Université de Montréal

Effect of foot angle changes on body joints and segments during standing and running

Effet de changement d'angle au pied sur les articulations et les segments lors de l'équilibre debout et de la course

by

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RÉSUMÉ

Le pied contribue à l'équilibre en station debout et participe à une démarche harmonieuse. Un désalignement du pied et de la jambe pourrait perturber cet équilibre et modifier l'amortissement des chocs et entraîner des charges excessives aux articulations. Les chaussures et les orthèses servent à aligner le pied et la jambe correctement. Malgré le soulagement apparent des symptômes liés aux blessures, près de 40% des coureurs n'en retirent peu ou aucun bénéfice. Les trois études formant le corps de cette thèse portent sur l'effet des modifications de l'alignement du pied et de la jambe sur les articulations et les segments proximaux lors de l'équilibre postural et à la course.

Au moins onze sujets masculins ont participé à ces études. Les mesures de la posture ont été réalisées au moyen d'un système vidéo tridimensionnel comprenant cinq caméras. Une cale en bois a été placée sous les côtés antérieur, postérieur, latéral et médial du pied dominant afin de perturber l'équilibre debout sur une jambe. Lors des expériences à la course, des données vidéo et de plate-forme de forces ont été collectées simultanément.

Le premier objectif de cette recherche était de tester comment une cale qui réoriente le pied modifie l'alignement des articulations du membre inférieur, du bassin et du tronc en station debout sur une jambe. Lorsque comparée à la condition sans cale, la variabilité angulaire dans le plan frontal pour l'articulation talo-calcanéenne était environ 6 fois plus importante qu'avec une cale médiale. Pour les cales antérieures et postérieures, la variabilité angulaire de la cheville et de la hanche dans le plan sagittal ainsi que celle du basin et du tronc dans le plan transverse était environ 2 à 3 fois plus élevée.

La seconde étude porte sur les mouvements à l'avant et à l'arrière pied et leurs effets sur la rotation tibiale lors de la phase de support durant la course. À la réception, le couplage cinématique de l'avant-pied par rapport à l'arrière-pied était déphasé. Au milieu de la phase de support, le couplage devient moins déphasé. Du

milieu de la phase de support à la poussée, la déphase se reproduit. Par comparaison à la course avec pieds nus, la course avec sandale a démontré une relation moins déphasée que celle qui était entre l'adduction/abduction à l'avant pied et l'éversion/inversion à l'arrière pied. Il a été démontré que la rotation tibiale n'était pas modifiée par rapport à celle à l'arrière pied lors du contact du talon par le mouvement à l'avant pied dans le plan horizontal.

Le troisième objectif était de tester si l'éversion à l'arrière pied et la rotation tibiale interne agissent sur le moment d'adduction maximal au genou et sur la force de réaction au sol lors de la phase de support à la course. Les mouvements de l'arrière-pied et du tibia ont été modifiés avec l'utilisation des orthèses lors des essais à la course. Une corrélation positive a été observée entre le moment d'adduction maximal au genou et l'éversion à l'arrière-pied. Les résultats indiquent qu'une modification du mouvement à l'arrière-pied dans le plan frontal pourrait être associée à une réduction du moment en adduction au genou excessif mais non à un amortissement de la force verticale de réaction au sol.

En général, ce travail de recherche souligne l'importance du pied en relation à ses articulations et segments proximaux lors de l'équilibre en station debout et lors de la course. Il est anticipé que ce travail pourrait aider les cliniciens à développer de meilleures orthèses.

ABSTRACT

The foot contributes to standing stability and participates to provide a smooth gait. Changes in foot and leg alignment could modify the stabilizing and shock-absorbing role of the lower-limb and, in turn, cause irregular loading on the body joints. Shoewear and foot orthoses have been advocated to align the foot and leg properly. Despite apparent relief of symptoms from injuries, up to 40% of runners were found to gain little or no benefit through the application of foot interventions. The three studies comprising the core of this thesis are intended to establish the contributions of the foot-angle changes in relation to the angular variability and amplitude of lower-limb joints, during standing and running.

At least eleven able-bodied male subjects participated to these three studies. Posture measurements were performed by means of a three-dimensional video-based system consisting of five cameras. A wooden wedge was placed under the anterior, posterior, lateral and medial sides of the dominant foot to perturb single-limb stance. In the running experiments, video and force-plate data were collected simultaneously.

The first objective of this research was to see how single-limb standing posture is affected by the lower-limb joints, pelvis and trunk, when a wedge re-orients the foot. Compared to the no wedge condition, the frontal plane angle variability for the subtalar joint was about 6 times greater for the medial wedge. For the anterior and posterior wedges, angle variability of the ankle and hip in the sagittal plane and the pelvis and trunk in the transverse plane was about 2 to 3 times higher.

The second study determined the forefoot-rearfoot motion patterns and their effects on tibial rotation during the stance phase of running. The measure of forefoot-rearfoot motion patterns was manipulated by sandal during running. Forefoot-rearfoot coupling was more in out-of-phase at heel-strike. This transitioned into an in-phase relationship by mid-stance. From mid-stance to toe-off, this coupling pattern transitioned back to an out-of-phase relationship. The coupling pattern of forefoot adduction/abduction and rearfoot eversion/inversion, was more in-phase during the

heel-strike phase of shod running than in barefoot running. Nevertheless, there was no statistically significant reduction of tibial internal rotation.

The third specific objective tested the contributions of rearfoot eversion and tibial internal rotation to peak knee adduction moment and ground reaction force during the stance phase of running. Rearfoot and tibial motions were manipulated with the use of foot orthoses during running. A positive correlation was observed between peak knee adduction moment and rearfoot eversion amplitude. Findings imply that modifying rearfoot frontal plane motion with the use of orthoses could be related to a reduction of excessive knee adduction moment but not to a cushioning of the vertical ground reaction force.

In general, this research work underlines the importance of the foot segment in relation to its proximal joints and segments during standing and running. It is anticipated that this work can help clinicians to develop better orthotics.

MOTS CLÉS

- **Appareils orthopédiques**
- **Articulations**
- **Cinématique**
- **Cinétique**
- **Posture**
- **Avant pied**
- **Arrière pied**
- **Moment au genou**
- **Rotation tibiale interne**
- **Force de réaction au sol**

KEY WORDS

- **Orthotic devices**
- **Joints**
- **Kinematics**
- **Kinetics**
- **Posture**
- **Forefoot**
- **Rearfoot**
- **Knee moment**
- **Tibial internal rotation**
- **Ground reaction force**

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LIST OF ABBREVATIONS

AGRF	Active Ground Reaction Force
ANOVA	Analysis of Variance
AW	Anterior Wedge
COP	Centre of Pressure
CRP	Continuous Relative Phase
EV/TIR	Rearfoot Eversion and Tibial Internal Rotation ratio
$\mathbf{FF}_{\mathbf{ad/ab}}$	Forefoot Adduction/Abduction
$\mathbf{FF}_{\mathbf{d/p}}$	Forefoot Dorsi/Plantar flexion
$\mathbf{FF}_{\mathbf{ev/in}}$	Forefoot Eversion/Inversion
KAM	Knee Adduction Moment
LW	Laterally Wedge
ML	Medially Wedge
PW	Posterior Wedge
REV	Rearfoot Eversion
$\mathbf{RF}_{\mathbf{ev/in}}$	Rearfoot Eversion/Inversion
RMS	Root Mean Square
TIR	Tibial Internal Rotation

Praise be to Allah, the Cherisher and Sustainer of the worlds

This Thesis is dedicated

To memory of my Mom and Father who first taught me the Faith

To my wife, Elham, for all her love, support, patience, and encouragement

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Chapter 1

1. INTRODUCTION

The foot is an integral mechanical part of the lower extremity. It contributes to standing stability and participates to provide a smooth gait. Changes in the alignment of foot and leg could modify the stabilizing and shock-absorbing role of the lower-limb and, in turn, cause irregular loading on the more proximal parts of the body (Radin et al., 1991). Shoewear and foot orthoses have been advocated to align the foot and leg properly. These serve to increase stability during standing and reduce overloading in lower-limb joints during gait (Nigg et al., 2003). Foot orthoses, irrespective of designs or postings, were reported to be ineffective at reducing postural instability (Hertel et al., 2001). Furthermore, despite apparent relief of symptoms from injuries, up to 40% of runners were found to gain little or no benefit through the application of orthoses (Gross et al., 1991). Thus, an understanding of the interactions of foot function and upper body joints during standing and running is needed to find better solutions for wedge/orthotic fitting. The three studies comprising the core of this thesis are intended to establish the contributions of the foot-angle changes in relation to the angular variability and amplitude of lower-limb joints, during standing and running.

This chapter describes the anatomy of the foot and ankle related to the purposes of the three studies. Next, the role of the subtalar joint, and its more proximal joints and segments will be described in maintaining posture during standing. Then, abnormal motions of the subtalar joint involving excessive rearfoot and tibial rotations will be addressed to emphasize the importance of these rotations in lower-limb injuries. This will be followed by a description of the effect of forefoot motions on rearfoot and tibia motions during running. The use of foot orthoses in the prevention rearfoot and tibial rotation, and their relation to ground reaction forces and knee moments, will follow. Finally, the overall structure of thesis will be presented.

1.1 Foot and Ankle Joints and Axes of Motion

The foot is comprised of 28 bones and 33 joints. Three major segments make up the foot, namely forefoot, midfoot and rearfoot. The forefoot includes the five metatarsal, fourteen phalanges and two sesamoid bones. The navicular, cuboid and three cuneiform bones make up the midfoot and the rearfoot is comprised of the calcaneus and talus. The ankle includes the tibia, fibula and talus forming the ankle mortise. The joints between the base of five metatarsal, cuboid and three cuneiform is termed the tarsometatarsal joints. The transverse tarsal joint consists of talonavicular joint and calcaneocuboid joint. The joint between the talus and calcaneus is the subtalar joint (Figure 1.1). The ankle complex consists of the tibiotalar, fibulotalar, and tibiofibular joints.

The foot acts as a rigid platform in supporting the weight of the entire body, as standing on single-limb, or quite flexible platform in ground contact, as in running barefoot on the sand. The transition from rigid lever to shock-absorbing platform depends on the orientations of rotations axes in subtalar and transverse tarsal joints.

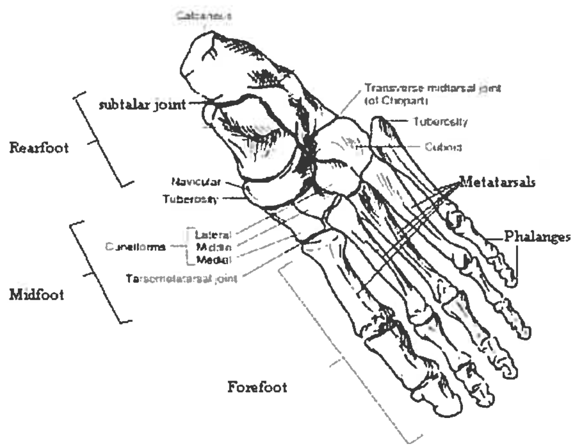

Figure 1.1 A dorsal view of foot joints and bones (adapted from Nordin & Frankel, 2001)

Manter (1941) determined that the subtalar axis of rotation is oriented upward at an angle of 42° from the plantar surface and medially 16° from the midline (Figure 1.2). Inversion and eversion occurs primarily in this axis.

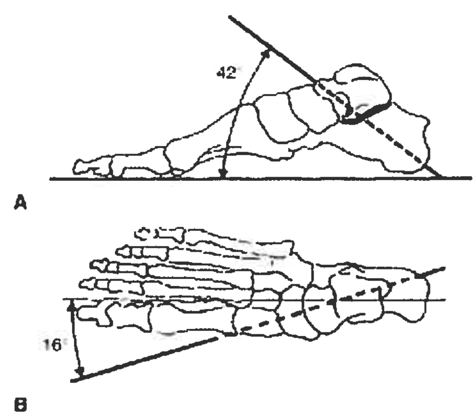

Figure 1.2 Subtalar joint axis. A, Sagittal plane axis rises up at a 42° angle from the plantar surface (lateral view). B, Transverse plane (top view). The axis is oriented 16° medial to the midline of the foot *(Manter 1941)*

In the transverse tarsal joint, a longitudinal axis and an oblique axis of rotation were determined. The longitudinal axis is oriented 15° upward from the horizontal and 9° medially from the longitudinal axis of the foot. Inversion and eversion occur in this axis (Figure 1.3). The oblique axis is oriented 52° upward from the horizontal and 57° anterior-medially (Figure 1.4). Flexion and extension occur primarily about this axis.

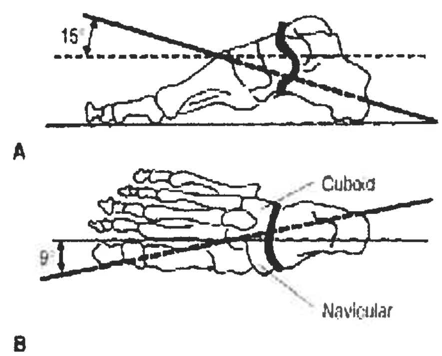

Figure 1.3 Longitudinal axis of the transverse tarsal joint. A, Lateral view. B, Top view *(Manter 1941)*

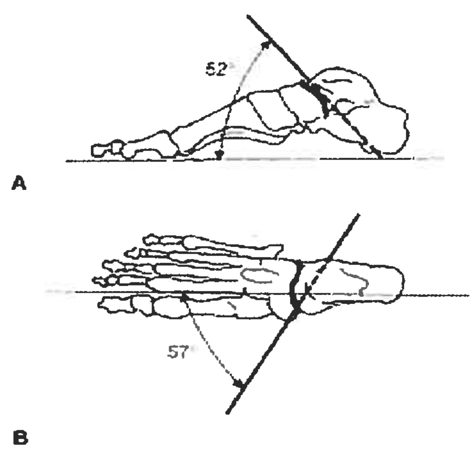

Figure 1.4 Oblique axis of the transverse tarsal joint. A, Lateral view. B, Top view (Manter 1941)

Generally, the transverse tarsal joint and subtalar are responsible for transforming tibial rotation into forefoot motion or vice verse. The transverse tarsal joint allows the forefoot to adjust itself against the rearfoot. By doing so, the anterior footplate is able to maintain full contact with the supporting surface.

The subtalar joint along with the ankle transfer the motion from the tibia to the foot in order to reduce stress. Mann (1993) described the coupling motion of the subtalar joint and ankle in a mitered hinge model. This model explains that as the tibia internally rotates in the transverse plane, the rearfoot at the subtalar joint everts in the frontal plane. This occurs during early stance phase of gait. Conversely, external rotation of the tibia during the late stance phase, causes inversion of the rearfoot (Figure 1.5). Section 2.1 will describe how the orientations of rotation axes of transversetarsal joint could affect on rigidity or flexibility of the foot.

