midstance were reduced from a barefoot angle of -1.2° to a maximum angle shod angle of 4.0°. Custom and standard orthoses did not exceed the shod mean value (4.0° SD 4.9°) and differed by 0.3° (Table 7.9).

**Table 7-9 Between group mean calc angles at midstance**

<table>
<thead>
<tr>
<th>(n=32)</th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (°)</td>
<td>-1.2 ± 2.5</td>
<td>4 ± 4.9</td>
<td>3.1 ± 4.0</td>
<td>3.4 ± 4.6</td>
</tr>
</tbody>
</table>
The hypotheses accepted were,

Shod walking changed the mean calcaneal angle at midstance when compared to barefoot mean walking calcaneal angles at midstance. \((p<0.001)\) (Graph 7.3).

Custom orthoses change the mean walking calcaneal angle at midstance when compared to barefoot mean calcaneal walking angles at midstance. \((p<0.001)\) (Graph 7.3).

Standard orthoses change the mean walking calcaneal angle at midstance when compared to barefoot mean walking calcaneal angles at midstance. \((p<0.001)\) (Graph 7.3).

**7.7.3 Tibial angles at heel strike.**
Between group mean tibial angles at heel strike were not significantly different than barefoot trials \((p>0.05)\) at heel strike (Graph 7.4).

![Between Group Mean Tibial Angle At Heel Strike With SD Bars (n=31)](image)

**Graph 7-4 Shod, custom and standard orthotic tibial angles at heel strike were not significantly different than barefoot trials at heel strike. \((p>0.05)\)**

The greatest mean tibial angular change over barefoot trials (mean 6.4° SD 2.6°) was achieved by the custom orthotic trials (mean 5.1° SD 5.2°) although this did not reach statistical significance \((p=0.228)\) (Graph 7.5).
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Graph 7-5  The greatest mean tibial angular change achieved at heel strike between groups (p>0.05)

This was the smallest p-value achieved between conditions. No significant mean tibial angle difference resulted between trials or between orthotic conditions (Table 7.10).

Table 7-10 Between group mean tibial angle at heelstrike with SD bars

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean(°)</td>
<td>6.4 ± 2.6</td>
<td>5.7 ± 2.7</td>
<td>5.1 ± 5.2</td>
<td>5.9 ± 2.6</td>
</tr>
</tbody>
</table>

All alternative hypotheses were rejected (Appendices L) and null hypotheses retained.

The retained hypotheses were,

H₀ There is no mean tibial angle difference between shod and barefoot walking at heel strike.

H₀ There is no mean tibial angle difference between barefoot walking at heel strike and wearing custom orthoses at heel strike while walking.

H₀ There is no mean tibial angle difference between shod walking at heel strike and wearing custom orthoses at heel strike while walking.

H₀ There is no mean tibial angle difference between shod walking at heel strike and wearing standard orthoses at heel strike while walking.
There is no mean tibial angle difference between custom orthoses walking at heel strike and wearing standard orthoses at heel strike while walking.

7.7.4 Tibial angles at midstance.

Between group shod, custom and standard orthotic mean tibial angles at midstance were significantly different than barefoot trials (p<0.001) (Graph 7.6).

Graph 7-6 Shod, Custom and Standard orthoses trial tibial angles were significantly different than barefoot trials at midstance. (p<0.001)

Mean barefoot values (9.4° SD 2.6°) were reduced the greatest amount in custom and standard orthoses (mean 6.6° SD 2.6°). The shod condition resulted in a lesser angular change (mean 6.8° SD 2.6°) (Table 7.11).

Table 7-11 Between group mean tibial angle and SD bars at midstance

<table>
<thead>
<tr>
<th>(n=32)</th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (°)</td>
<td>9.4 ± 2.6</td>
<td>6.8 ± 2.6</td>
<td>6.6 ± 2.6</td>
<td>6.6 ± 2.6</td>
</tr>
</tbody>
</table>

The smallest p-value achieved was between shod and custom orthoses. (p=0.156) This did not reach significance at .05 level (Graph 7.7).

Shod and standard orthoses were p=0.285 and custom and standard orthoses p=0.853.
The hypotheses accepted were,

\( H_a \) Shod walking changes the mean tibial angle at midstance when compared to barefoot mean walking tibial angles at midstance \((p<0.001)\) (Graph 7.6).

\( H_a \) Custom orthoses change the mean walking tibial angle at midstance when compared to barefoot mean tibial walking angles at midstance \((p<0.001)\) (Graph 7.6).

\( H_a \) Standard orthoses change the mean walking tibial angle at midstance when compared to barefoot mean walking tibial angles at midstance \((p<0.001)\) (Graph 7.6)

**7.7.5 Calcaneal range of motion at heel strike.**

Shod, custom and standard orthotic, calcaneal range of motion at heel strike were not significantly different between groups \((p>0.05)\) (Graph 7.8).
Graph 7-8 Mean calc range of motion was not significantly different between group trials at .05 level.

The mean calc range of motion was reduced from a barefoot value of 2.1° to 1.3° in shod trials. Though not statistically significant this was the largest effect on range of motion. Custom and standard orthotic trial mean ranges of motion were 1.8° and 1.7° respectively (Table 7.12).

Table 7-12 Between group mean calc range of motion and SD at heel strike

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>(n=32) Mean(*)</td>
<td>2.1 ± 2.3</td>
<td>1.3 ± 3.1</td>
<td>1.8 ± 2.7</td>
<td>1.7 ± 2.3</td>
</tr>
</tbody>
</table>
The smallest p-value was reached between barefoot and shod trials (p=0.70) (Graph 7.9).

**Graph 7-9** Barefoot and shod trials resulted in the greatest reduction in range of motion but did not reach significance at the .05 level.

All alternative hypotheses were rejected (Appendices L) and null hypotheses retained.

The retained hypotheses were,

**H₀** There is no mean calcaneal range of motion difference between shod and barefoot walking at heel strike.

**H₀** There is no mean calcaneal range of motion difference between barefoot walking at heel strike and wearing custom orthoses at heel strike while walking.

**H₀** There is no mean calcaneal range of motion difference between shod walking at heel strike and wearing custom orthoses at heel strike while walking.

**H₀** There is no mean calcaneal range of motion difference between shod walking at heel strike and wearing standard orthoses at heel strike while walking.

**H₀** There is no mean calcaneal range of motion difference at heel strike when walking in custom orthoses or walking in standard orthoses.
7.7.6 Calcaneal range of motion at midstance.

Between group mean calc range of motion in shod, custom and standard orthoses trials were significantly different than barefoot trials at midstance (p<0.001) (Graph 7.10).

The greatest mean range of motion difference was reach between barefoot (mean 1.3° SD 1.3°) and shod trials (mean 4.5° SD 4.6°) p<0.001.

The mean difference between custom orthoses and shod trials was 0.9° and custom versus standard orthoses was 0.4° (Table 7.13).

Table 7-13 Between group mean calcaneal range of motion and SD at midstance

<table>
<thead>
<tr>
<th>Between Group Mean Calc Range Of Motion And SD At Midstance (n=32)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Barefoot</td>
</tr>
<tr>
<td>Mean (°)</td>
</tr>
<tr>
<td>Barefoot</td>
</tr>
<tr>
<td>Shod</td>
</tr>
<tr>
<td>Custom Orthotic</td>
</tr>
<tr>
<td>Standard Orthotic</td>
</tr>
</tbody>
</table>

There was no significant mean difference between shod, custom and standard orthoses. The smallest p-value was achieved between shod and custom trials (p=0.122) (Graph 7.11).
Graph 7-11  The smallest p-value between group trials shod and custom (p=0.122) failing to reach significance at .05 level.

The Hypotheses accepted were,

\( H_a \) Shod walking changes the mean calcaneal range of motion at midstance when compared to barefoot walking mean calcaneal range of motion at midstance.

\( H_a \) Custom orthoses change the mean calcaneal range of motion at midstance when compared to barefoot walking mean calcaneal range of motion at midstance.

\( H_a \) Standard orthoses change the mean calcaneal range of motion at midstance when compared to barefoot walking mean calcaneal range of motion at midstance.

All rejected hypotheses are detailed in appendix L.
7.7.7 Tibial range of motion at heel strike.

Between group mean tibial range of motion at heel strike was significantly different than barefoot walking trials when subjects were shod, in custom or standard orthoses (p<0.001) (Graph 7.12).

**Graph 7-12**  Mean tibial range of motion at heel strike was significantly different in shod, custom and standard orthoses when compared to barefoot trials (p<0.001).

The mean tibial range of motion at heel strike in barefoot trials was 1.1° and reduced five-fold to a mean range of 0.2° in shod and custom orthoses (Table 7.14).

**Table 7-14 Between group mean tibial range of motion and SD at heel strike**

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean(°)</td>
<td>1.1 ± 1.0</td>
<td>0.2 ± 0.3</td>
<td>0.2 ± 0.2</td>
<td>0.2 ± 0.1</td>
</tr>
</tbody>
</table>

There was no significant mean tibial range of motion difference between shod and custom orthoses, (p=0.412) (Graph 7.13). Shod and standard orthoses (p=0.141) or custom and standard orthoses (p=0.315).
Chapter Seven 3-D Tibio-Calcaneal Kinematics In Runners Wearing Orthoses

**Graph 7-13** Shod and custom mean tibial range of motion at heel strike was not significant at .05 level.

The hypotheses accepted were,

\( H_a \) Shod walking changes the mean tibial range of motion at heel strike when compared to barefoot walking mean tibial range of motion at heel strike.

\( H_a \) Custom orthoses change the mean tibial range of motion at heel strike when compared to barefoot walking mean tibial range of motion at heel strike.

\( H_a \) Standard orthoses change the mean tibial range of motion at heel strike when compared to barefoot walking mean tibial range of motion at heel strike.

All rejected hypotheses are detailed in appendices L.

**7.7.8 Tibial range of motion at midstance**

Between group mean tibial range of motion was significantly different in shod, custom and standard orthoses when compared to barefoot trials at midstance \( (p<.001) \) (Graph 7.14).
Graph 7-14  Mean tibial range of motion was significantly different in shod, custom and standard orthoses than barefoot trials at midstance (p<0.001).

The mean tibial range of motion at midstance was reduced in barefoot trials from 1.7° to 0.8° in custom and standard orthoses. Shod range of motion decreased 0.6° from barefoot trial values (mean 1.7° SD 1.1°) (Table 7.15).

Table 7-15 Between group mean tibial range of motion at midstance

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom</th>
<th>Standard</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (°)</td>
<td>1.7 ± 1.1</td>
<td>1.1 ± 1.1</td>
<td>0.8 ± 0.9</td>
<td>0.8 ± 0.9</td>
</tr>
</tbody>
</table>

There was a significant tibial range of motion mean difference between shod and custom orthoses trials (p=0.048) (Graph 7.15).
Chapter Seven 3-D Tibio-Calcaneal Kinematics In Runners Wearing Orthoses

Graph 7-15 The mean tibial range of motion at midstance was significantly different between shod and custom orthoses trials at the .05 level.

Shod and standard orthoses did not reach significance (p=0.129) (Graph 7.16). Custom and standard orthoses trials were not significantly different (p=0.910) (Graph 7.17).

Graph 7-16 Mean tibial range of motion at midstance in shod and standard orthoses trials were not significantly different at .05 level.
The hypotheses accepted were,

$H_a$ Shod walking changes the mean tibial range of motion at midstance when compared to barefoot walking mean tibial range of motion at midstance.

$H_a$ Custom orthoses change the mean tibial range of motion at midstance when compared to barefoot walking mean tibial range of motion at midstance.

$H_a$ Standard orthoses change the mean tibial range of motion at midstance when compared to barefoot walking mean tibial range of motion at midstance.

$H_a$ Custom orthoses change the mean tibial range of motion at midstance when compared to shod walking mean tibial range of motion at midstance.

All rejected hypotheses are detailed in appendix L.
7.8 DISCUSSION
This study aimed to investigate any effect orthoses may have on pronation or tibial axial rotation during walking. The subjects selected for inclusion were athletes with a history of overuse running injuries. Orthoses had been prescribed at least six months previous to participation in the study. Orthoses were manufactured from the same materials and techniques though the individual prescriptions were unique to each subject. A standard generic pre-moulded device was also investigated. The findings suggest footwear is the most important factor influencing pronation, tibial rotation and range of motion at heel strike and midstance.

7.8.1 Calc angle at heel strike
Barefoot calc angle mean values at heel strike were 2.3°. In shod trials the calc angle increased by almost 11.1° and the introduction of orthotic gained a further 1.0°. This increase over barefoot walking was considered a clinically significant reaching statistical significance at the 0.05 level, (Table 7.16).

7.8.2 Calc angle at midstance
Progressing through to midstance table 7.17, the barefoot mean calc angle was -1.2°, a reduction of approximately 3.1° from the mean angle at heel strike. This was considered clinically good indication of data through the progression of the cycle giving support to the experimental setup. The introduction of footwear altered the barefoot calc angle by approximately 5.0°, a clinically and statistically significant effect. Orthotic trials reduced the calc angle by less than 1.0°. This finding for orthotic effect was disappointing as change of 1.0° was not considered clinically significant finding. Statistically significance was reached over barefoot mean calc angles and in orthotic trials at midstance only, (Table 7.17).

7.8.3 Calc range of motion at heel strike
The barefoot mean calc range of motion at heel strike was 2.1°. No significant clinical or statistical significance was generated in range of motion in trials with footwear or orthoses. The generated ranges of motion values were typically less than 1.0° (Table 7.18). This was not considered a clinical dilemma as the heel strike event involves a heel strike transient between 10 and 20 milliseconds and following this the foot moves into foot flat and the whole process occurring by 8% of the gait cycle (Whittle 1991).
The orthotic function at the heel strike phase may be primarily aimed at inducing structural and angular change that was achieved and clinically and statistically significant at heel strike. Range of motion at heel strike in clinical terms may be more significant as we progress through the stance phase.

7.8.4 Calc range of motion at midstance
At midstance the calc range of motion in shod and orthotic trials were statistically significant over barefoot trials. Orthoses did not reach statistical significance over shod trials and between orthotic mean calc ranges of motion values were less than 0.5° (Table 7.19). Clinically the custom orthotic performed better though the achieved value was small however both orthoses achieved a reduction in range of motion over shod only values. This finding supports past reports of motion control through midstance.

7.8.5 Tibial angle at heel strike
The mean barefoot tibial angle at heel strike was 6.4° (Table 7.16). Shod and orthotic trials did not reach statistical significance at the 0.05 level over barefoot trials. When subjects wore shoes and custom orthotics the tibial heel strike angle changed by 1.3° a small subjective clinical finding (Table 7.16). When taken in a clinical context the tibial transverse angle at heel strike may be considered irrelevant as injury is proposed through range of motions occurring as the gait cycle progresses.

7.8.6 Tibial angle at midstance
The mean tibial barefoot angle at midstance was 9.4°. The introduction of footwear reduced the tibial angle by 2.6° (Table 7.17). Surprising the tibial angle reduced by only a further 0.2° when orthoses were placed in footwear. It was anticipated the greatest effect would be observed at this phase of the gait cycle within the tibial segment transverse motion. Though this was the case, the effect of the orthoses was minimal with footwear having the greatest clinical and statistically significant effect overall.
7.8.7 Tibial angle range of motion at heel strike
In shod trials a mean tibial angle range of motion at heel strike was 1.1°. This range of motion reduced almost five-fold to 0.2°, a value that was not considered clinically useful or significant, although statistically this value was reached significance at the 0.05 level (Table 7.18). There was no range of motion difference between shod and orthoses or between orthotic types. This was not a clinically disappointing result as a possible reason may be due to the motion being absorbed by calc at heel strike and the large angular change at heel strike (Table 7.16).

7.8.8 Tibial range of motion at midstance
Barefoot mean tibial range of motion was 1.7°. Shod trials reduced the barefoot mean range of motion by 0.6°. This was considered too small to be clinically useful and not significant though the reduction reached statistical significance at the 0.05 level.
Orthoses trials were similarly statistically significant over the barefoot range of motion with custom orthoses achieving further statistical significance over shod trials (Table 7.19).

| Table 7-16 Calc and tibia at heel angle strike summary of significance p<0.05 |
|-------------------------------|-------------------|-------------------|
| Condition | Calc angle at heel strike (°) | Tibial angle at heel strike (°) |
| Barefoot | 2.3 | 6.4 |
| Shod | -9.2* | 5.7 |
| Custom orthotic | -10.2* | 5.9 |
| Standard orthotic | -10.3* | 5.9 |

*Significant over barefoot at the 0.05 level.

| Table 7-17 Calc and tibia angle at midstance summary of significance p<0.05 |
|-------------------------------|-------------------|-------------------|
| Condition | Calc angle at midstance (°) | Tibial angle at midstance (°) |
| Barefoot | -1.2 | 9.4 |
| Shod | 4.0* | 6.8* |
| Custom orthotic | 3.1* | 6.6* |
| Standard orthotic | 3.4* | 6.6* |

*Significant over barefoot at the 0.05 level.
Table 7-18 Calc and tibia RoM at heel strike summary of significance p<0.05

<table>
<thead>
<tr>
<th>Condition</th>
<th>Calc RoM at heel strike (°) Mean</th>
<th>SD</th>
<th>Tibial RoM at heel strike (°) Mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Barefoot</td>
<td>2.1</td>
<td>2.3</td>
<td>1.1</td>
<td>1.0</td>
</tr>
<tr>
<td>Shod</td>
<td>1.3</td>
<td>3.1</td>
<td>0.2*</td>
<td>0.3</td>
</tr>
<tr>
<td>Custom orthotic</td>
<td>1.8</td>
<td>2.7</td>
<td>0.2*</td>
<td>0.2</td>
</tr>
<tr>
<td>Standard orthotic</td>
<td>1.7</td>
<td>2.3</td>
<td>0.2*</td>
<td>0.1</td>
</tr>
</tbody>
</table>

*Significant over barefoot at the 0.05 level.

Table 7-19 Calc and tibia RoM at midstance summary of significance p<0.05

<table>
<thead>
<tr>
<th>Condition</th>
<th>Calc RoM at midstance (°) Mean</th>
<th>SD</th>
<th>Tibial RoM at midstance (°) Mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Barefoot</td>
<td>1.3</td>
<td>2.3</td>
<td>1.7</td>
<td>1.1</td>
</tr>
<tr>
<td>Shod</td>
<td>4.5*</td>
<td>3.1</td>
<td>1.1*</td>
<td>1.1</td>
</tr>
<tr>
<td>Custom orthotic</td>
<td>3.4*</td>
<td>2.7</td>
<td>0.8**</td>
<td>0.9</td>
</tr>
<tr>
<td>Standard orthotic</td>
<td>3.8*</td>
<td>2.3</td>
<td>0.8*</td>
<td>0.9</td>
</tr>
</tbody>
</table>

*Significant over barefoot at the 0.05 level.  **Significant over shod and barefoot conditions at 0.05 level.