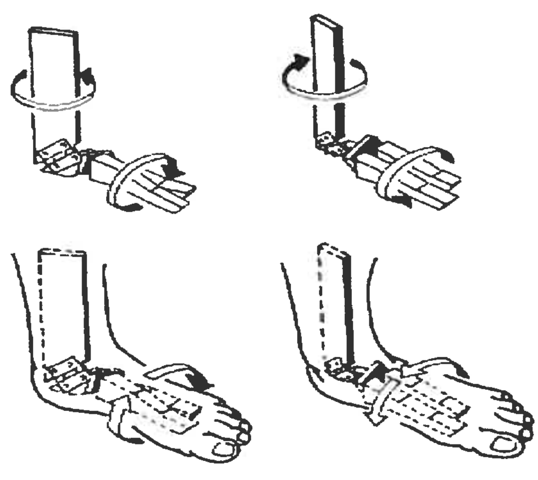

Figure 1.5 Mitered hinge model of leg, ankle, and subtalar joint motion (Nordin & Frankel, 2001)

1.2 Subtalar Joint and Its Proximal Joints and Segments in Standing Posture

The subtalar joints and ankles with their more proximal segments contribute to postural adjustments during standing (Hoogvliet et al., 1997; Tropp & Odenrick, 1988). To control standing balance, the subtalar joints and ankles cause the body to oscillate as an inverse pendulum in the frontal and sagittal planes (Hoogvliet et al., 1997; King & Zatsiorsky, 2002). As balance becomes more of a challenge, during such activities as standing on an unstable surface, the body moves as a multi-link segment about the hips, pelvis and trunk. This takes place when adjustments at the ankles and subtalar joints are no longer sufficient to control standing balance (Tropp & Odenrick, 1988).

Foot and leg positions affect the maintenance of posture by the subtalar joints and ankles (Hoogvliet et al., 1997; King & Zatsiorsky, 2002). It is believed that when the foot and leg are misaligned, mechanical and proprioceptive properties are altered (Nigg et al., 1999; Nordin & Frankel, 2001). The mechanical changes could be related to the orientation of the axes of the calcaneocuboid and talonavicular joints (Nordin & Frankel, 2001). Elftman (1960) reported that when the subtalar joint is everted, the axes of these joints are parallel and the foot is mobile. As the subtalar joint inverts, these axes converge to lock the transverse tarsal joint, rigidifying the midfoot as shown in Figure 1.1. In terms of proprioceptive properties, the maximization of muscle activity and the resulting fatigue can occur due to joints misalignment and increasing soft tissue vibration (Nigg et al., 1999). Therefore, modifying the orientation of the foot-joint axes and propriceptive properties could increase the mobility or rigidity of the subtalar joint and ankle. This could perturb the contributions to standing posture by the more proximal joints and segments.

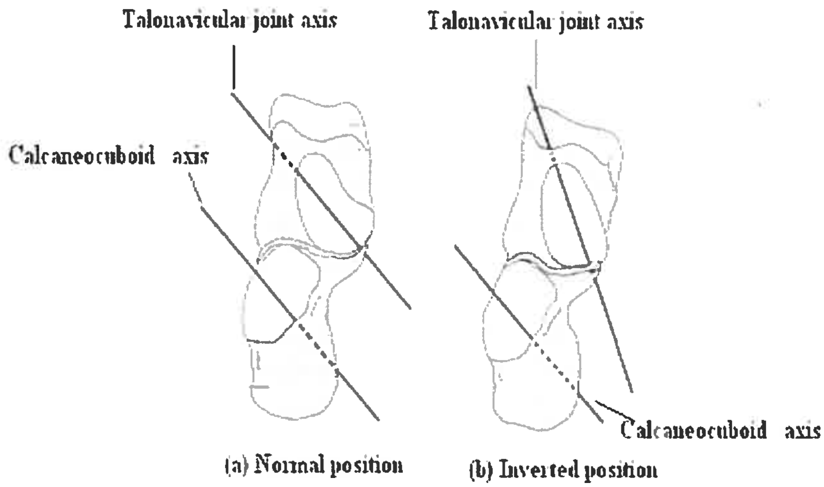

Figure 1.6 Orientation of the calcaneocuboid and talonavicular joints axes in (a) normal, and (b) inverted position of foot (Nordin & Frankel, 2001).

The amplitude and velocity of the centre of pressure (COP) are used to assess standing stability (Baier & Hopf, 1998; Hertel et al., 2001). The COP is the point of application of the ground reaction forces within the base of support. Surface covered by its excursion or sway area is indicative of standing imbalance. Hertel et al. (2001) reported that individuals with cavus feet display a large sway area. They suggested that this large sway area could be due to the limited range of motion between the subtalar and midtarsal joints but the large motions at the hips and proximal segments. Tropp et al. (1984) concluded that functional instability as demonstrated by sway strategy could not show any limitation of lower-limb joints motions. These could be related to the selected parameters which could not demonstrate any contributions of lower-limb joints during standing. Furthermore, the effect of feet positions on standing balance was not address in their study. The first study of this thesis focuses on the inter-relationships between the subtalar joint and its more proximal joints, pelvis and trunk when the foot is oriented in different directions in a single-limb stance test. Single-limb stance can better demonstrate the contributions of joints and segments in maintaining posture because of the increasing the challenge of maintaining equilibrium compared to double-limb stance (Riemann et al., 2003).

A major assumption in the first study resided in the inference of foot interaction with the tibia resulting in lower-limb injuries. These interactions are highly dynamic and cannot satisfactorily be demonstrated in standing condition (Nigg, 1987). Thus, the second and third studies assess foot and tibia coupling motion in different running conditions.

1.3 Forefoot-Rearfoot Coupling Motion and Tibial Rotation during Running

Excessive rearfoot eversion and excessive tibial rotation were associated with various running injuries (Stacoff et al., 2000). Excessive rearfoot eversion was related to Achilles tendonitis (Clement et al., 1981; Smart et al., 1980) and shin splints (Viitasalo & Kvist, 1983) whereas excessive tibial internal rotation was associated with the development of knee injuries (van Mechelen, 1992). It was speculated that excessive rearfoot eversion forces the Achilles tendon laterally producing an asymmetric stress distribution across the tendon and leading to Achilles tendonitis (Clement et al., 1981). Additionally, excessive tibial rotation could track excessively the patella and cause femoral pain syndrome (Stergiou, 1996). There is evidence that motion at the midfoot contributes significantly to overall foot motion and tibia during walking and running (Hunt et al., 2001; Pohl et al., 2006).

Forefoot motion with respect to rearfoot was modeled as a twisted plate (Hunt et al., 2001). This model suggests that, during running, the forefoot produces counter motions with respect to the rearfoot segment (Hunt et al., 2001; Sarraffian, 1993). Nordin and Frankel (2001) suggested that from heel-strike through foot-flat, the rearfoot is everted and the forefoot is flexible for absorbing shock and adapting itself to irregularities in the ground floor surface. Further, Johanson et al. (1994) reported that a deformity in the frontal plane motion of forefoot, such as the forefoot varus, resulted in excessive rearfoot eversion, which allowed the medial metatarsal heads to contact the weight-bearing surface. Shoes or wedges under the foot could cause the load to shift from the lateral side of foot to its medial side with maximum pressures under the first and second metatarsal heads (Soames, 1985). This may be related to changes in the directions of the ground reaction forces and ankle moment (Nordin & Frankel, 2001).

While clinical studies frequently model the foot as a single rigid body (Nigg et al., 1993; Stacoff et al., 2000), its many articulations keep it from acting as a simple hinge joint (Hunt et al., 2001; Pohl et al., 2006). Furthermore, in vivo studies of the forefoot motions, subjects were tested in barefoot condition to enable tracking of markers on the forefoot (Hunt et al., 2001; Pohl et al., 2006). Since footwear could affect on the three-dimensional forefoot motion coupling with the rearfoot, their effects on tibial rotation remained unknown. In the second study of this thesis, the effect of shoewear on forefoot-rearfoot coupling motions will be investigated with respect to tibial angular motion in running. Understanding of forefoot-rearfoot coupling motion patterns and their effects on tibial rotation could point out the importance of forefoot posted orthoses in controlling excessive motion in tibia (Johanson et al., 1994).

1.4 Rearfoot and Tibial Rotations in Relation to Ground Reaction Forces and Knee Moments

Runners are potentially at risk to lower-limb joint injuries because they experience a large number of repetitive ground reaction forces (Cole et al., 1995; Hreljac et al., 2000). The amplitude and timing of rearfoot eversion, tibial internal rotation and knee flexion were proposed as key components in the reduction of external forces in the lower-limb during the first half of stance phase (Hreljac et al., 2000; Naster et al., 2003; Stergiou & Bates, 1997). Foot orthoses are often prescribed to control these motions, which may reduce impact shock absorbing property, at the cost of increased loading at the knee during running (Bellchamber & van den Bogert, 2000). The third part of this thesis will address this issue.

Wedged foot orthoses are a common treatment to redistribute of loads to the lower-limb joints. Hurwitz et al. (2000) and Hunt et al. (2006) proposed that increased load on the medial aspect of the tibial plateau and femoral condyle was primarily due to knee adduction moment. This could be responsible for greater iliotibial band friction in runners (Andriacchi et al., 1985) and knee pain in patients with osteoarthritis (Yasuda & Sasaki, 1987). Andriacchi et al. (1985) reported that a large knee adduction moment was associated with the distribution of load in the medial compartments of the knees. Hunt et al. (2006) proposed that the resultant ground reaction force and its lever arm in the frontal plane are primarily two independent variables in the knee adduction moment. Foot orthoses could change these two variables by manipulating foot and leg movements.

Conflicting results, however, were reported on rearfoot and tibial rotation control by means of foot orthoses to reduce peak adduction moment during walking (Kakihana et al., 2005; Keating et al., 1993). Yet, there has been little investigation on knee adduction moment in runners who are affected by rearfoot and tibial rotations brought about by foot orthoses. The third study will attempt to determine changes in the peak ground reaction force and knee adduction moment during the stance phase of running. Here, the rearfoot and tibial rotations were manipulated using semi-rigid orthoses. It was postulated that the foot orthoses might not affect only reduction of rearfoot eversion and tibial internal rotation, but also ground reaction forces and joint moments which are affected by control of lower-limb motions.

1.5 Structure of the Thesis

The general objective of this research project is to determine the effect of foot-angle changes on the kinematics of lower limb joints and their relationships with ground reaction force and knee adduction moment. A review of the role of lower-limb joints and upper segments in standing posture, and of lower-limb joint motions related to running injuries is presented in Chapter 2. Specific objectives of this thesis conclude this chapter. Chapter 3 presents the kinematic and kinetic models of foot and ankle and describes the experimental methods used to estimate three-dimensional movement of lower-limb joints and knee moments. The following three chapters are papers, two of which have been published. The first paper (Chapter 4) presents an assessment of the variability in the subtalar joint and the ankle, and their more proximal joints and segments. Of particular importance, is an examination of the foot in different orientations during single-limb stance. The second paper (Chapter 5) provides an estimate of the forefoot-rearfoot coupling patterns and their effects on tibial rotation, in both barefoot and shod running conditions. This is followed by a determination of the rearfoot and tibial rotations, as they relate to peak vertical ground reaction force and knee adduction moment, during shod running and shod with orthoses in Chapter 6. The findings are discussed in Chapter 7 and this is followed by the conclusions in Chapter 8.

Chapter 2

2. REVIEW OF LITERATURE

Despite the wealth of literature regarding to lower-limb joint mechanics, the role of foot orientations in standing posture and normal gait is not well understood. In standing, ankle and hip strategies are well documented in relation to the control of posture after an external perturbation (Horak & Nashner, 1986; Nashner & McCollum, 1985). However, little known about how posture through body joints and segments can be affected by the alteration of foot positions. In addition, lower-limb injuries resulting from excessive motions in the rearfoot and tibia are relatively well-understood during running; few have investigated the factors related to excessive motions.

This chapter reviews the studies related to the kinematics of lower-limb joints, pelvis and trunk, and their role in the maintenance of posture during double and single-limb standing. Additionally, the excessive coupling motion of rearfoot and tibia during gait will be reviewed, and the effect of forefoot motion on this coupling motion will be discussed. The contribution of rearfoot eversion and tibial internal rotation to both, the vertical ground reaction force and knee moment will be outlined. Finally, the chapter ends with a delineation of this thesis's specific objectives.

2.1 Postural Strategies in Maintaining Standing Posture

In adults, postural adjustments during double-limb stance on a flat surface are achieved using ankle and hip strategies in the sagittal plane. In the ankle strategy, muscle activity extends primarily from the distal to the proximal joints (Horak & Nashner, 1986). The hip strategy involves the generation of torque at the hip, rather than at the ankle, and extends motion at the trunk, pelvis and hip, using a proximal-distal sequence of muscle activations (Horak & Nashner, 1986). The hip strategy makes larger corrections possible, (Nashner & McCollum, 1985) whereas the ankle strategy is limited by the foot's ability to exert torque as it makes contact with the surface (Tropp & Odenrick, 1988).

Although research in postural stability has been mainly on double-leg stance, single-leg stance occurs frequently in the course of daily living, as well as in many sport activities such as running. In addition, the challenge of maintaining single-leg equilibrium may better clarify the contribution of different joints and segments in maintaining posture. Hoogvliet et al. (1997) described two frontal plane strategies to maintain posture during single-limb stance. The first refers to body rotation with the subtalar joint acting as the center of rotation. The second is a hip strategy which occurs about the hip with a relatively large displacement of the centre of pressure. Riemann et al. (2003) reported that the ankle was the main source of posture maintenance during single-limb standing. As the challenge of balance became greater- for example, on unstable surfaces- controlling posture at the hip and trunk was demonstrated by increasing their angular displacements. They calculated angular displacement from the vector sum of the three separate angular position vectors. Inman and Mann (1978) reported that initial control of posture was by the subtalar joint and ankle movements. Further, Tropp and Odenrick (1988) explain that, if the abnormalities such as ankle injuries are generated in the foot and leg, compensatory motions to maintain posture will appear in the upper segments of the body.

A clearly drawn definition of compensatory motion comes from Nicolopoulos et al. (2000), who described it as a change of position or function of one part of the body, as it adjusts to a deviation of structure, position or function of another part. Structural or positional abnormalities create a recurrent or persistent demand for compensation which may result in pathology (Albert & Chen, 1996). From a practical viewpoint, when the foot is tilted during standing, compensation is accomplished with large movements in the upper joints and segments in order to keep the centre of mass within the base of support (Nashner & McCollum, 1985; Tropp & Odenrick, 1988). Aligning the foot and leg in order to maintain posture, may increase the contribution of subtalar joint and ankle, while decreasing upper segment contributions.

Foot orthoses, combined with postings which act as wedges, are often prescribed to improve posture by attempting to align the foot and leg. Because foot and leg segments are linked by the ankle and subtalar joint, tilting the foot by means of a wedge affects the COP position. In a Rocker Shaped model of foot during single limb stance, Hoogvilet et al. (1997) reported that the amplitude and velocity of the

COP could decrease once the foot is tilted to a given amplitude and direction in the frontal plane. This, in turn, could have adverse effect on postural control. Therefore, upper joints and segments could compensate by increasing their contributions in maintaining posture (Nicolopoulos et al., 2000). These contributions are still unknown in a single-limb stance.

Guskiewicz and Perrin (1996) reported that subjects fitted with foot orthoses following ankle injuries, sway more than the uninjured people when assessed on a single-limb stance. In contrast, Hertel et al. (2001) found that orthotics, irrespective of design or posting, were ineffective at reducing postural sway after lateral ankle sprain. Similarly, Tropp et al. (1984) concluded that functional instability as demonstrated by sway strategy could not show any limitation of ankle motions. This finding could be related to the selected parameters, which failed to demonstrate any contributions of lower limb joints in a single-limb stance test.

In summary, the literature reveals that when the foot is tilted in different directions during single-limb stance, greater compensatory actions were taken by the upper joints and segments. Determining the amplitude and velocity of the COP is insufficient to detect these compensations (Baier & Hopf, 1998; Tropp et al., 1984). A kinematic approach can establish the contributions made by both, the subtalar joint, and its upper joints and segments in single-limb stance testing.

2.2 Lower Extremity Coupling Kinematics during Running

The forefoot-rearfoot motion patterns in the mid-foot joint, along with their effects on tibial rotations, are not well understood during the stance phase of running. These patterns are highly dynamic that cannot be demonstrated with standing conditions. Pronation of the subtalar joint, in respect to the talus, consists of eversion, abduction, and dorsiflexion of the calcaneus (Donatelli, 1993). Pronation occurs in the first half of stance phase of the walking or running cycle, allowing the foot to accommodate to uneven surfaces to better attenuate shock (Isman & Inman, 1969; Lundberg, 1989; Root et al., 1966). During pronation, when the calcaneus is fixed to the ground, it cannot abduct relative to the talus. Therefore, due to the tight ankle mortise, the tibia internally rotates as the talus adducts. Thus, rearfoot eversion, and tibial internal rotation occur relatively synchronously during the first half of stance (Buchbinder et al., 1979; Levens et al., 1948; Tiberio, 1987). Abnormalities in foot function may influence the timing and amplitude of these segments during gait.

Multi-segment foot models have provided evidence that the mid-foot joints contribute more to the overall foot motion than was previously believed (Hunt et al., 2001; Pohl et al., 2006). According to Saraffian (1987), the foot behaves as a twisted plate, in that the arch raises or lowers according to the counter motions of the forefoot and rearfoot segments. During heel-strike through foot-flat, the rearfoot is everted and the forefoot should be flexible from the mid-tarsal joints to absorb shock and adapt itself to irregularities in the ground floor surface in an efficient gait (Nordin & Frankel, 2001). Lundberg et al. (1989) found that the frontal plane motion occurred primarily at the talonavicular joint, rather than at the talocalcaneal joint. Johnson et al. (1999) reported that when an inversion contracture of the forefoot occurs at the mid-tarsal joints, abnormal rearfoot pronation results, allowing the medial metatarsal heads to contact the weight-bearing surface. Excessive and prolonged pronation causes abnormal delay in both, external and internal rotation, resulting in various symptoms in the lower-limb (Bate et al., 1978; Hamill et al., 1992). A less efficient gait is the consequence of these biomechanical abnormalities, and can result in an overuse syndrome (Weik & Martin, 1993). These investigations indicate that mid-foot joints are complex structures that contribute to the overall foot motion during locomotion. It will be invaluable to have a clear picture of the three dimensional motion which occurs at the mid-foot, relative to that at the ankle complex. Armed with this knowledge, the clinician will be better able to manage joint dysfunction and the effective prescription of orthotic devices.

In the stance phase of running, the subtalar joint has a coupling motion with the forefoot and tibia. The orientation of the subtalar joint axis influences its range of motion. Root et al. (1966) reported that, with a subtalar joint axis orientation of 41°, the range of motion was between 22° and 55°. At 42°, the range of motion was lower from 29° to 47° (Manter, 1941). Lundberg et al. (1989) reported an average orientation of subtalar joint axis of only 32° with a range of 14° to 39,8°. It is difficult to measure the orientation of the subtalar joint axis directly without invasive techniques in vivo study. Thus, a number of authors have examined the relative amounts of both rearfoot eversion and tibial internal rotation motion (EV/TIR), which is indicative of the orientation of the subtalar joint (Nawoczenski et al., 1995; Nigg et al., 1993; Stacoff et al., 2000 ; Williams et al., 2001).

The EV/TIR ratio provides a measure of the relative motion between rearfoot eversion and tibial internal rotation excursions. It is measured from heel-strike to the respective peaks which occur around mid-stance. This ratio also may be altered with the use of footwear. The EV/TIR ratio varied between 0,65 in the normal shod (Stacoff et al., 2000) and 2,40 in the barefoot conditions (Pohl et al., 2006). The EV/TIR ratio is often used to determine if the tibia has a relatively greater motion with respect to the rearfoot, in regard to a discrete data point (Nawoczenski et al., 1995; Nigg et al., 1993; Williams et al., 2001). Thus, it may not be helpful in understanding two segments coupling patterns throughout the stance phase (DeLeo et al., 2004).

Dynamic systems theory is another technique used to examine the coupling motions in two adjacent segments throughout the stance phase. This technique uses a continuous relative phase (CRP) measure to detect in-phase or out-of-phase relations between two adjacent segments. Briefly, the CRP is calculated by first generating a phase plane portrait of normalized angular velocity, as plotted against normalized angular position for two segments or joints as shown in Figure 2.1. Phase angles are then calculated for all points in the phase plane portrait. Finally, the CRP angle is plotted by subtracting the phase angle of the distal segment from the phase angle of the proximal segment. CRP values can range between -180° and 180°. A zero value indicates complete in-phase coupling, while 180° or -180° indicates complete out-of-phase coupling (Hamill et al., 1999; Li et al., 1999; Stergiou et al., 2001).

For a group of healthy runners, Hamill et al. (1999) reported a CRP angle of approximately 45° for the rearfoot and tibia coupling motion pattern at the foot-strike phase. This pattern transitioned quickly into a more in-phase relationship (CRP approximately 10°) that was maintained throughout the remainder of stance.

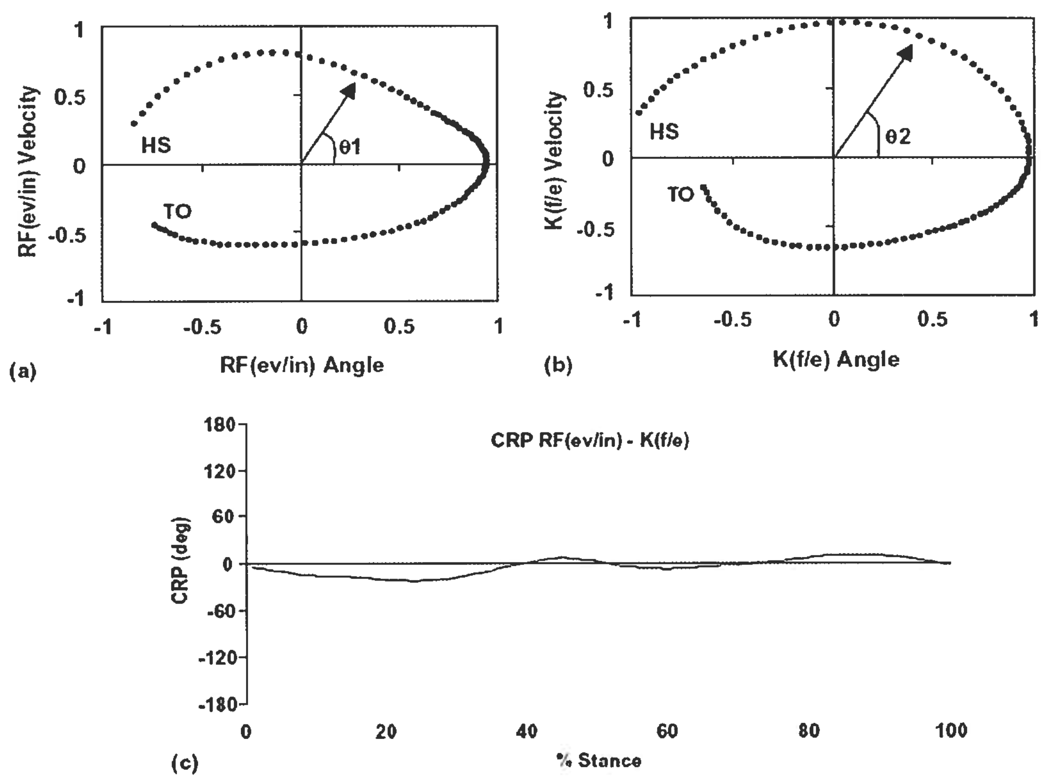

Figure 2.1 (a) Phase angle plots of normalized angular displacement versus normalized angular velocity curves for rearfoot eversion/inversion, (b) tibial rotation, and (c) the continuous relative phase (CRP) plot for coupling motion of rearfoot eversion/inversion (RF ev/in) and knee flexion/extension (K f/e) (DeLeo et al., 2004).

Ferber et al. (2002) used the CRP to examine joint coupling in healthy and injured runners. These authors reported an in-phase relationship for rearfoot eversion-tibial internal rotation for the healthy group and a more out-of-phase relationship for the injured group, throughout stance. These data suggest that a more out-of-phase relationship for rearfoot eversion-tibial internal rotation may be related to injury. Stergiou et al. (2001) studied rearfoot eversion-tibial abduction coupling in the frontal plane and reported an out-of-phase relationship at heel strike which transitioned into an in-phase relationship by mid-stance. From mid-stance to toe-off, rearfoot eversion-tibial abduction transitioned back to an out-of-phase relationship. These data suggest that (1) coupling relationships are different for different segments or joint combinations, and (2) that they may change throughout stance phase.

Hunt et al. (2001) calculated the three-dimensional angular motions of the forefoot with respect to the rearfoot, during walking. They reported that the angular range of motion of forefoot was 12°, 4° and 10° in the sagittal, frontal and transverse planes during the stance phase, respectively. They did not, however, describe the relationship motion between forefoot and rearfoot during stance phase. Pohl et al. (2006) used a cross-correlation technique to evaluate the relationship between forefoot and rearfoot motion during barefoot running. They indicated that rearfoot eversion/inversion was highly correlated to both, forefoot plantar/dorsiflexion ($r < -0,85$) and abduction/adduction ($r > 0,94$), with no phase shift during the stance phase of barefoot running. However, no significant relationship was observed between rearfoot eversion/inversion and forefoot eversion/inversion ($r = -0,02$). This finding shows that forefoot and rearfoot coupling motion in the frontal plane had a non-linear relationship. Because cross-correlations are based on the assumption that linear relationships exist between two adjacent segments (Pohl et al., 2006), they are not useful in determining the degree of linkage between segments that have a non-linear relationship (Sideway et al., 1995). Using the CRP technique could determine an in-phase or out-of-phase relationships between forefoot and rearfoot. Also, forefoot-rearfoot coupling motion can be compared with the rearfoot-tibia and tibia-knee coupling motions during running, which has not yet been described.

In summary, forefoot and rearfoot variations as well as the amount of tibial rotations could have a significant effect on foot function in gait and running. The use of relative angular motion could not detect this coupling during the stance phase, because this value is measured from particular discrete time event. Additionally, cross correlations determine the similarity of the motion of two segments which have a linear relationship. To provide a description of continuous forefoot-rearfoot coupling motions, the CRP could determine an in-phase or out-of-phase relationship at any point within the stance phase of gait.

2.3 Rearfoot and Tibia Rotations in Relation to Knee Moment and Ground Reaction Force

The control of excessive rearfoot eversion and tibial internal rotation is considered one of the most important correction functions performed by foot orthoses. These would work to correct, align, or limit the skeletal movement of foot and leg (Nigg et al., 2001). Hreljac et al. (2000) point out that the control of these motions during the stance phase may not be the primary function of such interventions. In fact, variations in the amplitude of lower-limb kinematics could contribute to changes in the kinetic parameters. For example, moment is primarily calculated as the product of force and its lever arm. Foot orthoses could change the amount of forces and lever arms by aligning and limiting foot and leg movement affecting the loading distribution on the proximal joints.

The peak knee adduction moment was associated to overuse running injuries. It was suggested that knee adduction moment may cause an increase in load in the medial aspect of the tibial plateau and femoral condyle thereby, increasing knee pain in runners (Hurwitz et al., 2000; Hunt et al., 2006). It is proposed that knee injuries due to high loading specifically due to the adduction moment, can be reduced by variations in the amplitude of lower-limb kinematics with posted orthoses.

In addition, repeated overloading resulted in degenerative changes to the articular cartilage in animal models (Radin et al., 1985). These results indirectly suggest that runners who experience high loading may be at risk to degenerative joint disease. Ground reaction force measurements and particularly the vertical force have typically been used to describe the loading conditions in running (Andriacchi, 1994; Cole et al., 1995; Perry & Lafortune, 1995; Nigg et al., 2001). Messier et al. (1988) and Grimston et al. (1994) reported that the magnitude of active ground reaction force was a significant discriminator between groups of injured and uninjured runners with stress fractures. Figure 2.2 illustrates a double hump pattern of the vertical ground reaction force during the stance phase of 4 m/s heel-toe running. The active reaction force is the peak vertical force that occurs during mid-stance of running. The impact force occurs when the subtalar joint was inverted earlier than 50 ms after first contact (Frederick et al., 1981).

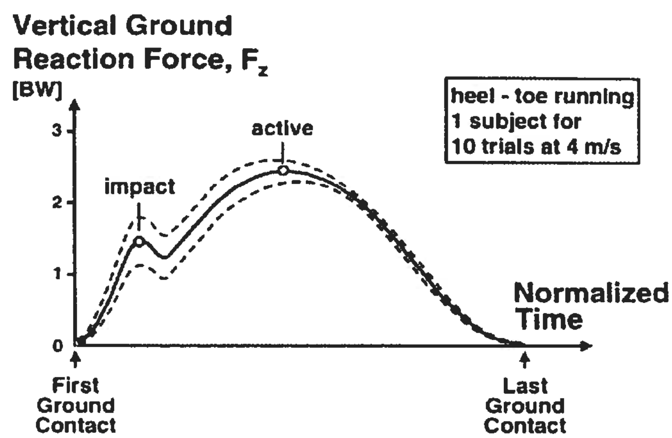

Figure 2.2 Illustration of impact and active vertical ground reaction force for 10 trials at the speed of 4 m/s (Nigg, 2001).

Ground reaction force can be attenuated during early stance phase of running. This can happen through synchronous timing and proper movement of rearfoot eversion, tibial internal rotation and knee flexion (Stergiou & Bates, 1997). Perry and Lafortune (1995) indicated that the active ground reaction force was increased when normal rearfoot eversion was prevented. To the contrary, no reduction was seen when normal rearfoot eversion was excessive during running. Mündermann et al. (2003) reported that impact force was reduced when maximum rearfoot inversion was increased. No significant changes, however, were found for the active peak ground reaction force when rearfoot eversion was decreased. These findings suggest that the amplitude of lower-limb motion could contribute to the vertical ground reaction force during running. It is unknown if using foot orthoses to control rearfoot eversion and tibial rotation is effective at cushioning the active vertical ground reaction force.

Bates et al. (1978) suggests that timing between the subtalar joint and knee actions can reduce ground reaction forces. These actions are accompanied by internal tibial rotation (Nuber, 1988). During the support phase of running, peak knee flexion and peak pronation occurs at approximately the same time during mid-stance (Bates et al., 1978). Prolonging rearfoot eversion and tibial internal rotation later than mid-stance phase could cause a disruption in the timing pattern, leading to a failure to absorb the ground reaction force. To our knowledge, no investigation has verified that foot orthoses ameliorate this timing disruption and could therefore better absorb forces and reduce joint loading.

To summarize, when foot orthoses control the foot pronation, the peak vertical reaction force might be changed, because the cushioning forces could be due to the synchronous timing and amplitude of foot pronation. This could also affect loading distribution on the proximal joints. Previous studies have compared only the effect of different orthoses on lower extremity kinematic and kinematic (Mündermann et al., 2003; Nester et al., 2002). Even so, the relationships among amplitude and temporal characteristics of lower extremity kinematics and ground reaction forces and moment have not been investigated. Thus, one of the aspects of this thesis was to ascertain if the peak ground reaction force and knee moment could be altered when rearfoot eversion and tibial internal rotation are manipulated by means of foot orthoses.