7.8.9 The clinical relevance of the findings

Excessive or prolonged pronation has been identified as a causative factor in many overuse injuries including shin splints, Achilles tendinitis, plantar fasciitis and iliotibial band friction syndrome (Clement et al., 1981). Underpinning the treatment and management of excessive pronation is the use of specialised running shoes, taping techniques and in-shoe orthotics. The latter are prescribed with varying degrees of rearfoot or forefoot posts to bring the ground to the foot or redirect ground reaction forces. The subtalar joint and its orientation historically coupled with tibial rotation may be controlled by the introduction of one or a combination of these three techniques to help aide recovery. An accepted fact of lower limb biomechanics is pronation related injury occur only during weight-bearing activities primarily walking and running. At the moment of heel strike the calc everts and the subtalar joint pronates, the talus abduct everts and plantarflexes. This combination of triplanar motion absorbs the ground reaction forces at impact.

The findings of this investigation reinforce the importance of appropriate footwear. There were significant footwear effects over barefoot trials at heel strike and midstance.
in calc and tibial angles. Typically these angles and ranges of motion were between 2-3°. Though at heel strike mean calc angles were 10.0° with a mean SD of 6.0° over barefoot mean values in shod and orthotic trials (Table 7.16). This was surprising given the footwear selected for use the study was a baseline neutral shoe design with no inbuilt manufacturers motion control or support system. And the orthoses only induced a 1.1° change over the shod trials. The small but clinically significant results are in keeping in part with those reported previously by Donoghue et al., (2008) who investigated orthoses in achille’s tendon injury. They concluded calcaneal eversion was increased despite the use of in-shoe devices. This reported as their main finding is also an outcome of this investigation and justifies further investigation. A possible explanation for the observed increase in calc eversion in the study by Donoghue et al., (2008) may be the duration of hours subjects had worn their respective orthoses. This was not reported hence the findings may be complicated by material compression and lack of a wedge effect to invert or resist calcaneal eversion, which may be a factor when EVA foams are used. This may also be a limitation of this study as subjects had been wearing EVA orthoses for six months and remained asymptomatic. This may be a reason for the similar outcome of increased calc eversion in this study. The effects of treadmill on foot placement, abduction angles or transverse tibial rotation was not provided. The results of this study demonstrate a mean transverse plane internal tibial rotation reduction of 2.8°. Clinically, though a small angular change it was an unexpected finding in the light of the increased calcaneal eversion.

Researchers have reported the process of removing a portion of the heel counter of the shoe to provide a window for attaching skin markers or a cluster mount to the calcaneus reducing the overall performance of the shoe. It was accepted this may be a limitation of our study but anticipate any effect was negligible. The results show, shod trials were statistically significant in almost all trials over barefoot conditions despite footwear modification. Taken with respect to previous research literature the use of sandals, and wedged outsole modification limit the internal validity of their findings. As a clinical observation, recreational runners, walkers through to elite sports men and women, the findings of sandal studies cannot be extrapolated to these populations. A similar problem arises with markers placed on the heel of footwear. The findings of such research may mask the pure motion of the heel within the shoe exaggerating false
positive outcomes or vice versa. The present study used a bench grinding technique to minimise disruption of the structural integrity of the shoe. It was anticipated the calcaneal movement represented raw calcaneal motion.

A previous investigation attempted to control tightness of fit of the footwear when worn by the subjects. Knots were tied strategically in the laces so when footwear removed and replaced between trials there would be consistence. No information was given concerning the type of shoe or foot. A cavoid foot may have a higher mid tarsal joint subsequently the lace may be too loose for the next subject. This study did not constructively plan for tightness of fit though all footwear was fitted by the same investigator for all subjects between all trials. This was necessary as the calc cluster and mount had to be fitted into the shoe carefully to avoid contact with the rear of the shoe. The calcaneal cluster was mounted by the same investigator using the same procedure that was repeated throughout all trials. Therefore the tightness of fit was consistent with the investigators discretion.

The model accepted for decades and taught at undergraduate level defines the “ideal foot” as pronating and supinating around a neutral vertical calcaneus (Root et al., 1977). This has been questioned (Hamill et al., 1999). The finding of this study support the theory behind an orthotic altering the calc angle at heel strike and the tibial angle at midstance. Also supported, the hypothesis that orthoses change the mean range of motion of the calc at midstance, and the tibia at heel strike and midstance. There was no significant mean change of the calc range of motion at heel strike. A reason for this may be explained because the calc heel strike is an event, a single point in time. Secondly the calc cluster mount was at heel strike prone to vibration and unwanted movement due to ground reaction forces. This has been referred to as “wand dance”. Initial pilot tests were conducted to determine the optimum length of the mounting rod. The chosen length was sufficiently long to protrude from the attachment at the calc base plate through the aperture in the shoe, but not too long to minimize wand dance artefact. (Appendix Q)

Researchers may filter raw data in a smoothing process. All data used in this study was raw data unfiltered. In the authors opinion initial pilot trials on the data time series curves appeared sufficiently smooth to justify no filter process. (Appendix Q)
be due to the high speed camera specification 200 fps and the walking pace of subjects 1.1 m/sec over faster running studies were smoothing may be necessary.

The tibial wand mounted on the antero-lateral shank was also subject to wand dance and noise artefact. Though this is a limitation of the study, the range of motion and angular values are small and in agreement with previous research. Cornwall and McPoil (1995) employed soft orthoses and inflatable arch fillers and concluded shoes alone or an accommodative orthotic reduce maximum tibial rotation compared with barefoot walking with the overall reduction achieved was less than 2.0°. In this study tibial rotation range of motion values were 1.1° and 0.2° at heel strike and standard deviation of 0.1°. The author acknowledges this may be instrumental error and not representative of the segment motion, although these values did reach statistical significance over barefoot trials (Table 7.18 and 7.19).

The choice of retroreflective sphere was 23 mm diameter. This was selected as a fundamental theme of this investigation was to introduce 3-D motion analysis into the main stream clinical setting. The use of larger markers was chosen to facilitate the automatic tracking process. Previous researchers have used similar size markers to assess the accuracy and precision of motion capturing systems within systematic parameters (Windolf et al., 2008) but most laboratory based research has employed smaller 6 mm to 12 mm diameter spheres (Stergio and Bates 1997; Baltali et al., 2008) with several investigators omitting to include this information (Collins et al., 2009; De Wit et al., 2000; Ashford and Shippen 2003; Ferber et al., 2005). A consensus of conclusion with respect to marker sizes used is accuracy and precision must be established within each laboratory setting (Windolf et al., 2008).

7.9 CONCLUSION
Clinicians must be aware of the importance of footwear effect. This may be further enhanced by the use of in-shoe orthoses. Orthoses increased the calc eversion angle at heel strike and midstance, the calc range of motion at midstance and reduced the tibial internal tibial rotation angle and the range of motion at heel strike and midstance.

Custom orthoses performed better than standard preformed devices reaching a statistical significance in internal tibial rotation at midstance only. The small values observed overall may indicate a clinical effect to small to have been measureable with
conventional clinical chairside techniques of examination and 2-D motion analysis methods. Comfort is generally accepted as related to fit, and fit may be linked to a better outcome in injured subjects. Anecdotal reports from the subjects in this study expressed they remained pain free in their EVA devices after six months of use. Further research is warranted in orthotic material science, patient reported preferences in material selection and system noise if attained values are clinically small as calculated in this study.
Chapter 8 TIBIO-CALCANEAL VARIABILITY; A 3-DIMENSIONAL KINEMATIC STUDY OF RUNNERS WEARING IN-SHOE ORTHOSES

INTRODUCTION
Movement is governed by complex neural processes controlling muscle activations and relaxations across the joints of the skeletal system. Podiatric biomechanics has classified the foot based on morphology (Clement et al., 1981) structural alignment (McPoil and Cornwall 1996) and flexibility (McMillan and Payne 2008). The pilot study (Chapter three) investigated dynamic motion of the rearfoot in the frontal plane using a 2-D analysis. The foot was modelled as a single rigid segment, calcaneal (Calc) inversion /eversion representing the motion of the tarsus distal to the rearfoot complex. Within the clinical environment this method has been adopted perhaps due to the ease of use and the origins of theoretical paradigm suggesting the foot follows a sequence of events beginning with heel strike and ending with heel lift, a procedure that occurs uninterrupted and on time (Root et al., 1977). More recently dynamic studies have investigated the function of specific joints within the midfoot (Carson 2001) and forefoot (Dannanberg 1999) a progression from traditional single segment “triangle at the bottom of the leg” to multi-segmental models of the tarsus (Nester 2009) discussed in chapters two and three.

Movement science has described the body as a multi-segmental system, with analogies of a chain link type structure were each segment moves relative to each other through coupling processes and preferred postural pathways (Nasher and McCollum 1985). Postural pathways have been considered as reducing agents for large deviations of movement producing a tighter motor effect. The concept of large deviations being reduced or dampened has been accepted worldwide (Nasher and McCollum 1985). The ankle strategy (McCollum and Mcleen 1989) proposed that musculature acting across the ankle joint aides balance were postural sway amplitudes are less than 20.0° and a hip strategy reducing amplitudes above 20.0° (Nasher and McCollum 1985). Traditionally deviation and variability has been inferred as dysfunction or erroneous (Schmidt and Lee 1998). Non-linear dynamic and chaos in motion theories have provided new insight and direction in lower limb function (Glass and McKay 1988; Thompson and Stewart 2002).
Variability has been defined in statistical terms as,

“The variance of data dispersed about the mean quantified by the size of the standard deviation (Riley and Turvey 2002)”.  

It has been linked with noise associated with the instrument, study design and setup. Noise has been defined as,

“Random fluctuations that compromise the deterministic relation between input and output at different levels of analysis in the sensorimotor system (Newell and Corcos 1993; Slifkin and Newell 1999ab)”. 

To quantify the amount of error associated with a 3-D dynamic motion measure an investigation was conducted using a validated rod established and accuracy of 0.5 ± 0.1 mm (chapter four). A further investigation using subjects determined a repeatability intra-class correlation coefficient of moderate to high (r= 0.786) chapter six. 

Motion theorists suggest that variability is inherent and has a functional role in protection, providing adaptability and flexibility during dynamic motion (Slifkin and Newel 1999b; van Emmerick and van Wegen 2000; Latash et al., 2002; Riley and Turvey et al. 2002; Davids et al., 2003). Hamill et al., (1999) suggested coupling variability can be related to overuse injury. They proposed low variability was indicative of injury and high variability a healthy state, in their investigation of patellofemoral pain sufferers. Furthermore restriction of variability within a segment may predispose increased tissue stresses, cartilage damage and eventual degenerative change. Heiderscheit et al. (2002) repeated the study by Hamill et al., (1999) using a modified vector coding technique (Sparrow et al., 1987). They concluded patellofemoral pain sufferers have less variability in the affected limb than in the non-affected limb. Interestingly when the affected knee was strapped variability increased, giving further support to the findings of Hamill et al., (1999) and the injured subject adjusting or restricting (freezing degrees of freedom) the mechanics of movement to prevent or reduce pain hence reducing variability.

Using an in-shoe orthotic has been regarded as introducing an environmental change that can affect overall segment dynamic coupling processes (Kamm et al., 1990). Excessive STJ pronation may cause compensatory motions by increasing the variability
of the knee, hip, pelvis and trunk (Nicolopoulos et al., 2000; Johanson et al., 1994). Esalami et al., (2007) studied 14 healthy males in a static standing posture. A 4.6° wooden wedge was placed under the rearfoot and forefoot. Frontal plane angular variability was six times greater than barefoot variability trials for a medially placed wedge. Hip and knee transverse plane variability was also 2-3 times greater than barefoot trials. They concluded wedged positions may alter the STJ angular variability and its proximal joints and segments in their respective planes of movement. This disruption in the respective coupling pattern may provide an avenue for the higher centres of the brain and musculoskeletal system to find a potentially better one (Todorov 2002).

The research is not clear, sufficient but not excessive varied joint motion may aid in the redistribution of stresses across the foot (Hamill et al., 1999). Too much variability may indicate joint laxity, hypermobility and inadequate control, whereas a reduction in variability may be detrimental, reducing the ability to adjust to uneven surfaces or reduce ground reaction force shock transients. Podiatric biomechanists assessing the foot make clinical decisions based on subjective findings of palpation and manipulation of the foot determining joints as mobile, semi-mobile or rigid. The identification of foot hypermobility may be treated with in-shoe devices to reduce excessive movement. Though not clear is how much variability is normal and acceptable. Underpinning the body of foot mechanical evidence over the last four decades is dependence on characteristics that contribute to successful motor performance or predispose injury (Bartlett 1999). Adoption of this theoretical analogy of an ideal foot type or single dynamic gait cycle that is the ‘optimal,’ may have obscured research progress in this area. For example if an athlete with a proven field or track record presents in the clinical setting their foot profile may be considered the most advantageous and subsequently extrapolated to others as an ideal (Brisson and Alain 1996). A problem with this is that there is no allowance for individual uniqueness within the adopted clinical model of the ideal osseous and structural alignment of the neutral foot model. Deviation in RCSP or in dynamic gait has been considered a clinical or erroneous deviation and not as a unique pathway derived from multiple sensory inputs which may decrease the need for sustained muscle activation through passive dampening. The choice of walking and running speed is selected by each individual and may be a process
of adaption to surface and obstacles in their path, enhancing stability and minimising energy consumption per unit distance (Margaria 1976). Jordan et al., (2006) provided an analogy of the pendulum in swing while Bellchamber and Bogert (2000) describe a vehicle drive shaft. Both tolerate stresses as does the lower limb during walking and running. In acceleration and slowing the potential for muscular strain exists. The plantarflexors and dorsiflexors of the foot are stressed while increasing the speed of locomotion and may require additional force to produce dampening (Jordan et al., 2006). As subjects walk at speeds above their preferred walking speed it becomes necessary to control more actively movement output while at slower speeds than their preferred walking speed the excursions of the centre of mass increases (Orendurff et al., 2004). Damage to the articular cartilage and the underlying bone structures above some threshold is required (Gill and O Connor 2003). Below this threshold value the body’s natural repair mechanisms may help prevent progressive damage to articular cartilage (Radin and Parker 1973). Orthotics may alter this threshold, in addition to exerting structural and angular change.

Variability exists between subjects, between steps, between walking, running, over time and over ground (Steinwender et al., 2000; White et al., 1999) Motion analysis studies are designed to control intrinsic and extrinsic error. Controlling the environment, the subject and the intervention error may still remain. Recently this error or variability has provided a new insight into gait in healthy and injured subjects. Uninjured subjects demonstrate successive movements are similar but not identical with each repetition (Turvey et al., 1990). This unique study undertaken in a clinical environment investigated the error which is synonymous with the standard deviation and termed variability in a group of runners who had patellofemoral pain that was treated with in-shoe orthoses six months prior to participation, and who are now pain free and continue to wear their orthotic devices. The rationale for this study was measurement of joint angles and range of motion between specific points on time series curves to provide objective data the clinician can use to justify an intervention with the aim of restoring or improving lower limb function and understanding in the injured patient.

This study chapter will examine the precise contribution of in-shoe orthoses on 3-D rearfoot and tibial motion variability. The specific aims were,
1. To describe the effect of in-shoe orthotics (specifically EVA and a standard generic polypropylene orthotic) on the 3-dimensional kinematics variability of the rearfoot and tibia during walking.

2. To determine the extent of effect between a custom made (high density EVA) orthotic and a standard generic (semi rigid polypropylene orthotic) on the heel strike and midstance phase variability angles during walking.

8.1 Hypothesis # 5
In-shoe orthoses will change the mean frontal plane (Calcaneal to ground) and transverse plane (tibial internal/external rotation) motion variability at the heel strike and midstance phase of walking.

8.2 Methods
The methodology for this study has been described previously (Chapter 7).

8.2.1 Procedure
The procedures for this study are identical to those used (Chapter 7).

8.2.2 Definitions
All definitions are identical to those used (Chapter 7).

Variability is synonymous with variance of data about the mean and is quantified in this investigation as the standard deviation (SD). The unit of measure is degrees (°)

8.2.3 Kinematics
Angular kinematic definitions are identical to those used (Chapter 7).

8.3 Variability calculation processes for statistical analysis
8.3.1 Within-subject angular variability
Ten trials were recorded for each subject in each experimental condition. Within-subject variability SD data was generated as an ensemble mean SD trial of ten trials in each experimental condition.

8.3.2 Within-subject range of variability
Ten trials were recorded for each subject in each experimental condition. Within-subject range of variability was generated from an ensemble SD trial of ten trials in each experimental condition. The minimum ensemble trial value was subtracted from the
maximum ensemble trial value within each experimental condition to determine the range of variability.

### 8.3.3 Between-subject angular variability

Ten trials were recorded for each subject in each experimental condition. A single ensemble SD value was generated across ten trials for each experimental group in each experimental condition.

### 8.3.4 Between-subject range of variability

Ten trials were recorded for each subject in each experimental condition. Between-subject group range of variability was generated from a group ensemble trial of ten trials. The minimum ensemble trial value was subtracted from the maximum ensemble trial value within each experimental condition to determine range of variability.

### 8.4 RESULTS

#### 8.4.1 Calc angle variability at heel strike.

Between group mean calc angle variability at heel strike was significantly different than barefoot trials ($p<0.001$) (Graph 8.1).

![Graph 8-1 Shod, custom and standard orthoses mean calc angle variability was significantly different than barefoot trials at heel strike ($p<0.001$).](image)

Calc angle variability at heel strike resulted in a mean angle variability change of 11° in shod and custom orthotics. Standard orthotic trials were 12° when compared to barefoot mean 0.9° (Table 8.1).
There were no significant differences at heel strike between shod and custom (p=0.496) custom and standard (p=0.698) and shod and standard devices (p=0.444).

**Table 8-1 Between group mean calc angle variability at heel strike with 95% CI (n=31)**

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Mean(°)</strong></td>
<td>0.9 ± 0.8</td>
<td>12.2 ± 14.5</td>
<td>12.7 ± 13.3</td>
<td>13.1 ± 13.3</td>
</tr>
</tbody>
</table>

The hypotheses accepted were,

**Hₐ** Shod walking changes the mean calc angle variability at heel strike when compared to barefoot walking calc angle variability at heel strike.

**Hₐ** Custom orthotics change the mean calc angle variability at heel strike when compared to barefoot walking calc angle variability at heel strike.

**Hₐ** Standard orthotics change the mean calc angle variability at heel strike when compared to barefoot walking calc angle variability at heel strike.

All rejected hypotheses are detailed in appendix L.
8.4.2 Calc angle variability at midstance.

Between group calc angle variability at midstance was significantly different than barefoot trials (p<0.001) (Graph 8.2).

**Table 8-2 Between group mean calc angle variability at midstance with 95% CI**

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Mean (°)</strong></td>
<td>0.8 ± 0.6</td>
<td>8.7 ± 14.2</td>
<td>10.1 ± 13.6</td>
<td>10 ±14.0</td>
</tr>
</tbody>
</table>

The only statistically significant difference between orthoses was in the shod versus custom orthotic group (p=0.024) (Graph 8.3). Shod and standard orthoses were p=0.094 and custom versus standard orthoses p=0.859.
Graph 8-3  Custom orthoses mean calc angle variability was significantly different than shod only trials at midstance (p=0.024).