2.4 Specific Objectives of Thesis

The effect of foot angle changes on the kinematics of lower-limb joints and their effects on knee moment and ground reaction forces, is the core of this thesis. In the first study, we hypothesized that wedges located under the foot will affect equally the plane of movement joint angle variability at the lower limb joints, pelvis and trunk. Additionally, changes in angle variability will occur equally at these joints and segments to maintain posture during single-limb stance.

In the second study, we hypothesized that tibial internal rotation is increased when the forefoot-rearfoot coupling patterns are modified to a more in-phase relationship with shoewear during the stance phase of running. The purposes were: i) to compare the excursion of tibial internal rotation and rearfoot eversion from heel-strike to peak value during the stance phase of running in barefoot versus shod conditions, ii) to determine differences in mean relative phase angle of the forefoot eversion/inversion and rearfoot eversion/inversion, forefoot dorsi/plantarflexion and rearfoot eversion/inversion, forefoot adduction/abduction and rearfoot eversion/inversion during the stance phase of barefoot versus shod running.

The objectives of third study were to make observations under shod and shod-with-orthoses conditions, and compare them in light of the magnitude and temporal characteristics of the rearfoot eversion, tibial internal rotation, peak ground reaction force and knee adduction moment. We hypothesized that i) foot orthoses decrease the amplitude of rearfoot eversion, tibial internal rotation, and knee adduction moment, but increases the peak ground reaction force; ii) foot orthoses synchronize time to peak rearfoot eversion, tibial internal rotation, peak ground reaction force and knee adduction moment; iii) the amplitude of rearfoot eversion, tibial internal rotation are correlated to the peak ground reaction force and knee adduction moment.

Chapter 3

3. KINEMATIC AND KINETIC MODELS OF THE FOOT AND ANKLE

This chapter describes a three-rigid segment kinematic model of foot and ankle. It was adapted from Kidder et al. (1996) by adding virtual markers on the foot and tibia to facilitate data acquisition and joint angle calculation. Furthermore, the method to calculate the three-dimensional (3D) ankle and knee moments is presented where the foot and tibia are represented as rigid segments. These models will be applied to assess the coupling motions of forefoot-rearfoot as well as rearfoot-tibia described in Chapter 5 and to estimate the relationship between foot kinematics and knee moments reported in Chapter 6.

3.1 Tibia and Foot Kinematic Model and Three-Dimensional Joint Angles Calculations

To determine the 3D movements of the forefoot with respect to the rearfoot and their effect on tibial rotation, a three-segment rigid body kinematic model based on Kidder et al. (1996) was adapted by modifying marker configurations in the forefoot and tibia as well as including virtual markers during shod running. A local coordinate system (LCS) was defined on the tibia, rearfoot and forefoot segments by determining the inter-segment axes and rotations according to the ISB Joint Coordinate System recommendation (Wu et al., 2002). Then, virtual marker positions were calculated with respect to their local coordinate system. The three-dimensional joint angles were obtained by the method proposed by Grood and Suntay (1983). This recent method is a widespread clinical method for expressing a distal segment orientation relative to the next proximal segment during gait. Since the rearfoot is fixed on the ground during the first of stance phase of running, tibial internal rotation is defined as transverse motion of the foot with respect to the tibia by this method (DeLeo et al., 2004). The three-segment rigid body kinematic model was based on that of Kidder et al. (1996) and four additional virtual markers were used to describe the three-dimensional rotations of the forefoot (FF), rearfoot (RF) and tibia (TB). The three-rigid segment model is shown in Figure 3.1 with marker placement depicted in Figure 3.2. Firstly, the rearfoot motion was expressed relative to the tibia to represent the combined subtalar and talocrural joints motions (Wu et al., 2002). Secondly, the forefoot motion was described with respect to that of the rearfoot representing mid-foot motions (Pohl et al., 2005). The inter-segment axes and rotations were defined according to the Joint Coordinate System recommendation (Wu et al., 2002).

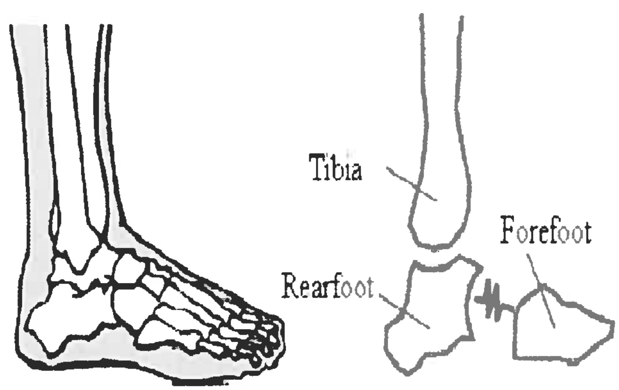

Figure 3.1: Schematic diagram of the three segment model of the foot segments and tibia (Carson et al., 2001).

3.1.1 Marker Configuration

Table 3.1 and Figure 3.2 present thirteen 16 mm diameter reflective markers fixed to the right foot and tibia. Three types of markers were used, namely, anatomical, virtual and technical. Of these, ten markers were positioned on bony landmarks to define the anatomical coordinate system of the segments. Of these, four markers were detached for the experiments after calibration. They were detached during running with sandal because in the second study, we proposed to evaluate changes in forefoot-rearfoot coupling brought about by the use of sandals. Since the sandals cover the base and middle parts of the forefoot, marker placement on these parts would be impossible without altering the footwear. Furthermore, marker dropout, skin movement artifacts and hidden markers could occur particularly on the medial side of the lower limb during the running trials. These four markers are called virtual markers since there were absent during the experimentation. Besides these ten markers, three technical markers were placed on the tibia and forefoot. These technical markers were used to estimate the virtual marker positions during the running experiments and calculate the rotation matrices. Generally, the virtual markers allow to identify the location of key anatomical landmarks with respect to other markers in order to determine the anatomical motions of each rigid segment during running trials. The technical markers, however, cannot indicate the anatomical motion of each rigid segment.

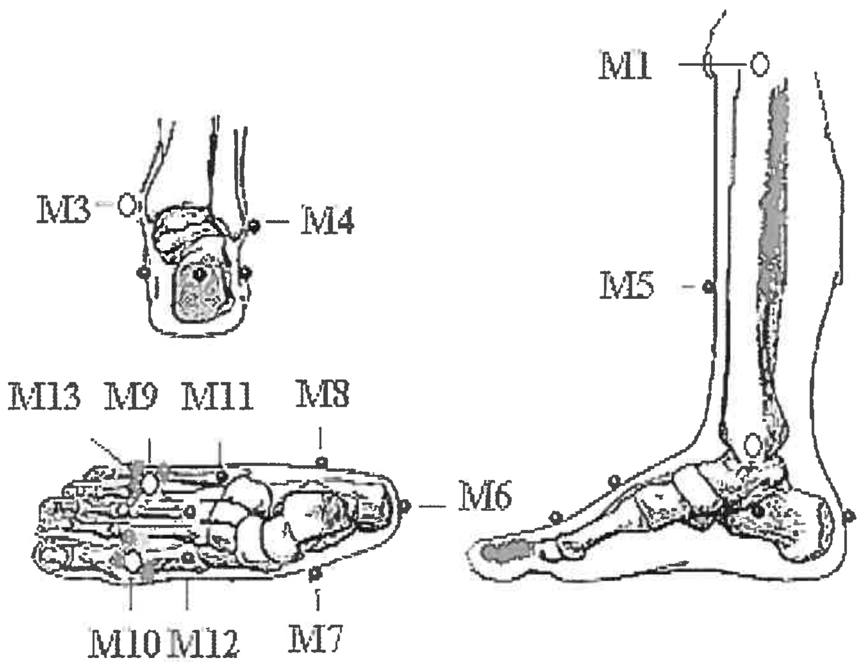

Figure 3.2 Markers configuration located on the tibia, rearfoot and forefoot based on Kidder et al. (1996) as well as four virtual markers (○). The alpha-numeric symbols are described in Table 3.1 (Kidder et al., 1996).

Table 3.1 Anatomical, Virtual and Technical Markers, Symbols and their Locations on the Tibia, Rearfoot and Forefoot.

Segmentss	Symbols	Anatomical locations	Type of markers
Tibia			
	M1	Medial tibial tubercle	Virtual marker/Anatomical marker
	M2	Lateral tibial tubercle	Anatomical marker
	M3	Medial malleolus	Virtual marker/Anatomical marker
	M4	Lateral malleolus	Anatomical marker
	M5	Anterior middle aspect of the tibia	Technical marker
Rearfoot			
	M6	Posterior calcaneus	Anatomical marker
	M7	Medial calcaneus	Anatomical marker
	M8	Lateral calcaneus	Anatomical marker
Forefoot			
	M9	Fifth metatarsal head	Virtual marker/Anatomical marker
	M10	First metatarsal head	Virtual marker/Anatomical marker
	M11	Fifth metatarsal base	Anatomical marker
	M12	First metatarsal base	Technical marker
	M13	Between metatarsals II and III	Technical marker

3.1.2 Global and Local Coordinate Systems

A global coordinate system (GCS) is required to match force-plate data with video-based information in a fixed reference frame. The origin of the global coordinate system is located in one of the corners of the force plate, and its positive axes are shown in Figure 3.3. A local coordinate system (LCS) was defined on the tibia, rearfoot and forefoot segments (i), of the right limb. The X_i axis is oriented forwards; the $\mathbf{Y}_i$ axis upwards and $\mathbf{Z}_i$ axis is directed towards the right.

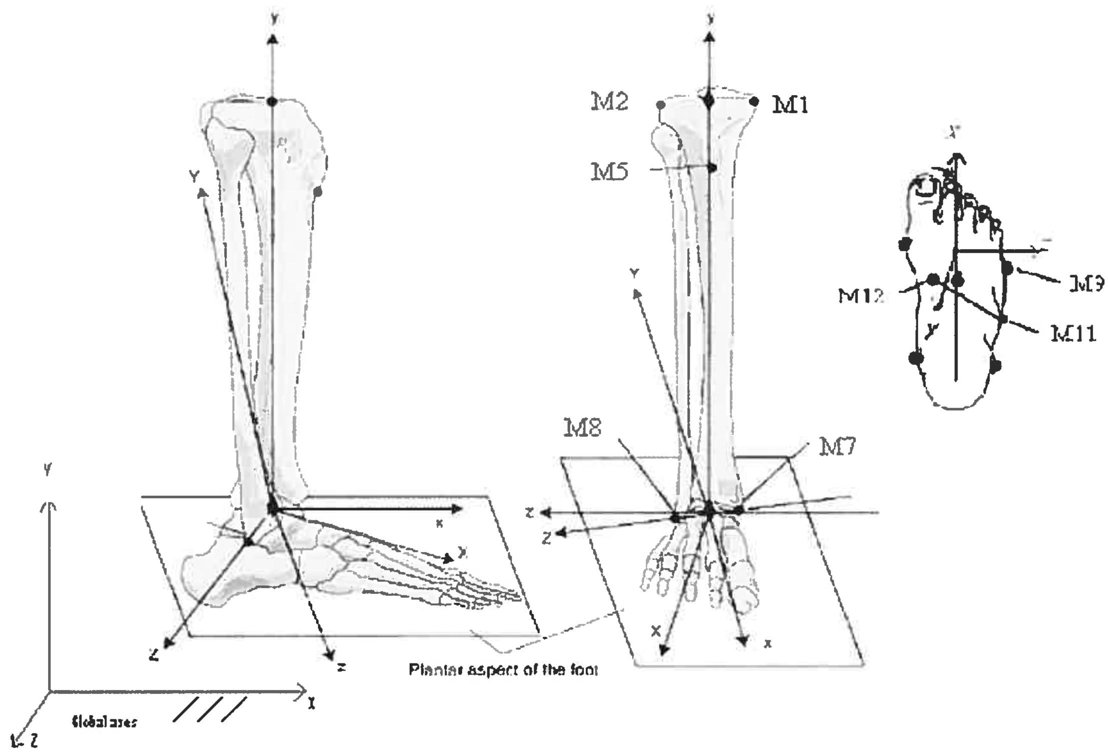

Figure 3.3 Global coordinate system and local coordinate systems of the tibia (XYZ), rearfoot (xyz) and forefoot (xyz) (Carson et al., 2001; Wu et al., 2002).

The origin (O_i) of the LCS of segment i was located at the joint center. The tibia joint center coordinates $\mathbf{O}_{TB}$ were

$$\mathbf{O}_{TB} = \frac{\mathbf{M}_3 + \mathbf{M}_4}{2} \qquad (3.1)$$

where the vector position $\mathbf{O}_{RF}$ and $\mathbf{O}_{FF}$ are for the rearfoot and forefoot, respectively.

$$\mathbf{O}_{RF} = \frac{\mathbf{M}_7 + \mathbf{M}_8}{2} \qquad (3.2)$$

$$\mathbf{O}_{FF} = \frac{\mathbf{M}_9 + \mathbf{M}_{10}}{2} \qquad (3.3)$$

Positive anatomical Z axes were calculated from the lateral and medial markers in each segment as

$$\mathbf{Z}_{TB} = \mathbf{M}_4 - \mathbf{M}_3 \qquad (3.4)$$

$$\mathbf{Z}_{RF} = \mathbf{M}_8 - \mathbf{M}_7 \qquad (3.5)$$

$$\mathbf{Z}_{FF} = \mathbf{M}_9 - \mathbf{M}_{10} \qquad (3.6)$$

The line joining the ankle centre to the knee center is an interim y_{TB} axis given by

$$y_{TB} = \frac{\mathbf{M}_1 + \mathbf{M}_2}{2} - \mathbf{O}_{TB} \qquad (3.7)$$

Because it is not exactly at right angle to the tibia anatomical Z axis, the anatomical Y axes for the rearfoot $\mathbf{Y}_{RF}$ and forefoot $\mathbf{Y}_{FF}$ are

$$\mathbf{Y}_{RF} = \mathbf{Z}_{RF} \times x_{RF} \qquad (3.8)$$

$$\mathbf{Y}_{FF} = \mathbf{Z}_{FF} \times x_{RF} \qquad (3.9)$$

where x_{RF} and x_{FF} are interim x axes define as

$$x_{RF} = \mathbf{M}_7 - \mathbf{M}_6 \qquad (3.10)$$

$$x_{FF} = \mathbf{M}_9 - \mathbf{M}_{11} \tag{3.11}$$

The anatomical anterior-posterior axis of the tibia, $\mathbf{X}_{TB}$ is

$$\mathbf{X}_{TB} = y_{TB} \times \mathbf{Z}_{TB} \tag{3.12}$$

The anatomical $\mathbf{X}_l$ of the rearfoot and forefoot as well as the $\mathbf{Y}_{TB}$ for the tibia were calculated from the cross product of the two other orthogonal axes to ensure that the three anatomical axes are at right angles to each other.

$$\mathbf{X}_{RF} = \mathbf{Y}_{RF} \times \mathbf{Z}_{RF} \tag{3.13}$$

$$\mathbf{X}_{FF} = \mathbf{Y}_{FF} \times \mathbf{Z}_{FF} \tag{3.14}$$

$$\mathbf{Y}_{TB} = \mathbf{Z}_{TB} \times \mathbf{X}_{TB} \tag{3.15}$$

To obtain unit vectors (i, j, k) for each axis, the above anatomical vectors need to be normalized with dividing them by their respective norm.

$$i = \frac{\mathbf{X}_l}{\|\mathbf{X}_l\|} \tag{3.16}$$

$$j = \frac{\mathbf{Y}_l}{\|\mathbf{Y}_l\|} \tag{3.17}$$

$$k = \frac{\mathbf{Z}_l}{\|\mathbf{Z}_l\|} \tag{3.18}$$

For each segment a technical coordinate system (TCS) was calculated from coordinates of three non-collinear markers that remained during the running trials. The procedure to calculate the TCS was exactly the same as for the LCS. The origin of TCS was, however, arbitrary located on one anatomical or technical marker. The calculation of the virtual marker positions using the TCS and GCS is described below.

3.1.3 Calculation of Virtual Markers

During the experimentations, four markers (M1, M7, M9, M10) were removed and their positions $\left({}^{L}\mathbf{r}\right)$ were calculated from the product of the rotation matrix of the global coordinate system ${}^{L}_{G}\mathbf{A}$ and the virtual marker position with respect to the global coordinate system (${}^{G}\mathbf{r}$).

$${}^{L}\mathbf{r} = {}^{L}_{G}\mathbf{A}\;{}^{G}\mathbf{r} \qquad \textbf{(3.19)}$$

Reorganizing,

$${}^{L}\mathbf{r} = {}^{G}_{L}\mathbf{A}^{\mathrm{T}}\;{}^{G}\mathbf{r} \qquad (3.20)$$

where ${}^{G}_{L}\mathbf{A}^{\mathrm{T}}$ is the transpose matrix of the technical coordinate system with respect to global coordinate system.

Since the virtual marker coordinates are fixed to technical coordinate systems of the rigid segment, an average position of the virtual marker with respect to the technical coordinate system ($\overline{{}^{L}\mathbf{r}}$) is estimated. Then, the virtual marker coordinates are expressed as a function of time, t in the global coordinate system as

$${}^{G}\mathbf{r}(t) = {}^{G}_{L}\mathbf{A}(t)\,\overline{{}^{L}\mathbf{r}} \qquad (3.21)$$

expanding

$$
{}^{G}\begin{pmatrix} X \\ Y \\ Z \\ 1 \end{pmatrix} = \begin{pmatrix} \mathbf{R}_{3\times3} & \mathbf{T}_{3\times1} \\ \mathbf{0} & 1 \end{pmatrix} {}^{L}\begin{pmatrix} X \\ Y \\ Z \\ 1 \end{pmatrix} \qquad (3.22)
$$

where *X, Y, Z* are the coordinates of the virtual markers in the global, G and local, L, coordinate systems. R and T are the rotation and translation matrices between those two systems. From the equations 3.21 and 3.22, the positions of the virtual marker can be calculated in the global coordinate system during the running trials.

3.1.4 Joint Angles Calculation

Three-dimensional joint angles were obtained by the method proposed by Grood and Suntay (1983). The first rotation (γ) is about the medial/lateral axis (Z) of the proximal segment (*P*) and corresponds to flexion/extension. Eversion/inversion is given by the second rotation (α) about posterior/anterior axis (X) of distal (floating axis). The third and final rotation (β) is about proximal/distal axis (Y) of the distal segment (*D*). It is performed after the first and second rotations and indicates internal/external rotation. The orientation matrix, ${}^{P}_{D}\mathbf{A}$ is given by

$$
{}^{P}_{D}\mathbf{A} = \mathbf{R}(Z,\gamma)\mathbf{R}(X,\alpha)\mathbf{R}(Y,\beta) \qquad (3.23)
$$

where R is the rotation matrices about their respective axis. These rotations can be expressed as

$$
\begin{pmatrix} \cos\gamma\cos\beta - \sin\gamma\sin\alpha\sin\beta & -\sin\gamma\cos\alpha & \cos\gamma\sin\beta + \sin\gamma\sin\alpha\cos\beta \\ \sin\gamma\cos\beta + \cos\gamma\sin\alpha\sin\beta & \cos\gamma\cos\alpha & \sin\gamma\sin\beta - \cos\gamma\sin\alpha\cos\beta \\ -\cos\alpha\sin\beta & \sin\alpha & \cos\alpha\cos\beta \end{pmatrix} \quad (3.24)
$$

where the angles α, β and γ were given by

$$\alpha = \sin^{-1}\left({}_{D}^{P}A_{3,2}\right); \tag{3.25}$$

$$\text{if } \alpha < \frac{\pi}{2}\ ;\ \beta = \tan^{-1}\left(-{}_{D}^{P}A_{1,2},\ {}_{D}^{P}A_{2,2}\right)\ \ \&\ \ \gamma = \tan^{-1}\left(-{}_{D}^{P}A_{3,1},\ {}_{D}^{P}A_{3,3}\right) \tag{3.26}$$

$$\text{if } \alpha > -\frac{\pi}{2};\ \beta = 0\quad \&\quad \gamma = -\tan^{-1}\left(-{}_{D}^{P}A_{1,3},\ {}_{D}^{P}A_{1,1}\right) \tag{3.27}$$

These equations were extracted from an algorithmic notation in Matlab 7.0.4.

3.1.5 Determination of Foot and Tibia Angular Velocities

The angular velocity of foot and tibia was calculated by the angular velocity vector of the local coordinate system with respect to the global coordinate system, represented in the local coordinate system (${}^{L}\tilde{\boldsymbol{\omega}}_{L}^{G}$). This vector is given in the tilde notation (~) indicating a skew-symmetric matrix. It is the product of rotation matrix (${}_{G}^{L}\mathbf{A}$) and its time derivative (${}_{G}^{R}\dot{\mathbf{A}}$).

$${}^{L}\tilde{\boldsymbol{\omega}}_{L}^{G} = {}_{G}^{L}\mathbf{A}\,{}_{G}^{R}\dot{\mathbf{A}} \tag{3.28}$$

where

$${}^{L}\tilde{\boldsymbol{\omega}}_{L}^{G} = \begin{pmatrix} {}^{L}\tilde{\boldsymbol{\omega}}_{xL}^{G} \\ {}^{L}\tilde{\boldsymbol{\omega}}_{yL}^{G} \\ {}^{L}\tilde{\boldsymbol{\omega}}_{zL}^{G} \end{pmatrix} \tag{3.29}$$

Because of the experimental inaccuracies and numerical differentiation, the product ${}_{G}^{L}\mathbf{A}\,{}_{G}^{R}\dot{\mathbf{A}}$ was not strictly a skew-symmetric matrix. Therefore, each component of ${}^{L}\tilde{\boldsymbol{\omega}}_{L}^{G}$ was estimated by:

$$^{L}\tilde{\boldsymbol{\omega}}_{L}^{G}=\frac{\tilde{\boldsymbol{\omega}}_{i,j}-\tilde{\boldsymbol{\omega}}_{j,i}}{2} \text{ with } \{i,j\}=\begin{Bmatrix}3,2\\1,3\\2,1\end{Bmatrix} \tag{3.30}$$

These velocities were need in the third paper (Chapter 6) where the objective was the associations of peak knee moment with the rearfoot and tibial rotations during running.

3.2 Kinetic Analysis of the Ankle and Knee

The Newton-Euler inverse dynamics method was used to calculate the ankle and knee moments during running. With this approach, measured kinematic data were combined with the estimation of segmental inertial properties and ground reaction force data to estimate the resultant acting forces and moments at the ankle and knee in a model of the subjects' lower-limb. This information was required in the third paper (Chapter 6).

3.2.1 Forces and Moments at the Ankle and Knee

The foot and tibia were modeled as rigid bodies. In this model, the rearfoot motion was expressed relative to the tibia to represent the combined subtalar and talocrural joints motions. The mass, centre of mass location and moment of inertia about the principal axes were calculated for each segment using anthropometric tables proposed by De Leva (1996). The local coordinate system of tibia was described in the section 3.1.2. Using Wu et al. (2002), the origin of the local coordinate system of the foot was estimated at its centre located at marker $\mathbf{M}_6$. The positive medial axis of the foot, X_F was defined by

$$\mathbf{X}_F = \mathbf{M}_{14} - \mathbf{M}_6 \tag{3.31}$$

An interim z axis was determined from the line joining the medial calcaneus to the posterior calcaneus as.

$$z = \mathbf{M}_7 - \mathbf{M}_6 \tag{3.32}$$

A vector perpendicular to the $\mathbf{X}_F$ z plane was determined by the cross product such as

$$\mathbf{Y}_F = z \times \mathbf{X}_F \tag{3.33}$$

Then a vector perpendicular to the $\mathbf{Y}_F \mathbf{X}_F$ plane was calculated by

$$\mathbf{Z}_F = \mathbf{X}_F \times \mathbf{Y}_F \tag{3.34}$$

Unit vectors (i, j, k) for each axis were obtained by dividing them by their respective norm then three-dimensional joint angles were obtained by the method proposed by Grood and Suntay (1983) as described in section 3.1.4.

3.2.2 Newton-Euler Inverse Dynamics Method

Newton's equation states that the sum of all forces acting on a rigid body (i) is equal to its mass times its acceleration.

$$m\,{}^{G}\ddot{\mathbf{r}}_C = \sum_i {}^{G}\mathbf{F}_i \tag{3.35}$$

where

m = mass of the body

${}^{G}\ddot{\mathbf{r}}_C$ = the acceleration of the centre of mass

$\sum_i {}^G\mathbf{F}_i$ = sum of all the external forces

Euler's equation expresses the moment sustained by a body at its center of mass (${}^G M_C$) as

$${}^G M_C = {}^L\mathbf{J}_C \; {}^L\dot{\boldsymbol{\omega}}_L^G + {}^L\tilde{\boldsymbol{\omega}}_L^G \left({}^L\mathbf{J}_G \; {}^L\boldsymbol{\omega}_L^G\right) \qquad (3.36)$$

where

${}^L\mathbf{J}_C$ = segment moments of inertia matrix about the centre of mass

${}^L\dot{\boldsymbol{\omega}}_L^G$ = angular acceleration of the local coordinate system with respect to the global coordinate system, represented in the local coordinate system

$\tilde{\boldsymbol{\omega}}$ = skew matrix of the angular velocity

${}^L\mathbf{J}_G$ = segment moments of inertia matrix about global coordinate system

${}^L\boldsymbol{\omega}_L^G$ = angular velocity of the local coordinate system with respect to the global coordinate system, represented in the local coordinate system.

Three-dimensional inverse dynamics method was used to calculate the joint forces and moments acting at the ankle and knee. The joint reaction forces were

$$\mathbf{F}_{T\to F} = m_F(\ddot{\mathbf{r}}_F - \mathbf{g}) - \mathbf{F}_{GRF} \qquad (3.37)$$

$$\mathbf{F}_{Fe\to T} = m_T(\ddot{\mathbf{r}}_T - \mathbf{g}) - \mathbf{F}_{T\to F} \qquad (3.38)$$

where

$\mathbf{F}_{T\to F}$ and $\mathbf{F}_{Fe\to T}$ = the vector describing the force from tibia to foot and femur to tibia, respectively.

m_F and m_T = the mass of foot and tibia

$\ddot{\mathbf{r}}_F$, $\ddot{\mathbf{r}}_T$ = the linear acceleration vector of the centre of mass of foot and tibia,

$\mathbf{F}_{GRF}$ = the ground reaction force vector

The moments were calculated from distal to proximal segment. The ankle and knee moments were resolved from equations 3.36 to 3.38.

$$^{G}\mathbf{T}_{T\rightarrow F} = {}^{G}\dot{\mathbf{H}}_F + \tilde{\mathbf{d}}_F\, m_F \ddot{\mathbf{r}}_F - \left[(\tilde{\mathbf{d}}_F\, \mathbf{W}_F) + \mathbf{T}_Z + \tilde{\mathbf{l}}_F\, \mathbf{F}_{GRF}\right] \qquad (3.39)$$

$$^{G}\mathbf{T}_{Fe\rightarrow T} = {}^{G}\dot{\mathbf{H}}_T + \tilde{\mathbf{d}}_T\, m_T \ddot{\mathbf{r}}_T - \tilde{\mathbf{d}}_T\, \mathbf{W}_T + {}^{G}\mathbf{T}_{T\rightarrow F} + \tilde{\mathbf{l}}_T\, \mathbf{F}_{T\rightarrow F} \qquad (3.40)$$

where

$^{G}\mathbf{T}_{T\rightarrow F}$ and $^{G}\mathbf{T}_{Fe\rightarrow T}$ = moment vectors from tibia to foot and from femur to tibia, respectively.

$^{G}\dot{\mathbf{H}}_F$ = rate of change of foot angular momentum

$^{G}\dot{\mathbf{H}}_T$ = rate of change of tibial angular momentum

$\tilde{\mathbf{d}}_F$ = skew matrix of the vector position from ankle joint centre to foot centre of mass

$\tilde{\mathbf{d}}_T$ = skew matrix of the vector position from knee joint centre to tibia centre of mass

$\mathbf{W}_i$ = vector of gravitational terms

$\mathbf{T}_Z$ = free moment of force applied on the platform at the center of pressure

$\tilde{\mathbf{l}}_F$ = the skew matrix of the vector position from ankle joint centre to centre of pressure

$\tilde{\mathbf{l}}_T$ = the skew matrix of the vector position from ankle joint centre to knee joint centre

$\mathbf{F}_{GRF}$ = ground reaction force vector

$\mathbf{F}_{T\rightarrow F}$ = vector describing force from tibia to foot

Figure 3.4 illustrates the components of the resultant moment in the joint coordinate system. For the ankle, flexion/extension moment was about the medial/lateral axis (Z-axis) which is fixed in the foot. Internal/external rotation moment is about the proximal/distal axis (Y-axis) which is fixed in the tibia, and abductor/adductor moment is about the third axis which is mutually perpendicular to the other two axes (X-axis).

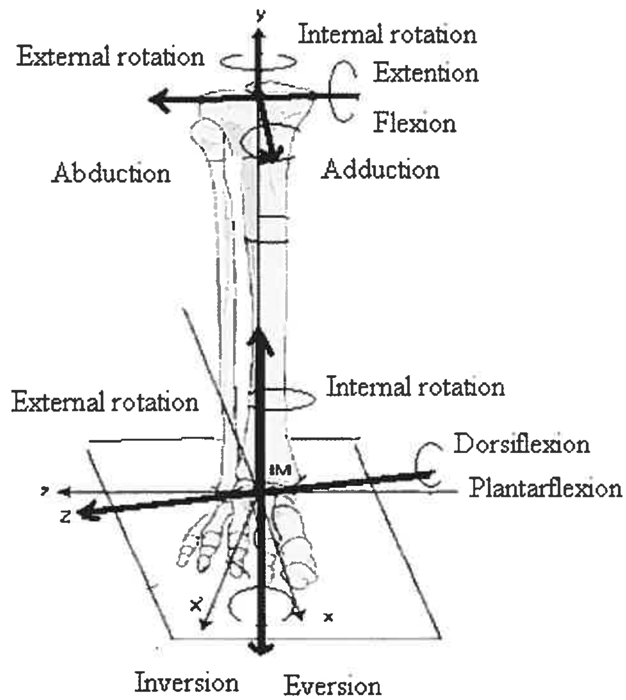

Figure 3.4 The components of the resultant moment in the joint coordinate system at the ankle and knee.