The hypotheses accepted were,

**Hₐ** Shod walking changes the mean calc angle variability at midstance when compared to barefoot walking mean calc angle variability at midstance.

**Hₐ** Custom orthoses change the mean calc angle variability at midstance when compared to barefoot walking mean calc angle variability at midstance.

**Hₐ** Standard orthoses change the mean calc angle variability at midstance when compared to barefoot walking mean calc angle variability at midstance.

**Hₐ** Custom orthoses change the mean calc angle variability at midstance when compared to the shod walking mean calc angle variability at midstance.

All rejected hypotheses are detailed in appendix L.

**8.4.3 Tibial angle variability at heel strike**

Between group tibial angle variability at heel strike was significantly different than barefoot trials (p<0.001) (Graph 8.4).
Graph 8-4  Shod, custom and standard orthoses mean tibial angle variability was significantly different than barefoot trials at heel strike (p<0.001).

Calc angle variability at heel strike in shod, custom and standard orthoses resulted in a significant mean difference of 0.8° in shod and 1.0° in custom orthotics over 0.7° in barefoot trials (Table 8.3).

Table 8-3 Between group mean tibial angle variability at heel strike with 95% CI

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (°)</td>
<td>0.7 ± 0.4</td>
<td>1.5 ± 1.1</td>
<td>1.7 ± 1.3</td>
<td>1.6 ± 1.1</td>
</tr>
</tbody>
</table>

Similar to a significant calc variability angle at midstance, custom versus shod trials were significantly different (p=0.029) (Graph 8.5).
Graph 8-5 Custom versus shod trials were significantly different (p=0.029). There were no significant differences between shod and standard orthotic trials, (p=0.382) or custom and standard orthoses (p=0.111).

The hypotheses accepted were,

\( H_a \) shod walking changes the mean tibial angle variability at heel strike when compared to barefoot walking tibial variability angle at heel strike.

\( H_a \) custom orthoses change the mean tibial angle variability at heel strike when compared to barefoot walking tibial variability angle at heel strike.

\( H_a \) standard orthoses change the mean tibial angle variability at heel strike when compared to barefoot walking tibial variability angle at heel strike.

\( H_a \) custom orthoses change the mean tibial angle variability at heel strike when compared to shod walking tibial variability angle at heel strike.

All rejected hypotheses are detailed in appendix L.
8.4.4 Tibial angle variability at midstance.

Between group mean tibial angle variability at midstance was significantly different than barefoot trials (P<0.001) (Graph 8.6).

Graph 8-6  Shod, custom and standard orthoses mean tibial angle variability was significantly different than barefoot trials at midstance (p<0.001).

Tibial angle variability at midstance in shod, standard and custom orthotic trials resulted in a significant mean difference of 1.2° in shod trials and 1.3° in custom and standard orthotic devices over 0.7° in barefoot only trials (Table 8.4).

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (°)</td>
<td>0.7 ± 0.5</td>
<td>1.9 ± 1.4</td>
<td>2 ± 1.1</td>
<td>2 ± 1.1</td>
</tr>
</tbody>
</table>

There were no significant mean differences at midstance between shod and custom trials (p=0.178), shod and standard trials (p=0.123) and custom and standard orthoses (p=0.636).
The hypotheses accepted were,

**H₁** shod walking changes the mean tibial angle variability at midstance when compared to barefoot walking tibial angle variability at midstance.

**H₂** custom orthoses change the mean tibial angle variability at midstance when compared to barefoot walking tibial angle variability at midstance.

**H₃** standard orthoses change the mean tibial angle variability at midstance when compared to barefoot walking tibial angle variability at midstance.

All rejected hypotheses are detailed in appendix L.

### 8.4.5 Calc range of variability at heel strike

Between groups mean calc range of variability at heel strike was significantly different than barefoot trials ($p<0.001$) (Graph 8.7).

![Graph 8-7](image)

**Graph 8-7** Shod, custom and standard orthoses mean calc range of variability was significantly different than barefoot trials ($p<0.001$).

Calc range of variability in shod, custom and standard orthotic trials and resulted in mean differences of 1.1° in shod trials and 1.0° in custom and standard orthotic trials over 0.3° in the barefoot only condition (Table 8.5).
Table 8-5 Between group mean calc range of variability at heel strike with 95% CI

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Mean (°)</strong></td>
<td>0.3 ± 0.6</td>
<td>1.4 ± 3.4</td>
<td>1.3 ± 2.7</td>
<td>1.3 ± 3.1</td>
</tr>
</tbody>
</table>

There were no significant differences at heel strike between shod and custom (p=0.866) shod and standard trial, (p=0.649) and custom and standard orthoses (p=0.803).

The hypotheses accepted were,

H₀ Shod walking changes the mean calc range of variability at heel strike, when compared to barefoot walking calc range of variability at heel strike.

H₀ Custom orthotics change the mean calc range of variability at heel strike when compared to barefoot walking calc range of variability at heel strike.

H₀ Standard orthotics change the mean calc range of variability at heel strike when compared to barefoot walking calc range of variability at heel strike.

All rejected hypotheses are detailed in appendix L.
8.4.6 Calc range of variability at midstance

Between group mean calc range of variability at midstance was significantly than barefoot trials (p<0.001) (Graph 8.8).

Graph 8-8 Shod, custom and standard orthoses mean calc range of variability was significantly different than barefoot trials at midstance (p<0.001).

Calc range of variability in shod, custom and standard orthotic trials resulted in a significant mean difference of 3.1° in shod trials, and 2.8° in custom orthotics. The greatest significant mean difference was in the standard orthotics group 3.2° over barefoot group mean value of 0.2°. There were no significant differences at midstance between shod and custom (p=0.620) shod and standard (p=0.911) or custom and standard orthoses (p=0.471).

The hypotheses accepted were,

**H₀** Shod walking changes the mean calc range of variability at midstance when compared to barefoot mean calc range of variability at midstance.

**H₀** Custom orthotics change the mean calc range of variability at midstance when compared to barefoot walking calc range of variability at midstance.

**H₀** Standard orthotics change the mean calc range of variability at midstance when compared to barefoot walking calc range of variability at midstance.

All rejected hypotheses are detailed in appendix L.
8.4.7 Tibial range of variability at heel strike

Between group mean tibial range of variability at heel strike was significantly different than barefoot trials in shod and custom trials only (p<0.05) (Graph 8.9).

Graph 8-9  Shod and custom orthotic tibial range of variability at heel strike reached significance at p<0.05 level.

Tibial range of variability at heel strike resulted in mean value of 0.1° CI 0.5° in shod trials (p=0.020) graph 8.10, and 0.2° CI 0.5° in custom trials (p=0.003) graph 8.11, over the barefoot mean value 0.2° SD 0.4° (Table 8.6).

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Mean (°)</strong></td>
<td>0.2 ± 0.3</td>
<td>0.3 ± 0.5</td>
<td>0.2 ± 0.5</td>
<td>0.2 ± 0.4</td>
</tr>
</tbody>
</table>
Graph 8-10  Shod orthoses were significantly different than barefoot trials at heel strike (p=0.020).

Graph 8-11  Custom orthoses were significantly different than barefoot only trials at heel strike (p=0.003).

The Hypotheses accepted were,

\( H_0 \) Shod walking changes the mean tibial range of variability at heel strike when compared to barefoot walking mean tibial range of variability at heel strike.

\( H_0 \) Custom orthotics change the mean tibial range of variability at heel strike when compared to barefoot walking mean tibial range of variability at heel strike.

All rejected hypotheses are detailed in appendix L.
8.4.8 Tibial range of variability at midstance

Between group mean tibial range of variability at midstance was significantly different than barefoot trials in all conditions (Graph 8.12).

Graph 8-12  Shod, custom and standard orthoses mean tibial range of variability was significantly different than barefoot trials at midstance (p<0.001).

Tibial range of variability at midstance in shod, custom and standard orthoses resulted in a mean difference of 0.4° in shod trials and 0.3° in custom and standard trials over a mean barefoot value of 0.2° CI 0.3° (Table 8.7).

Table 8-7 Between group mean tibial range of variability at midstance with 95% CI

<table>
<thead>
<tr>
<th></th>
<th>Barefoot</th>
<th>Shod</th>
<th>Custom Orthotic</th>
<th>Standard Orthotic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (°)</td>
<td>0.2 ± 0.3</td>
<td>0.6 ± 0.9</td>
<td>0.5 ± 0.7</td>
<td>0.5 ± 0.7</td>
</tr>
</tbody>
</table>

There were no significant differences between shod and custom trials (p=0.459), shod and standard (p=0.247) or custom and standard orthotic trials (p=0.592).

The hypotheses accepted were,

Hₐ Shod walking changes the mean tibial range of variability at midstance when compared to barefoot mean tibial range of variability at midstance.
Custom orthoses change the mean tibial range of variability at midstance when compared to barefoot mean tibial range of variability at midstance.

Standard orthoses change the mean tibial range of variability at midstance when compared to barefoot mean tibial range of variability at midstance.

Tables 8.8 to 8.11 summarise statistically significant findings. All rejected hypotheses are detailed in appendix L.