The error in the ankle joint complex caused by the skin movement artifacts was previously determined by Reinschmidt et al. (1997). They concluded that skin and shod markers gave a relatively good estimate of the actual tibiocalcaneal kinematic. In our study, three-dimensional coordinates filtered at 8 Hz with a low-pass zero phase shift fourth-order Butterworth filter. Since the coupling motion is subjective to the cut-off frequency (Hamill et al., 1999 & Stergiou et al. 2001), the frequency cut-off was determined based on residual analysis described by Winter (1990). All the kinematic parameters were calculated using a set of programs written in Matlab 7.0.4.

In summary, the goal of second paper (Chapter 5) was to determine the forefoot and rearfoot coupling motion patterns and their effect on the amount of tibial rotation in running. The aim of the third paper given Chapter 6 was to test if there is a relationship between foot kinematics and peak knee moment since the joint moment was proposed as an indirect measure of joint loading (Hunt et al., 2006). The model proposed by Kidder et al. (1996) was developed adding virtual markers during the shod running condition. They enabled the tracking of hidden markers at the forefoot and tibia during shod running. Furthermore, in the inverse dynamic approach, measured kinematic data are combined with estimated segmental inertial properties and ground reaction force data to estimate the resultant moment and force acting at each joint in the model. In the next chapter, angle variability of subtalar joint/ankle and their more proximal joints and segments will be assessed during single-limb stance.

Chapter 4

4. MANUSCRIPT 1

Title: **Effect of foot wedge positions on lower-limb joints, pelvis and trunk angle variability during single-limb stance**

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Journal: **The Foot (16); 208-213, 2006**

Key Words: **Orthotic devices, joints, kinematics, posture.**

ABSTRACT

Background: Wedge posted foot orthotics could prevent abnormal compensatory motions in the proximal joints by aligning the subtalar joint and ankle. However, the effect of wedge positions on the compensations of the subtalar joint and its proximal joints and segments in maintaining posture are not well understood.

Objective: To test the effect of four wedge positions on compensatory actions of the subtalar joint, ankle, knee, hip, pelvis and trunk by determining their angle variability during single-limb standing posture.

Method: Fourteen healthy male were tested in single-limb stance during 64 seconds. A wooden wedge with an inclination of 4,6° was placed under the anterior, posterior, lateral and medial sides of the dominant foot. A no wedge barefoot condition was also tested. Angle variability was measured by determining the root mean square (RMS) deviation value for each joint angle.

Results: The frontal plane angle variability for the subtalar joint was about 6 times greater for the medial and posterior wedge compared to the no wedge condition. For the anterior and posterior wedges, angle variability of the ankle and hip in the sagittal plane and pelvis and trunk in the transverse plane was about 2 to 3 times higher by comparison to the no wedge condition.

Conclusion: Wedge positions may affect differently the angular variability of the subtalar joint and its proximal joints and segments in their respective planes of movement. Similar patterns of changes in angle variability were found in the joints and segments which have the same plane of movement during single-limb standing posture.

INTRODUCTION

Foot orthoses combined with postings acting as wedges have widely been used to maintain the foot in a normal position and align it with the leg and thigh. A wedge could prevent abnormal compensatory motions in the proximal joints by aligning the subtalar joint and ankle. Abnormal compensation is the motion in which lower extremities and proximal segments adjust their position when the foot joints are misaligned (Nicolopoulos et al., 2000). For example, excessive eversion at the subtalar joint could produce abnormal compensatory motions by increasing the angular variability of the knee, hip, pelvis and trunk (Nicolopoulos et al., 2000; Johanson et al., 1994). A medial wedge designed to control the subtalar joint eversion in the frontal plane could correct abnormal compensatory internal rotation at the tibia, knee and proximal segments (Donatelli et al., 1988; Shaw, 1975). A forefoot wedge is also used to normalize the position of the forefoot relative to the rear foot, although a rear-foot wedge is thought to have a more direct effect than a forefoot wedge on subtalar joint motion in the frontal plane (Johanson et al., 1994). In the study of the effect of different wedge posted orthotics, the measurement of postural sway has been realized by means of the trajectory of the centre of pressure (Baier & Hopf, 1998; Hertel et al., 2001; Guskiewicz & Perrin, 1992). However, it still remains unknown if the foot joints and their proximal segments have normal interactions and compensations in maintaining posture.

Postural control is accomplished through joint interactions and compensations in different planes during standing balance. During quiet stance on a fixed support surface, the ankle and hip (Horak & Nashner, 1986; Nashner & Woollacott, 1979) as well as the knee (Nashner & Woollacott, 1979) control posture in the sagittal plane. Narrow stance width such as single-limb stance increases frontal plane motion at the subtalar joint and hip (Hoogvliet et al., 1997). It is generally believed that in a most challenging position when the subtalar joint and ankle cannot correct standing posture, the human body reacts as a multi-segmental chain with large joint angle variability at the knee, hip and trunk in different planes of movement (Nashner & McCollum, 1985; Tropp & Odenrick, 1998; Winter et al., 1996). Riemann et al. (2003) concluded that the reduction of the base of support in single-limb stance on a foam surface, increases joint angular motion at the ankle, hip and knee in order to control standing balance. Thus, single-limb stance could elicit proximal joints compensation actions characterized by an increase in joint angle variability.

The effect of wedge positions on the subtalar joint and proximal joints and segments acting in different planes are not well understood. We hypothesized that wedges located under the foot will affect equally the plane of movement joint angle variability at the lower limb joints, pelvis and trunk. Additionally, changes in angle variability will occur equally at these joints and segments to maintain posture during single-limb stance. The objective of this study is to test the effect of four wedges positioned under the medial (MW), lateral (LW), anterior (AW) and posterior (PW) sides of the foot on the subtalar joint, ankle, knee, hip, pelvis and trunk compensation actions by determining their angle variability during single-limb standing posture in a normal population.

METHOD

Fourteen able-bodied men having an average age of 31,1±6,0 years, weight of 84,7 ±11,3 kg and height of 179,4± 7,6 cm voluntarily participated in this study. Subjects would be included if they had no previous recent history of musculoskeletal or neurological ailments that could affect their postural balance. Individuals who used foot orthoses; had surgery at the lower limbs, or have hearing or visual impairments were excluded. The experimentation procedures were explained to all participants and those who volunteered signed the informed consent form approved by the Hospital Ethics Committee.

First a capture volume ($0,5\times 0,5 \times 2,0$ m^3) covering the whole body in standing position was used to calibrate a set of five cameras. The cameras were located around the capture volume in an ''umbrella'' configuration. Seventeen spherical markers (25mm in diameter) with double-sided adhesive tape were attached to the following identified land-markers:

left shoulder, right shoulder, iliac crest, left anterior superior iliac spine, right anterior superior iliac spine, superior lateral thigh, inferior lateral thigh, superior anterior thigh, inferior anterior thigh, superior lateral shank, inferior lateral shank, superior posterior shank, inferior posterior shank, superior heel, inferior heel; lateral mallelous and 5th metatarsal. Additionally ground right and left markers aligned with medial and lateral axis were used as fixed markers to measure the transverse plane angle changes.

Each subject was asked to stand inside the capture volume on his dominant single limb. The preferable foot for kicking a ball was chosen as a dominant limb. The contra lateral limb was maintained with the knee flexed at about 90 degrees beside the dominant limb. For all the subjects, the dominant limb was on the right side. The subjects were also instructed to fold their arm across the chest and to look at a visual marker located in front of them at shoulder level at a 2m distance (Tropp & Odenrick, 1998) (Figure 4.1).

A wooden wedge of 70mm×50mm dimension, with an elevation of 4mm and an inclination of 4,6 degrees was placed under the dominant foot (Figure 4.2). The orientation of the slope of the wedge was oriented towards the anterior, posterior, medial and lateral sides of the foot. For each of position, a series of three acquisitions were performed, each trial lasting for 64 seconds. Three trials of 64 seconds were also collected for the no wedge condition.

Three-dimensional spatial positions of the markers were recorded using the five cameras and a motion analysis system (Motion Analysis Corporation, Santa Rosa, CA, USA) at 60 Hz. To avoid beginning and ending effects during the single-limb stance, the middle 40 seconds of each trial was analyzed. Seven planar angles in three planes of movement were measured using the dot product. The inversion and eversion of the subtalar joint were defined as medial and lateral tilt of the heel relative to the shank in the frontal plane. Adduction and abduction motions in the hip were defined as medial and lateral tilt of thigh relative to the pelvis. In the sagittal plane, flexion and extension of the hip were defined as anterior-posterior motions of the line between the right shoulder, the iliac crest and superior and inferior thigh markers. For the knee, flexion and extension were defined as thigh motion relative to the shank in the sagittal plane. The ankle angle was defined as the shank sagittal motion relative to the foot. The trunk rotation was defined as the transverse plane motion of the shoulder relative to the fixed markers on the ground. The pelvis rotation was defined as the transverse plane motion of the pelvis relative to the fixed markers on the ground.

The dependant variable is the root mean square (RMS) deviation value across the three trials. The deviation values were measured from the average of angular displacements of the three trials for each wedge position from reference joint angles. The reference joint angle corresponded to the mean angular displacement value of the three trials taken in the no wedge condition for each joint angle.

To determine if all joints and segments compensated equally to maintain posture and if the compensations were dependent on the wedge positions, a two-factor (wedge positions by joints) repeated-measures analysis of variance (ANOVA) was performed. Protected t test comparisons were used to determine significant differences for joint angle variability between no wedge condition and each wedge condition when overall F test was statistically significant. The significance level was set at 5%.

RESULTS

No interaction effect was observed. Significant main effect was found for the wedge position ($F_{4, 52}$=5,76, P=0,002). Figure 4.3 illustrates the frontal plane subtalar and hip angle variability. For all wedge positions, the subtalar joint showed significant differences when compared with the no wedge condition. The average of subtalar joint variability in the MW condition was about 6 times greater than the NW condition (P =0,004), while it was respectively 4,7 and 5,6 times greater in the AW (P =0,008) and PW positions (P=0,001). In the LW condition the average subtalar joint variability was 3,1 times greater (P = 0,007) than NW condition. The average of hip joint variability in the frontal plane was statically significant in the AW (P =0,033) and MW (P =0,015) positions when compared with the NW condition. Hip variability increased significantly and was about 2,5 times higher than the no wedge condition in the AW and MW positions.

Figure 4.4 illustrates the sagittal plane angle variability of the hip, knee and ankle. A statistically significant increase in the angle variability values was revealed for the ankle in all positions compared to the no wedge condition. The average RMS value in ankle was 3,5; 3,1; 3 and 2,3 times higher in the AW (P =0,027), PW (P =0,006), MW (P =0,046) and LW (P=0,005) positions than the NW condition, respectively. For the hip, these values were respectively about 2, 1,53 and 1,56 times greater in the PW (P =0,048), LW (P=0,029), and MW (P=0,41) positions than the NW condition. For the knee joint, the RMS value was significantly increased at the LW (P =0,004) condition only, it was about 1,8 times higher than the NW condition.

Figure 4.5 illustrates the RMS values for the trunk and pelvis measured in the transverse plane for the NW condition and wedge positions. Compared to the NW condition, the RMS values of the trunk were increased in the AW (P = 0,003) and PW (P =0,002) positions by 2,5, and in the LW (P =0,019) and MW (P =0,033) positions by 2,0. The pelvis showed a greater RMS than the trunk in the AW (P =0,011), LW (P =0,052) and MW (P=0,075), but smaller in the PW (P =0,030), although in the LW and MW positions were not significant. Generally both of the trunk and pelvis showed a similar trend in the AW, LW and MW wedge positions.

DISCUSSION

All four wedge positions increased the angle variability of the subtalar joint, ankle, knee, hip, pelvis and trunk in their plane of movement when compared with the no wedge condition. Similar augmentations of the angle variability were observed in the joints and segments which have the same plane of movement. The finding illustrates that each wedge position could target specific joints and planes of movement to maintain posture during single-limb stance.

The applied wedges in this experiment could change related mechanical and proprioceptive properties, resulting in a greater variability of different joints and segments. In term of mechanical changes, a wedge position could change the orientation of subtalar joint axis of motion. The area of the contacting surface between talus and calcaneus could become smaller, providing a smaller base of support for talus. In addition, the muscle tendons' orientations are changed. These phenomena plus the perturbation on the ankle proprioceptions could play a major role in the observed variability.

Comparisons between each wedge position and the no wedge condition show that the posterior and medial wedges highly increased the frontal plane angle variability in the subtalar joint and hip. The similar pattern of variability in the subtalar joint and hip among wedge positions also shows an in phase relationship for maintaining posture in the frontal plane of motion. The frontal plane motion at the subtalar joint is thought to be influenced by the orientation of the sagittal axis of subtalar joint from the plantar surface (DeLeo et al., 2004; Williams et al., 2001). The posterior wedge could reposition the calcaneus in plantar flexion and decrease the sagittal axis orientation of subtalar joint. Furthermore, the medial wedge changes the Achilles tendon orientation and consequently leads to higher variability in the frontal plane than the other wedge positions.

In the lateral wedge position, greater angle variability at the ankle, knee and hip in the sagittal plane could be due to limited motion at the subtalar joint in the frontal plane. This result supports the idea that small angular changes in eversion at the subtalar joint could cause internal rotation of the tibia, affecting the motion pattern at the knee and hip (Inman & Mann, 1978) as well as that of the ankle (Nashner & Woollacott, 1978). Similarities in the angle variability changes at the ankle and hip across the four wedge positions show their positive interactions in maintaining posture in the sagittal plane during single-limb stance. Indeed, significant increase in the angle variability of the knee in the sagittal plane only for the lateral wedge position could show greater compensatory actions at the knee and could be due to the limited frontal plane motion of the subtalar joint and hip.

Greater compensatory actions of the pelvis and trunk rotations for the anterior and posterior wedge positions could be related to high variability of the sagittal plane motion of the ankle and hip, as well as the frontal plane of the subtalar and hip in order to maintain posture. Pelvis rotation in the transverse plane could occur to minimize the displacement of the centre of mass. The angle variability of the trunk in the transverse plane could be related to its higher mass and moment of inertia, which can be associated with quick adjustments of posture necessary for stability (Riemann et al., 2003). Furthermore; the trunk provides a damping effect on the pelvis rotation and contributes to a smoother movement (Leroux et al., 2002).

The use of wedges in different positions causes greater angle variability of the subtalar joint frontal plane motion and its proximal joints and segments in their respective planes of movement by comparison to the no wedge condition. Fitzpatrick (1992) claimed that a multi-link structure function of the body increases stability through decreasing the large inertia that would be associated with a large segment. Furthermore, at each segment, passive damping can occur, thereby decreasing the need for sustained muscle activation. On the other hand, it is reported that the need for the number of muscle activation and postural strategy possibilities are minimized with increasing stability. The minimization of muscle activity can occur due to two major strategies: first, by stabilizing joints and second, by minimizing soft tissue vibration (Nigg et al., 1999). However, the contribution of high joint variability in enhancement of postural stability remains unknown during single-limb stance.

Generally, a given wedge position could either bring the subtalar and ankle joints in their normal position to improve postural stability or misalign the foot joints, by affecting their mechanical and properioceptive properties. The application of the appropriate wedged foot orthoses could prevent abnormal compensation motions by improving the proprioception information as well as the mechanical properties of the foot.

CONCLUSION

Medial and posterior foot wedge positions highly increased the frontal plane angle variability of the subtalar joint; whilst high angle variability of the ankle and hip in the sagittal plane, and the pelvis and trunk in the transverse plane were seen for the anterior and posterior wedge positions. Similar patterns of changes in angle variability were found in the joints and segments which have the same plane of movement.

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Figure 4.1 Anterior and posterior views of marker placement during barefoot single-limb stance.

Figure 4.2 Structure of wedge (lateral view).

Figure 4.3 RMS values and standard deviations of angle changes for the hip and subtalar joints in the frontal plane for all wedge positions [anterior wedge (AW), posterior wedge (PW), lateral wedge (LW), medial wedge (MW)]. A statistical difference is indicated by (*) in comparison to the no wedge condition (NW).

Figure 4.4 RMS values and standard deviations of angle changes in the hip, knee and ankle joints in the sagittal plane for all wedge positions. A statistical difference is indicated by (*) in comparison to the no wedge condition (NW).

Figure 4.5 RMS values and standard deviations of angle changes for the trunk and pelvis in the transverse plane for all wedge positions. A statistical difference is indicated by (*) in compare to the no wedge condition (NW).

Figure 4 1

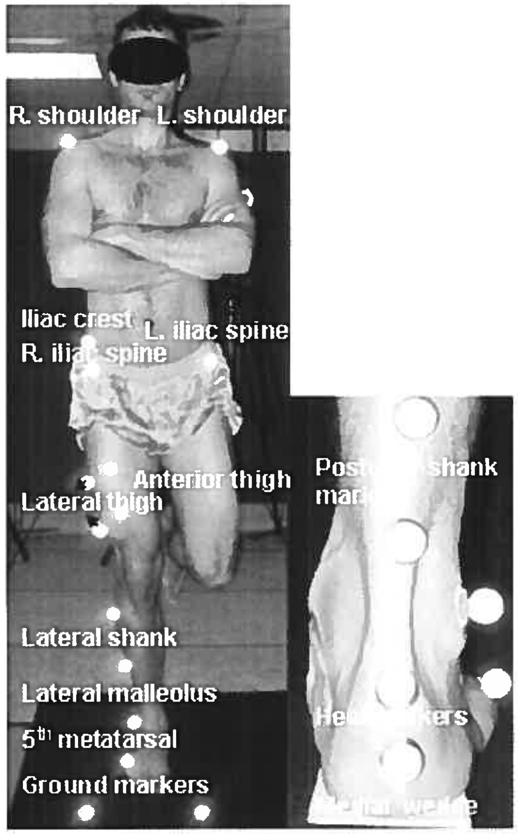

R. shoulder
L. shoulder
Iliac crest
L. iliac spine
R. iliac spine
Anterior thigh
Lateral thigh
Lateral shank
Lateral malleolus
5th metatarsal
Ground markers

Figure 4.2

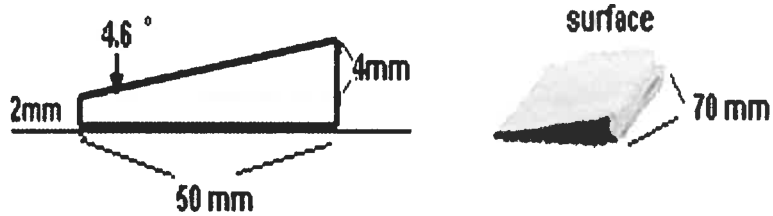


Figure 4.3

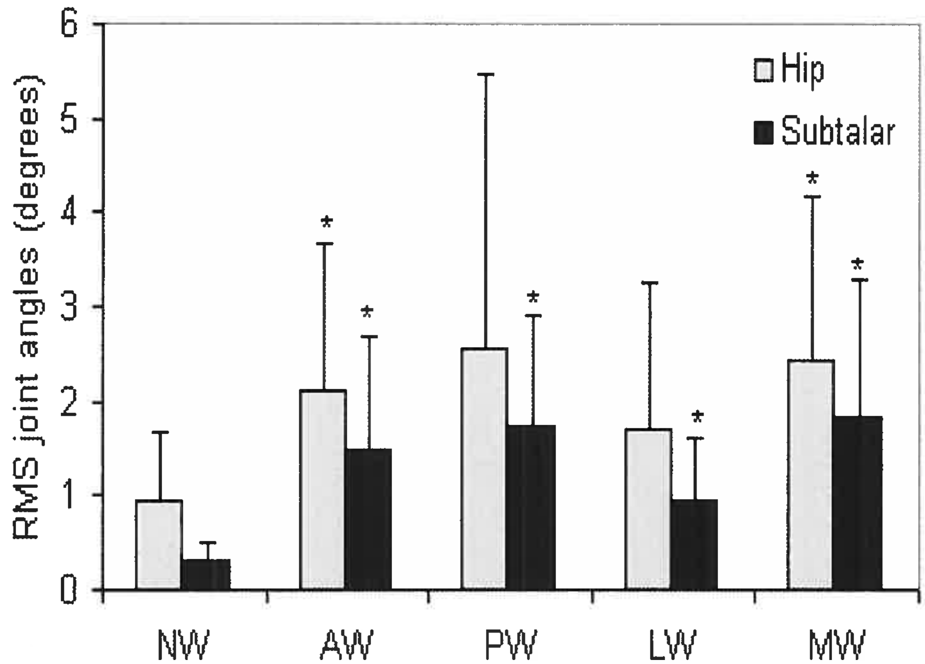


Figure 4.4

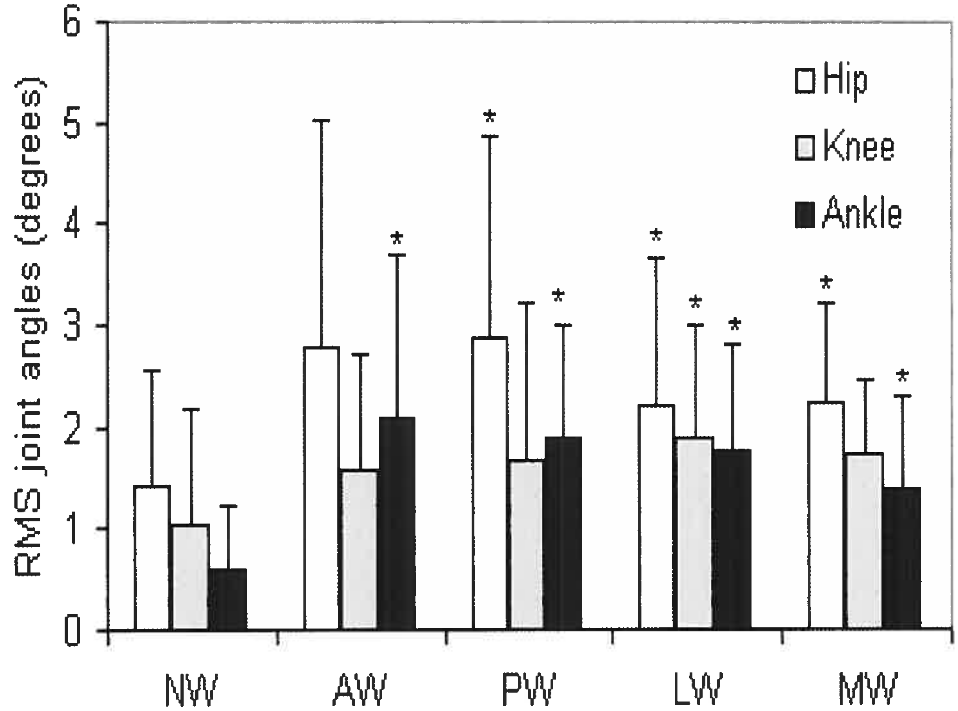
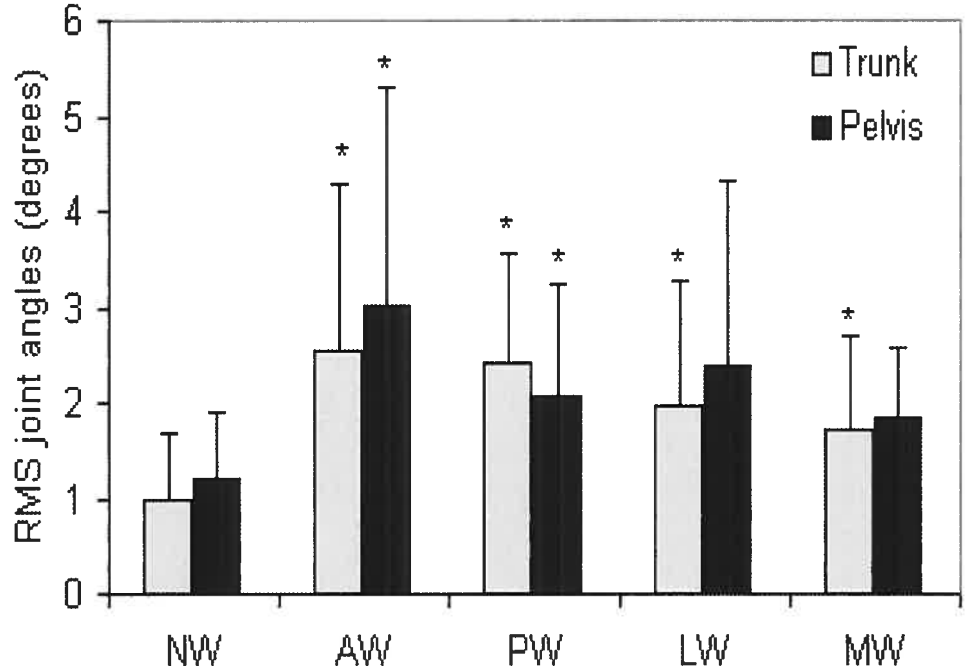

RMS joint angles (degrees)
0
1
2
3
4
5
6
Hip
Knee
Ankle
*
NW
AW
PW
LW
MW

Figure 4.5



Chapter 5

5. MANUSCRIPT 2

Title: **Forefoot-rearfoot coupling patterns and tibial internal rotation during stance phase of barefoot versus shod running**

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Keywords: **Forefoot; Rearfoot; Relative phase angle; Tibia internal rotation**

ABSTRACT

Background: Based on twisted plate and mitered hinge models of the foot and ankle, forefoot- rearfoot coupling motion patterns can contribute to the amount of tibial rotation. The present study determined the differences of forefoot-rearfoot coupling patterns as well as excessive excursion of tibial internal rotation in shod versus barefoot conditions during running.

Methods: Sixteen male subjects ran 10 times at 170 steps per minute under the barefoot and shod conditions. Forefoot-rearfoot coupling motions were assessed by measuring mean relative phase angle during five intervals of stance phase for the main effect of five time intervals and two conditions (ANOVA, $P<0,05$). Tibial internal rotation excursion was compared between the shod and barefoot conditions over the first 50% of stance phase using paired t test, ($P<0,05$).

Findings: Forefoot adduction/abduction and rearfoot eversion/inversion coupling motion patterns were significantly different between the conditions and among the intervals ($P<0,05$; effect size=0,47). The mean absolute relative angle was significantly modified to 37° in-phase relationship at the heel-strike of running with shoe wears. No significant differences were noted in the tibial internal rotation excursion between shod and barefoot conditions.

Interpretation: Significant variations in the forefoot adduction/abduction and rearfoot eversion/inversion coupling patterns could have little effect on the amount of tibial internal rotation excursion. Yet it remains to be determined whether changes in the frontal plane forefoot-rearfoot coupling patterns influence the tibia kinematics for

different shoe wears or foot orthotic interventions. The findings question the rational for the prophylactic use of forefoot posting in foot orthoses.

1. INTRODUCTION

Excessive tibial internal rotation coupling with rearfoot eversion during the first half stance phase of running was associated with patella-femoral pain syndrome, Achilles tendon pain and shin splint (Clement et al., 1981; Smart et al., 1980; Tiberio, 1987; Viitasalo et al., 1983). The amount of internal tibial rotation is proposed to be related to coupling motion patterns between the forefoot and rearfoot (Lundberg, 1989; Naster et al., 2002). A twisted plate model of the foot suggests that the forefoot produces counter motions with respect to the rearfoot segments during barefoot running (Hunt et al., 2001; Sarraffian, 1993). From heel-strike through foot-flat, the rearfoot is everted and the forefoot becomes flexible to absorb shock and adapt itself to irregularities in the ground floor surface (Nordin & Frankel, 2001). A cross-correlation between the rearfoot and forefoot motion indicated that rearfoot eversion/inversion was highly correlated to forefoot plantar/dorsiflexion ($r < -0{,}85$) and abduction/adduction ($r > 0{,}94$) with no phase shift during the stance phase of barefoot running (Pohl et al., 2006). Johanson et al. (1994) reported that a large forefoot inversion with respect to the rearfoot results in an abnormal gait pattern when resulting in compensatory subtalar joint pronation. Furthermore, using a mitered hinge model, rearfoot eversion in the frontal plane was found to be coupled with tibial internal rotation during gait (Pohl et al., 2006; Nigg et al., 1993). A high correlation value ($r = 0{,}99$) was reported between rearfoot eversion and tibial internal rotation during the first 50% stance phase of gait (Pohl et al., 2006; Nigg et al., 1993). Therefore, based on the twisted plate and mitered hinge models, the forefoot and rearfoot coupling motion patterns could contribute to the amount of tibial rotation.

In previous studies the rearfoot and tibia coupling motion was modelled as a single rigid segment because of technical difficulties associated with evaluating the forefoot motion in a shoe condition. Furthermore, *in vivo* studies on the forefoot motions, subjects were tested in barefoot condition to enable tracking of markers on the forefoot (Pohl et al., 2006; Hunt et al., 2001). Therefore, footwear effects on the three-dimensional forefoot motion coupling with the rearfoot frontal plane motion and their contributions to the tibial rotation remained unknown.

The use of forefoot posting in orthotic interventions to compensate excessive foot pronation is still misunderstood. Clinically, it is believed that abnormal foot pronation is associated with forefoot excessive motions with respect to the rearfoot (Johanson et al., 1994; Tillman et al., 2003). However, Johanson et al. (1994) indicated that posting in the rearfoot was more effective in controlling foot pronation than posting in the forefoot, even in the presence of a forefoot deformity. A better understanding of the forefoot and rearfoot coupling relationships and their contributions to the tibial rotation in asymptomatic feet will provide information of the importance of forefoot posting in the orthotic interventions in controlling tibial rotation.

A number of techniques have been used to examine coupling motion relationships between rearfoot and tibia during dynamic motions. Cross-correlations are based on the assumption that linear relationships exist between two adjacent segments. However, this technique is not useful in determining the degree of linkage between the segments that have a non-linear relationship (Sideway et al., 1995). Rearfoot eversion and tibial internal rotation (EV/TIR) excursion ratio is used to provide a measure of the relative motion between the rearfoot and tibia from heel-strike to the respective peaks around mid-stance (DeLeo et al., 2004). In the recent studies, the EV/TIR ratio varied between 0,65 in the normal shod (Stacoff et al., 2000) and 2,40 in the barefoot conditions (Pohl et al., 2006). These values suggest that the rearfoot is everted by 1° for every $1{,}54^{\circ}$ and $0{,}41^{\circ}$ tibial internal rotation in shod and barefoot conditions, respectively. In the present study, EV/TIR excursion ratio will be used to determine if the tibia has a relatively greater motion with respect to the rearfoot (Nawoczenski et al., 1995; Nigg et al., 1993; Williams et al., 2001). For example, runners with lower EV/TIR ratios display relatively more tibial rotation with respect to the rearfoot rotation and increasing the risk for knee related injuries (McClay & Manal, 1997; Williams et al., 2001). A continuous relative phase angle technique (CRP) was also proposed to describe the coupling motion relationships of two adjacent segments throughout the stance phase (Hamill et al., 1999). This technique indicates the amount of in-phase or out-of-phase relationship between two adjacent segments. Hamill et al. (1999) reported that the relationship between the rearfoot and tibia was more out-of-phase in the strike phase than the rest of stance in a group of healthy runners. However, there is no information regarding to coupling motion patterns of the forefoot and rearfoot during shod running in the literature. Thus, relative phase angle technique will be used to provide quantitative information on the forefoot-rearfoot coupling motion patterns throughout the stance phase of barefoot running versus running with sandals.