**Table 8-8 Calc and tibia angle variability at heel strike summary of significance**

<table>
<thead>
<tr>
<th>Condition</th>
<th>Calc angle variability at heel strike (°)</th>
<th>Tibial angle variability at heel strike (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Barefoot</td>
<td>0.9</td>
<td>0.7</td>
</tr>
<tr>
<td>Shod</td>
<td>12.2*</td>
<td>1.5*</td>
</tr>
<tr>
<td>Custom orthotic</td>
<td>12.7*</td>
<td>1.7**</td>
</tr>
<tr>
<td>Standard orthotic</td>
<td>13.1*</td>
<td>1.6*</td>
</tr>
</tbody>
</table>

*Significant over barefoot at the 0.05 level. **Significant over shod and barefoot conditions at 0.05 level

**Table 8-9 Calc and tibia angle variability at midstance summary of significance**

<table>
<thead>
<tr>
<th>Condition</th>
<th>Calc angle variability at midstance (°)</th>
<th>Tibial angle variability at midstance (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Barefoot</td>
<td>0.8</td>
<td>0.7</td>
</tr>
<tr>
<td>Shod</td>
<td>8.7*</td>
<td>1.9*</td>
</tr>
<tr>
<td>Custom orthotic</td>
<td>10.1**</td>
<td>2.0*</td>
</tr>
<tr>
<td>Standard orthotic</td>
<td>10.0*</td>
<td>2.0*</td>
</tr>
</tbody>
</table>

*Significant over barefoot at the 0.05 level. **Significant over shod and barefoot conditions at 0.05 level

**Table 8-10 Calc and tibia range of variability at midstance summary of significance**

<table>
<thead>
<tr>
<th>Condition</th>
<th>Calc Range of variability at heel strike (°)</th>
<th>Tibial Range of variability at heel strike (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Barefoot</td>
<td>0.3</td>
<td>0.2</td>
</tr>
<tr>
<td>Shod</td>
<td>1.4*</td>
<td>0.3*</td>
</tr>
<tr>
<td>Custom orthotic</td>
<td>1.3*</td>
<td>0.2*</td>
</tr>
<tr>
<td>Standard orthotic</td>
<td>1.3*</td>
<td>0.2</td>
</tr>
</tbody>
</table>

*Significant over barefoot at the 0.05 level.
### Table 8-11 Calc and tibia range of variability at midstance summary of significance

<table>
<thead>
<tr>
<th>Condition</th>
<th>Calc Range of variability at midstance (°)</th>
<th>Tibial Range of variability at midstance (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean</td>
<td>CI</td>
</tr>
<tr>
<td>Barefoot</td>
<td>0.2</td>
<td>0.4</td>
</tr>
<tr>
<td>Shod</td>
<td>3.3*</td>
<td>7.6</td>
</tr>
<tr>
<td>Custom orthotic</td>
<td>3.0*</td>
<td>6.8</td>
</tr>
<tr>
<td>Standard orthotic</td>
<td>3.4*</td>
<td>7.6</td>
</tr>
</tbody>
</table>

*Significant over barefoot at the 0.05 level

### 8.5 DISCUSSION

#### 8.5.1 Calc angle variability at heel strike

Frontal plane measurement of the rearfoot may be the most widely used method of assessment of the foot and ankle in clinical practice and the gait laboratory. The findings of this investigation demonstrate mean barefoot calc angle variability at heel was 0.9°. This value was considered small and may not be clinically measurable using traditional clinical measuring techniques such as observation and goniometry. Though small, this value was an objective measure that quantified a mean baseline value of multiple barefoot walking trial data. In footwear trials the mean calc angle variability at heel strike increased to 12.2°, a greater than ten-fold variability increase (Table 8.8). The introduction of in-shoe orthoses further increased calc angle variability over shod trials but the increase was less than 1.0°. Custom and standard generic orthotic devices were not clinically or statistically different (Table 8.8).

#### 8.5.2 Calc angle variability at midstance

At midstance the mean barefoot calc angle variability was 0.8°, a difference of 0.1° less than that at heel strike. Donning footwear increased the midstance mean shod calc angle variability to 8.7° once again a greater than ten-fold variability increase over barefoot trials (Table 8.9). Worthy of note, the calc angle variability reduced in shod and orthotic trials at midstance by 3.5°. This demonstrated mean calc angle variability was greatest at heel strike phase of gait and reduced as the gait cycle progressed. These values were considered clinically and statistically significant. Surprisingly the mean calc angle variability at midstance in custom devices was statistically significant over shod trials with standard orthotics failing to reach statistical significance at the 0.05 level with a 0.1° lesser value than the custom device. Clinically this small difference was considered insignificant and most probably instrumental (Table 8.9).
8.5.3 Calc range of variability at heel strike

The mean calc range of variability at heel strike in barefoot trials was 0.3°. This was considered clinically small but served as an objective mean baseline measure over multiple data trials. Similar to the calc angle variability at heel strike and midstance, the calc range of variability was increased four-fold when subject wore footwear. The mean variability range increased to 1.4° over barefoot and 1.3° in orthotic trials. There was no clinical or statistically significant difference between orthotic types (Table 8.10).

8.5.4 Calc range of variability at midstance

The mean calc range of variability at midstance in barefoot trials was 0.2° (Table 8.11). This value was considered a clinically stable and objective baseline measure and reflected similar values attained when compared to the baseline values generated for calc angle variability at heel strike and midstance (Table 8.8 and 8.9). When subjects donned footwear, similar to calc angle variability at heel strike and midstance, and the calc range of variability at heel strike, the midstance calc range of variability increased fifteen-fold over barefoot trials to 3.3° a clinically and statistically significant finding, (Table 8.12).

8.5.5 Tibial angle variability at heel strike

The mean barefoot tibial angle variability at heel strike was 0.7°. When subjects donned footwear the mean tibial angle variability doubled in value to 1.5° an increase of 0.8°. This value was statistically significant but was not considered clinically significant (Table 8.8). When the custom orthotic was inserted into the subject’s footwear the tibial angle variability increased to 1.7° a statistically significant result over barefoot and shod only trials. Overall the increase in tibial angle variability was 1.0° not considered clinically significant with a mean difference of 0.1° over the standard orthotic (Table 8.8). The small tibial angle variability increase observed was not surprising a possible explanation may be the angle of the tibia at the heel strike event, and the direction of the ground reaction force may subject the lower limb to more compressive forces as opposed to inducing transverse motion. Greater variability may have been attained if the study was conducted whilst subjects ran on the treadmill.
8.5.6 Tibial angle variability at midstance
The mean barefoot tibial angle variability at midstance did not change from the baseline barefoot value of 0.7° recorded at heel strike. In shod trials the tibial angle variability increased to 1.9° at midstance, an increase in variability of 1.2° as gait progressed from heel strike to midstance (Table 8.9). The increase in variability of 1.2° reached statistical significance but was a small clinical measure in terms of the tibial transverse motion available in a chair-side clinical examination. The introduction an orthotic increased the tibial angle variability by 0.1° irrespective of type, custom made or standard generic. A comparison of the mean variability values recorded at heel strike in each condition to those at midstance demonstrated an overall increased mean variability of the tibial angle (Table 8.8 and 8.9).

8.5.7 Tibial range of variability at heel strike
The mean barefoot tibial range of variability at heel strike was 0.2°. The increase in tibial variability at heel strike in shod trials was 0.1° over barefoot generating a variability increase of 0.1° with no change in variability in either orthotic trial (Table 8.10). These values while achieving statistical significance they were not considered clinically useful or meaningful. A possible explanation was instrumental noise.

8.5.8 Tibial range of variability at midstance
The mean barefoot tibial range of variability at midstance was 0.2°, no difference over the barefoot range of variability at heel strike. Variability increased three fold to 0.6° in shod trials, reducing variability by 0.1° when orthotics were introduced to the trial. Shod tibial range of variability and orthotic induced tibial variability trials at midstance were statistically significant but clinically questionable as the generated values were 0.5° respectively (Table 8.11).

The findings of this study demonstrate frontal plane calcaneal inversion / eversion angle and tibial angle variability is increased in shod and orthotic trials over barefoot walking at heel strike and midstance (Table 8.8 and 8.9). Statistical significance was reached at the 0.05 level. The calc mean range of variability at heel strike and midstance was similarly increased in shod and orthotic trials over barefoot trials. Tibial range of variability at heel strike was not considered clinically significant but did reach statistical significance in shod and custom trials over barefoot variability (Table 8.10).
through the gait cycle the range of tibial variability at midstance was statistically significant at the 0.05 level in shod and orthotic devices over barefoot trials (Table 8.11).

Traditionally clinical estimates of pronation have evolved from observational analysis of static stance positions of the rearfoot, morphology classification, arch indices, navicular drift, and goniometry which have been reported as having questionable reliability (McPoil and Cornwall 1996; Elveru et al., 1988a; Menz 1995). More recent the advent of high speed 3-D motion measurement has become available. This advance in technology has provided an alternative direction of research within clinical practice. Podiatrists have adopted a clinically useable model of static weight-bearing, semi-weight bearing and non weight-bearing assessment of foot kinematics. The findings of this study provided a wholly new insight into in-shoe orthoses in dynamic motion at heel strike within the clinical setting which is a progressive direction for the podiatric evidence base. Work previous to this has involved purpose built gait laboratories and institutions for higher education. As long as this situation remains mainstream clinical practice will be slow to adopt alternatives to treatment intervention and rejection of theories which may be now in the passing of time be redundant but continue to be used because of a lack of validated alternatives (Nester 2009). The angular data and variability obtained from this investigation raise questions about previous assumptions of the effects of in-shoe devices, it appears in part the angular change we observe and have observed for over three decades now appears to be a smaller factor in a combination of multiple events involving neural, muscular, ligamentous as well as skeletal alignment.

In this study shod calc angle variability at heel strike was increased both orthotic types over barefoot only trials. Despite the orthotics the largest effect was observed between barefoot and shod trials. This was a surprising finding but is in agreement with anecdotal evidence that the shoe is the most important orthotic. A clinical adopted theory of the foot function assumes the calc strikes the ground slightly inverted absorbing shock as it everts, and the foot pronating as the knee flexes (Root et al., 1977). Variability was increased at heel strike despite the addition of medial rearfoot wedging. Clinically one may have expected the wedge would have prevented the calc rolling medially and one may have recorded a less variable calc strike. Nigg et al.,
(1999) speculated the shoe is the first filter dampening ground reaction forces and providing supportive mechanism. The tightness of the shoe fit encompassing the dorsal, medial and lateral sides of the foot may assist and provide sensory feedback as hair follicles and a greater skin surface area is factored into proposed neuro-muscular control mechanisms. The fibro-fatty tissue of the plantar heel area may act as a second filter complementing the shoe adding yet more proprioceptive output and finally the third filter mechanism, the in-shoe orthotic may provide a link between a good fit and a foot hugging mould that brings the medial longitudinal arch into the equation when typically it may not be in contact with the shoe or inlay for example in a cavoid foot. A combination of these processes may result in a central nervous system motor function re-programming (Nigg et al., 1999). Any observed variability may not be system error or noise as has been previously thought (Newell and Corcos 1993). It could not be dismissed, the data in this study show the donning of footwear increased variability at all phases. A factor that supports previous reports the shoe may be most important orthotic (Nigg et al., 1999). If this were to be accepted, questions are raised concerning the injury processes that may occur within footwear, and the successful treatment outcomes following the introduction of an orthotic to the shoe.

Current thinking on footwear effects on runners has found, footwear alters the foot strike landing pattern in habitually shod runners over habitually barefoot runners (Lieberman et al., 2010). The findings of five subject groups from Kenya and America running across force plates and overground demonstrated habitual barefoot runners avoid heel strike landing most frequently between forefoot and midfoot strike patterns. In contrast habitually shod runners mostly heel strike, facilitated by the elevated and cushioned heel of the modern running shoe. These findings demonstrate barefoot runners generate smaller collision forces than shod rearfoot strikers which may protect the foot and lower limb from impact injury now experienced by a high percentage of runners (Lieberman et al., 2010). With respect to the clinical translation of these finding and the findings of this thesis, the observed ten-fold increases in variability of the calcaneus and tibia during heel strike and midstance periods of gait may when footwear was donned may be a compensatory protective reaction to the feedback processes generated while walking and running. Furthermore the range variability increased further when the orthotic device was introduced to the shoe, perhaps
enhancing the necessary variability compensatory process further in a complimentary fine tuning process.

Continuing through the stance phase to mid stance the mean calc angle in shod trials continued to reach significance over barefoot trials. However the custom orthotic increased calc variability further over the shod only condition which is in agreement with past research (Stacoff et al., 2000a; Stackhouse et al., 2004). The standard preformed device did not reach significance over shod or custom orthotics, table 8.9. Tibial angle variability is synonymous with internal/external tibial rotation. At heel strike tibial internal/external angle variability was increased over barefoot trials in shod, custom and standard orthotic trials. Anecdotally and clinical experience would have been to expect a custom prescribed orthotic to provided the greatest angular control and restrict excessive angular motion variability. The finding of increased variability supports theories of tissue stress models, bio-mechanisms and inherent neurological mechanoreceptor processes responding to the foot orthoses interaction (Smith et al., 2000). Recent research suggests variability is a natural inherent protective mechanism in an individual. An increase in variability may indicate inadequate control of a joint or a reduction in variability may be a substantial reduction in the ability to adjust to perturbation and alleviate impact shocks (Heiderscheit 2000; Holt et al., 1995; Hamill et al., 1999). The question of flexible foot type’s hypermobility and rigidity was accounted for in the design with subject’s foot posture index mean scores of 8.0. The muscles of the lower limb and foot act to restrict contain or dampen motion. The stresses and strains of activity are generally accepted as the factors in soft tissue injury. McPoil and Hunt (1995) proposed the tissue stress model in management of lower limb injury. If tissue stress causes injury through repetitive strain or end range of motion, then alleviation of tissue stress must be part of the management process which may include in-shoe orthoses or taping. Most common research reported strains are long tendinous muscles (Hamstring) and plantar fascia injuries through stresses generated in dynamic motion or in weight-bearing for long periods (Hicks 1953). Isometric motions have been linked and to determine the role in support, guidance and affecting a motion they have been surgically released resulting in greater range either guiding or restricting motions (Hintermann et al., 1995; Rassmussen 1985). A possible explanation for the increased variability over barefoot trials may be explained by the subjects chosen for the study.
The subjects were competitive runners with a past history of lower limb overuse injury six months prior to being prescribed in-shoe orthotics. Now six months post initiation they remain asymptomatic. This study supports the use of orthotics and when linked with the increased calc variability at heel strike is in agreement with the most recent theory proposals of variability redistributing the heel strike transients in a form of inherent injury prevention, whether articular or soft tissue. A question remains a subject may limit motion movement in a segment when injured observed by a low segment variability, then following the introduction of an intervention i.e. an in-shoe orthotic the variability increases. If this is accepted as beneficial at what stage should treatment begin after injury? Despite questions concerning specific therapeutic effects. A clinical question that remains is the use of materials in the manufacture and prescription of in-shoe devices. Often patients are given the same device despite having different symptoms or differing body weights. A finding of this study and proposal for future work is investigation into softer orthotic types. Accommodative or often labelled quasi-functional materials EVA is widely used in orthoses and footwear manufacture and was used in this study. A reason for their functionality being suspect may be lower density materials may deform and may provide less control of STJ and MTJ pronation. The findings of this investigation conclude this is not the case. Subjects had been using their respective devices for six months and continue to date. Obvious is compression of the device over time however the findings suggest this may be a significant beneficial factor in the success of their use given the variability of the calc angle and tibia angle at heel strike, which were significant over shod only trials and the preformed polypropylene orthotics were not. This finding suggests the variability may be a factor in the compressed or flexibility of the devices enhancing variability rather than restricting it.

8.5.9 Limitations
There are several limitations with this investigation. Care was taken to refine the methodology and data collection techniques to minimise instrument error by pilot work and successive trial captures. However the concentration and uniqueness of this thesis was that all investigations were conducted in a busy clinical practice. The emphasis was on adopting a 3-D motion system into patient assessment clinics. For this to be possible the patient marker and cluster setups were simple wand and non collinear designs.
Furthermore the retroflective markers were large to facilitate the automated tracking processes. Consequently some minimal Instrument error may be evident in data however this was necessary to reduce the overall time involved both by the patient and the clinician. This thesis moved the bias of 3-D motion analysis away from a laboratory based tool to the front line of patient assessment and treatment within the clinical setting. Finally, space was limited necessitating the use of a treadmill subsequently no over-ground analysis was performed.

8.6 CONCLUSION
Purpose of this study was to investigate the role of in-shoe orthoses on rearfoot and tibial variability during walking. The findings of this study support in-shoe orthoses increase variability in calc and tibial angles at heel strike and midstance. Similar increases are induced in calc range of variability at heel strike and midstance. Though clinically small measurements were observed they support current theory on variability in symptomatic subjects being reduced and increased in healthy individuals. This may be a limitation of previous clinical examination techniques and a limitation of 2-D kinematic techniques that provide limited data. Despite the financial constraints of high speed 3-D equipment this study endorses the adoption of 3-D motion analysis into the clinical assessment, not as a routine but as specialist service. Finally this research suggests preformed orthoses are effective in lower limb treatment when compared to custom orthotic devices. The finding support custom devices increasing calc variability during midstance and tibial variability at heel strike over barefoot walking, shod and standard orthotics. But fundamentally the greatest contributing factor in variability was the adoption of footwear.
Chapter 9 OUTLOOK FOR THE FUTURE

INTRODUCTION
The translation of this thesis into clinical practice as with the interpretation of current literature based on assessment of patients using 3-D motion analysis raises several questions. In terms of clinical significance within the clinical setting the findings of increased variability at heel strike and midstance to the clinician elude conventional chairside measuring techniques involving goniometry and visual observation and therefore may be seen as laboratory based research. As an outcome of orthotic treatment, how much change an orthotic may cause may not be apparent in the clinic because of the dynamic motion of the subject and the need for specialised motion analyses equipment. Increased variability in this sense must be accepted on the basis of clinical significance versus statistical significance. Hypothesis testing and p-values provide a level at which an investigator can reject the null hypothesis, but equally does not mean that it must be rejected as only in mathematics can things be disproved and in true effects in real life never equal zero (Hopkins 2002). Accepted in this thesis the changes in variability that were statistically significant were presented along with statistically non significant findings and their graphical representations. The reason for this was to demonstrate a trend that was supported anecdotally from patients used in this study who reported significant clinical improvement and change in their symptoms following the introduction of in-shoe orthoses. In addition patient reported preferences were evident from anecdotal verbal reports in the absence of statistical significance this was especially apparent with custom orthoses in comparison to preformed devices. This may suggest that there are multiple factors or processes affecting the overall functioning of an individual. Muscular strength has been proposed as a key factor in in reduced injury rates in runners through being sufficiently strong to allow a lesser impact transient while running (Lieberman et al., 2010). This conflicts with research in footwear science that focuses on engineering a running shoe that provides an optimal control and with an orthotic device that may be equally aimed at this goal. In hindsight if this paradigm were true orthoses may actually predispose an individual to injury through weakened muscles or tendinous structures. For the clinician who may be presented with a myriad of clinical symptoms and pathologies within the same patient, providing an in-shoe device may alter the rearfoot angle or
variability at heel strike or midstance but at what cost? Is this a trade off? A patient may report a clinically significant improvement in their condition following orthotic treatment but subsequently develop another complaint. A question arises is this an outcome of the initial intervention? A side effect, or a coincidence? The clinician must make an informed clinical decision based on objective findings in a differential of multiple factors to determine a treatment plan. The adoption of 3-D motion analysis into the clinical decision making processes may help quantify visual change but does not provide the clinician with all the answers. For example, unilateral plantar facial thickening may be apparent in diagnostic ultrasound with respect to the affected foot but what about opposite foot. Is this an outcome of a pathological process or as a result of wearing a shoe or in-shoe orthotic device? Subsequently what are the long term effects of in-shoe devices on the asymptomatic foot given that orthotic devices are provided as a pair of devices in order to maintain symmetry. The paradigm of the vertical calcaneus clinically is not apparent in the attending population. A patient may present with bilateral flat foot (Pes Planus) and marked rearfoot valgus yet have no symptom or history of injury. Taken in perspective 3-D motion analysis provides the clinician with new insight and direction into the functioning of the foot. Gait analysis in the clinical setting provides an objective measure of disease as it impacts on dynamic motion. It assists the clinical intervention pre-injury and post injury in rehabilitation and helps inform clinical decision making processes. While 2-D analysis has been used perhaps because of the ease of setup procedure, it falls short of 3-D techniques. It may be important information has been overlooked and individualism of the patient overlooked. Adaptability may be synonymous with variability clinically the techniques of manipulation of joint of the foot as with chiropractic and physiotherapy treatment may improve flexibility and range of motion within joint thereby improving the adaptability and variability measured of the structure to perturbation with resultant patient reports of clinical improvement. The in-shoe orthoses may also benefit the patient in this way. The stresses on specific joints or structures may be relieved, freeing up alternative methods of compensation, reducing the threshold for pain or injury. With the passing of time 3-D setup techniques are now less time consuming and the analyses faster and over greater numbers of trials. The complexity of analysis, vector coding techniques, cross correlations are dependent on the investigator and their aims
and setting. The use of 3-D may be kept clinically useable by refining the methodology. The podiatrist is uniquely placed to utilise clinical judgment complimented by this technology to move forward the evidence base in the role of lower-limb function and emerging theory. Future studies may investigate the effect of footwear on variability in specific foot types i.e. hypermobile feet to establish is variability increased with flexibility and reduced in rigidity, also in the cavoid foot which may exhibit reduced flexibility. The effect of pace may be a contributing factor in variability reducing as speed increases, and increasing as we slow. Conversely it may be the opposite in older people, slowing walking dynamics may be implicated in fall prevention as variability may be reduced. This has subtle undertones there may be a higher inherent controlling process.

9.1 Conclusions and Important contributions

Chapter 3.0 established the contribution a simple machined wedge adhered to the outside of a running shoe on the 2-D rearfoot angle. Despite a significant reduction in STJ pronation at midstance it became apparent the effect of the wedge was insignificant at heel strike. Further, it highlighted several important limitations of current 2-D kinematic analysis methods based on rearfoot angle measurement within the clinical setting.

Chapter 4.0 established the validity of a high speed 3-D motion system and custom made calibration frame within the clinical settings. Justification for the investigation and equipment cost over 2-D methods were accepted as measurement outcome has a direct clinical impact on the patient. It was concluded the 3-D motion system and custom made calibration frame are accurate and clinically useable tools for lower limb kinematic analysis, achieving dynamic length measurement values of 0.5 mm.

Chapter 6.0 addressed the repeatability of 3-D motion analysis in healthy subjects, treadmill walking over a three week period. The findings supersede the single point discrete data analysis method employed in Chapter three. The Repeatability of the investigation was moderate to excellent ICC of \( r=0.786 \). The findings also provided a baseline reference of gait variability in healthy subjects for further investigations.

Chapter 7.0 introduced a custom carved CAD/CAM scanning and orthotic manufacturing process. Subjects sampled had been injured but remained
asymptomatic for six months following treatment with CAD/CAM manufactured in-shoe orthoses. During dynamic gait rearfoot and transverse tibial angles and ranges of motion were reduced. Custom orthoses were statistically superior to preformed devices. Clinical reductions in angular kinematics were typically 1-3.0° this finding, while supported anecdotally by subjects were considered clinically small but significant.

**Chapter 8.0** progressing from the findings of chapter seven the results were analysed with respect to subject variability. Traditionally research has focussed on orthotic induced structural and positional kinematic joint changes. Unique to this study tissue stress theories mechanoreceptors and bio-feedback mechanisms have been suggested as inherent processes overlooked by previous assessment methods. Nowadays with advanced instrumentation individual uniqueness and adaptive movement processes in gait patterns are emerging. The results of this thesis support increased variability in subjects following the introduction of an in-shoe orthoses at heel strike and midstance phases of walking. The contribution of in-shoe devices exert a mechanical lever effect on the bone structure evident from chapters three and seven which may transfer into reductions in dynamic angular boney change, tissue stress, proprioceptive or neural feedback mechanisms, registered in the central nervous system as variability.
10.0 Glossary

Abduction  Abduction of the foot (or part) is a motion in the transverse plane around a vertical axis in which the distal aspect of the foot (or part) moves away from the midline of the body

Adduction  Adduction of the foot (or part) is a motion in the transverse plane around a vertical (frontal-sagittal axis) in which the distal aspect of the foot (or part) moves toward the midline of the body

Ambulation  Walking

Angular displacement  The rotational component of a body’s motion

Angular velocity  A rate of change of angular displacement

Anterior  The front of the body or a part facing toward the front, anterior in front or forward part of

Anthropometry  The study of proportions and properties of body segments

Biomechanics  The study of mechanical laws and their application to living organisms, especially the human body and its movement

Cadence  The number of steps taken over a period of time, usually steps per minute

Centre of mass  The midpoint or centre of the mass of a body or object

Centroid  The two-dimensional coordinates of the centre of an area

Co-planar  Lying or acting in the same plane

Coronal plane  Frame of reference for the body – viewed from the front

Direct linear transformation (DLT)  The common mathematical approach to constructing the three-dimensional location of an object from multiple two-dimensional images

Distal  The furthest position away from the point of reference

Dorsiflexion  To flex backward, as in upward bending of the fingers, wrists, foot, or toes
Double support  The stance phase of one limb overlaps the stance phase of the contralateral limb creating a period during which both feet are in contact with the ground

Electrogoniometer  A device for measuring changes in joint angle over time using either a potentiometer or strain gauge wire

EMG or electromyography  The study of the electrical activity of muscles and muscle groups

Eversion  Eversion of the foot (or part) is a motion in the frontal plane around a transverse-sagittal axis (longitudinal axis) in which the plantar surface of the foot (or part) moves away from the midline of the body

Extension  A movement which increases the angle between two connecting bones

Filtering  The process of manipulating the frequencies of a signal through analogue or digital processing

Flexion  A movement which decreases the angle between two connecting bones

Force platform  A device for measuring the forces acting beneath the feet during walking

frontal plane  Frame of reference for the body – viewed from the front

g  Acceleration due to gravity

Gait  The manner of walking

Gait analysis  The study of locomotion of humans and animals

Global frame of reference or GCS  A set of orthogonal axes, one of which is parallel with the field of gravity

Ground reaction force  The reaction force as a result of the body hitting or resting on the ground

Habituation  Becoming accustomed

Inferior  Located in the lower position;
Glossary

Inversion  Inversion of the foot (or part) is a motion in the frontal plane around a transverse-sagittal axis (longitudinal axis) in which the plantar surface of the foot (or part) moves toward the midline of the body

Kinematics  The study of the motion of the body without regard to the forces acting to produce the motion

Kinetic energy  The energy associated with motion, both angular and linear

Kinetics  The studies of the forces that produce, stop, or modify motions of the body

Lateral  Aside away from the midline

Linear displacement  Distance moved in a particular direction

Linear velocity  Speed at which an object is moving in a particular direction

Medial  Middle towards the midline

Midstance  The period from the lift of the contra lateral foot from the ground to a position in which the body is directly over the stance foot

Midswing  This is the period of swing phase immediately following maximum knee flexion to the time when the tibia is in a vertical position

Motion and movement analysis  A technique of recording and studying movement patterns of animals and objects

Non-collinear  Points that do not lie in a straight line (plane)

Plantarfexion  Plantarfexion of the foot is a motion in the sagittal plane around a transverse-frontal axis in which the distal aspect of the foot moves plantarwards, or away from the tibia, and the angle at the ankle become greater

Posterior  The back of the body or a part placed in the back of the body, posterior behind or to the rear of
**Pronation**  Pronation is a motion which has a component on all three body planes, but of these component motions none can take place independently of the other two. The motion of pronation is a complex motion comprised of ABDUCTION, EVERSION and DORSIFLEXION

**Proximal**  Nearest the point of reference

**Range of motion**  The angular through which a limb moves

**Rehabilitation**  Restoring a patient or a body part to normal or near normal after a disease or injury

**Sagittal plane**  Frame of reference for the body – viewed from the side

**Single support**  The period during the gait cycle when one foot is in contact with the ground

**Spatial**  Distance

**Stance phase**  The period when a foot is contact with the ground

**Step length**  Distance between two consecutive heel strikes

**Step time**  Time between two consecutive heel strikes

**Stride length**  Distance between two consecutive heel strikes by the same foot

**Stride time**  Time between two consecutive heel strikes by the same foot

**Superior**  Located in the higher position or top

**Supination**  Supination is a motion which has a component on all three body planes, but of these component motions none can take place independently of the other two. The motion of supination is a complex motion comprised of ADDUCTION, INVERSION, and PLANTARFLEXION

**Swing phase**  period when a foot is not in contact with the ground

**Temporal**  Timing
**Total support**  The total time the body is supported by one leg during one complete gait cycle

**Translation**  Movement in a particular direction

**Transverse**  Frame of reference for the body – viewed from above

**Walk mat and walkway**  A device to measure the temporal and spatial parameters of gait

**x y and z**  Linear velocities in x, y, and z directions
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11.1 Bibliography