With respect to the following three assumptions, sandals were used as footwear in the present study. Firstly, the sandals' adjustable straps and the bottom midsole design enable greater changes in the forefoot and rearfoot coupling motion patterns than running shoe. Secondly, the sandal allows tracking of the rearfoot and forefoot surface markers during running trials. Finally, sandals are often used to evaluate the effects of foot orthoses on the rearfoot and tibia coupling motions (Branthwaite et al., 2004; Nawoczenski et al., 1995). However, the confounding effects of the sandal on the outcome measures of these coupling motions were unknown in the literature.

In current study, we hypothesized that tibial internal rotation is increased when the forefoot-rearfoot coupling patterns are modified to a more in-phase relationship with shoewear during the stance phase of running. The purposes were: i) to compare the excursion of tibial internal rotation and rearfoot eversion from heel-strike to peak value during the stance phase of running in barefoot versus shod conditions, ii) to determine differences in mean relative phase angle of the forefoot eversion/inversion and rearfoot eversion/inversion ($FF_{ev/in}$-$RF_{ev/in}$), forefoot dorsi/plantarflexion and rearfoot eversion/inversion ($FF_{d/p}$-$RF_{ev/in}$), forefoot adduction/abduction and rearfoot eversion/inversion ($FF_{ad/ab}$- $RF_{ev/in}$) during the stance phase of barefoot versus shod running.

2. METHOD

Sixteen able-bodied healthy men having an average age of 28,2 (SD 5,2 years), weight of 82,3 (SD 10,4 kg) and height of 179,0 (SD 5,4 cm) volunteered to participate to this study. The experimentation procedures were explained to all participants and those who volunteered signed an informed consent form approved by the Hospital Ethics Committee.

2.1 Experimental Set-up

Six cameras (Motion Analysis Corporation, Santa Rosa, CA, USA) were arranged along two arcs on the left and right sides of a force plate (960 Hz, AMTI, Watertown, MA, USA) placed in the middle of a 10 m runway. The capture volume ($0{,}5\times 0{,}5 \times 0{,}75$ m^3) covered the lower limb motion. Video data were collected using the EVa 4.2 software (Motion Analysis Corporation, Santa Rosa, CA, USA) at 60 Hz. Accuracy of the spatial reconstruction was assessed by means of an artificial foot (prosthesis) where markers corresponded to the forefoot and rearfoot. The average angular standard deviation was found about 1,5° in fast motions.

Forefoot, rearfoot and tibia were modelled as three rigid segments. The motion of the forefoot with respect to the rearfoot was defined using Kidder's et al. (1996) model whereas the rearfoot motion with respect to the tibia complied with the

Joint Coordinate System recommendation (Wu et al., 2002). These joint representations accounted for the functional anatomy of the foot and allowed the greater kinematic analysis than previous simpler models.

Thirteen reflective skin markers (16 mm diameter) were attached to the right foot and shank. Of these, ten markers were fixed on predefined anatomical landmarkers to define the forefoot, rearfoot and tibia coordinate systems as shown in (Figure 5.1 a,b) and describe as:

-Forefoot: medial side of the fifth metatarsal head (M5MH), lateral side of the first metatarsal base (L1MB) and head (L1MH).

-Rearfoot: posterior calcaneus (POSTC), medial calcaneus (MEDC), lateral calcaneus (LATC).

-Tibia: tibial tubercle (TIBT), head of the fibula (HFIB), medial malleolus (MEDM), lateral malleolus (LATM).

Three technical markers were also placed on the anterior middle aspect of the tibia (ANTT), fifth metatarsal base (M5MB) and middle part of between the second and third metatarsals (M23M). To avoid from marker dropout, skin movement artifact and hidden markers during running trials, the markers at the tibial tubercle, medial malleolus, medial side of fifth metatarsal head, lateral side of the first metatarsal head were removed and calculated as virtual markers after recording a barefoot neutral standing position. The technical markers were used to define the coordination of the virtual markers during running trials.

Three-dimensional joint rotations were calculated using method of Grood and Suntay (1983). The sequence of rotations was first plantar/dorsiflexion about a fixed media-lateral axis of the proximal segment, abduction/adduction about the floating axis, then inversion/eversion about the anterior-posterior axis of the distal segment. The tibial internal/external rotation corresponded to the rearfoot abduction/adduction motion. All the kinematic parameters were calculated using a set of programs written in Matlab 7.0.4 from the three-dimensional coordinates previously filtered at 8 Hz with a low-pass zero phase shift fourth-order Butterworth filter.

2.2 Testing Procedure

A barefoot standing trial was recorded to define the coordinate systems of the tibia, rearfoot, and forefoot for 4 seconds. The standing trial allowed the calculation of virtual markers location with respect to technical system of coordinates. For recording a neutral position, thirteen markers were attached to the right forefoot, rearfoot and tibia. Subjects were instructed to stand with straight knee and ankle in neutral position and feet aligned parallel to the force platform representing the laboratory coordinate system. Then four markers (TIBT, MEDM, L1MH and M5MH) were removed and the subjects were given sandals to practice running along the runway. New sandals were selected to fit the subject's foot size. The three straps of the sandal surrounded the foot at the calcaneus on the posterior side, at the tarsal-metatarsal joints and at the metatarsal-phalangeal joints along the frontal side of foot (Figure 5.1.c). Each sandal was designed with the same midsole material for shock absorption. It had height of 30 mm, 25 mm and 20 mm in the heel, middle and forefoot segments, respectively. Ten running trials were performed in the barefoot and shod conditions in a block random order. In each experimental condition, the subject ran at a controlled cadence of 170 steps per minutes. A successful trial was defined as one where the subject's right foot landed on the force plate during running.

2.3 Data Analysis

The dependent variables were: excursion of rearfoot eversion, excursion of tibial internal rotation, over the time period from heel-strike to the maximum value around mid-stance. Excursions were calculated by determining the difference between the maximum value during first 50% of stance phase and the value at heel-strike.

For statistically analyzing the coupling motion patterns of the forefoot and rearfoot, the stance phase was divided into five intervals determined from vertical force and loading rate. The first two intervals were taken at heel-strike (0%) and foot-flat (5% to 25% of stance) since significant differences of loading rate and vertical force were observed during the first 25% of stance phase. Furthermore, the variation of the loading rate and vertical force at the heel-strike phase was higher than the other data points within the first 25% of stance phase. Because force-plate data followed approximately similar trends in the remaining of the stance phase, the last three intervals were taken at heel-rise (25% to 50% of stance), push-off (50% to 75% of stance phase) and toe-off (75% to 95% of stance phase). Since there was no velocity value for the end of the stance phase, the mean relative phase angle of the toe-off interval was calculated from 75% to 95% of the stance phase.

The phase angle profile for the forefoot and rearfoot were generated from the average of a point by point across the all trials. Phase angle were normalized and calculated as described in the Hamill et al. (1999) study. Relative angle was defined as difference between the normalized phase angles of the rearfoot as the proximal segment and the forefoot as the distal segment during the stance phase of running. The mean absolute relative phase angles for each interval of the stance phase were calculated over time according to the method outlined by Stergiou et al. (2001).

2.4 Statistical Analysis

Paired t tests were used to compare the tibial internal rotation, rearfoot eversion excursions and EV/TIR ratios. A two-factor repeated measures ANOVA (2 conditions × 5 intervals) and post-hoc analysis using Bonferroni test were performed to identify differences in the mean absolute relative phase of forefoot-rearfoot between barefoot and shod conditions and among the five time intervals of stance phase (α=0,05).

3. RESULTS

Figure 5.2 indicates that the rearfoot eversion and tibial internal rotation occurred from heel-strike to about mid-stance then the rearfoot inverts and the tibia rotates externally from mid-stance to toe-off. A similar pattern of rearfoot and tibial rotation coupling motions was observed from heel-strike to toe-off in the barefoot and shod running conditions. The steeper slope of the mean curve indicates a higher rearfoot frontal plane motion than the tibial rotation during stance phase. The EV/TR excursion ratio in the barefoot and shod running was 1,80 and 2,24, respectively (P>0,05). This finding shows that the rearfoot is everted by 1° for every 0,55° and 0,44° tibial internal rotation in the barefoot and shod conditions during the stance phase of running, respectively.

Table 5.1 illustrates the mean and standard deviation of the tibial internal rotation and rearfoot eversion in barefoot and shod conditions. Rearfoot eversion excursion increased by 2% while tibial internal rotation excursion decreased by 1,6% in the shod condition when compared to barefoot condition. However, these changes were minimal and statistically insignificant (P>0,05).

Table 5.2 and 5.3 present the mean absolute relative phase angles of $FF_{ev/in}$-$RF_{ev/in}$ and $FF_{d/p}$-$RF_{ev/in}$ for each interval of the stance phase in the shod and barefoot conditions. No statistical differences were observed in the relative phase angles of $FF_{ev/in}$-$RF_{ev/in}$ and $FF_{d/p}$-$RF_{ev/in}$ between shod and barefoot conditions for any intervals of the stance phase (P>0,05).

Statistical analysis for the absolute relative phase angle of $FF_{ad/ab}$- $RF_{ev/in}$ showed an interaction effect of the intervals of the stance phase and conditions (P<0,01). Effect size estimation indicated that the intervals of the stance phase contribute to 47% of the total variance (more important factor). In the barefoot condition, the relative phase angle was by 50° and 53° higher and in out-of-phase in the heel-strike compared to the foot-flat (P<0,01) and heel-rise (P<0,01), respectively. Furthermore, a statistically higher (by -22°) in-phase relationship was observed in the heel-rise phase than toe-off (P<0,05).Whereas in the shod condition, significant differences among the intervals were observed between foot-flat (20,7°) and toe-off (41,0°), (P<0,05) as well as between heel-rise (16,0°) and push-off periods (42,1°), (P<0,05), (Table 5.4). These findings indicate that $FF_{ad/ab}$- $RF_{ev/in}$ coupling motion have a more out-of-phase relationship during heel-strike compared to latter intervals during barefoot running. Contrary, a higher out-of-phase relationship (by 26°) of the $FF_{ad/ab}$- $RF_{ev/in}$ coupling motion was observed in push-off and toe-off compared to heel-rise during the shod running. Furthermore, a statistical difference between the shod and barefoot conditions was observed in heel-strike (P=0,01). This difference was higher by 37° in the out-of-phase relationship in the barefoot compared to the shod condition. This finding shows that the out-of-phase relationship of $FF_{ad/ab}$- $RF_{ev/in}$ at heel-strike in the barefoot condition is modified to a more in-phase relationship with the sandal.

4. DISCUSSION

The first purpose of this study was to compare the excursion of rearfoot eversion and tibial internal rotation from heel-strike to the peak value during the first half stance phase in barefoot versus shod running. The findings showed an insignificant change in the rearfoot eversion excursion and tibial internal rotation by using the sandals when compared to the barefoot running. This is consistent with the results obtained by comparing normal shod and barefoot running via the direct measurement of the markers mounted on bones (Stacoff et al., 2000). However, Stacoff et al. (1991)

showed differences in the rearfoot and tibia coupling patterns using skin and shoe mounted markers between barefoot and normal shod running. The functional frontal plane subtalar joint motion and tibial rotation during barefoot running was reported to vary from 8° to 15° and from 3° to 6°, respectively (McClay & Manal, 1997; Pohl et al., 2006; Stacoff et al., 2000). In the present study, the average range of the frontal plane rearfoot motion was $11{,}4^{\circ}$ (SD 4,3) in barefoot and $10{,}9^{\circ}$ (SD 4,9) in the shod condition. Furthermore, the average range of tibial rotation was $5{,}2^{\circ}$ (SD 2,4) and $5{,}3^{\circ}$ (SD 2,9) in the barefoot and shod conditions in healthy runners, respectively. This finding shows that the rearfoot eversion and tibial internal rotation excursions varied in the range reported in the previous studies during the selected running speed. This variation in the mean values may due to different experimental protocols and foot joint models utilized. It is suggested that the effect of normal shod on the rearfoot and tibia coupling motion could be observed when different type of feet are selected and tested during higher running speeds or when cutting movements are performed (Stacoff et al., 2000).

Second purpose of this study was to determine differences in the mean relative phase angle of the forefoot and rearfoot during the five intervals of stance phase between barefoot and shod running. No statistical difference was noted in the mean relative phase relationship of $FF_{ev/in}$-$RF_{ev/in}$ during the five intervals of stance phase in barefoot versus shod running. Cornwall and McPoil (2002) showed that forefoot inversion was coupled with rearfoot eversion during the heel-strike phase period of walking; in contrast, Pohl et al. (2006) found that the forefoot was everted with respect to the rearfoot in the heel-strike period of barefoot running. These findings suggest that the forefoot frontal plane motion with respect to rearfoot could vary during different gait patterns (running versus walking). Shod running had no significant effect on the mean relative angle of $FF_{d/p}$-$RF_{ev/in}$ during the five intervals of stance phase. Hunt et al. (2001) and Lundberg et al. (1989) showed that the forefoot sagittal and frontal plane motion patterns were linked to the collapse of the medial longitudinal arch. They believed that the talonavicular joint could contribute to the forefoot sagittal and frontal plane motions and arch behaviors. Generally, the sandals could not change $FF_{ev/in}$-$RF_{ev/in}$ and $FF_{d/p}$-$RF_{ev/in}$ coupling patterns as well as tibial internal rotation as compared to barefoot running. However, regarding to the twisted plate and mitered hinge models, significant changes of $FF_{ev/in}$-$RF_{ev/in}$ could likely affect on the amount tibial rotations during running. This needs to further investigations with different footwear structures and foot orthoses.

The mean relative angle of $FF_{ad/ab}$- $RF_{ev/in}$ was different among the five intervals of the stance phase and between the barefoot and shod conditions. A statistically significant coupling relationship was previously reported between forefoot transverse and rearfoot frontal plane motions (Naster et al., 2002; Pohl et al., 2006). Significant changes in the coupling relationships between the forefoot transverse and rearfoot frontal plane motions could not indicate the amount of tibial rotation during dynamic motions. This finding is in contrast to the concept that cutaneous receptors of the forefoot may motivate the contraction of inverting muscles leading to control of the rearfoot and tibial rotations. Rattanaprasert et al. (1999)

found that the tibialis posterior muscle support the arch of the foot and the frontal plane rearfoot motion was not affected by the loss of tibialis posterior muscle. They suggested that the motion of the forefoot relative to the rearfoot was mostly about the behavior of the longitudinal arch. This is in agreement with Buchanan and Davis (2005) who observed a significant relationship between forefoot angle and navicular drop ($r = 0{,}55$, $P < 0{,}001$) in healthy subjects. Lee et al. (1999) reported that the medial foot length was positively correlated with relative forefoot abduction while Aramantzios et al. (2005) found that the motion at the forefoot relative to the rearfoot is influenced by the mats with different hardness during landing. They suggested that the acting forces can not possibly be compensated by means of muscular actions in the forefoot motion. In general, it is speculated that the transverse plane motion of the forefoot with respect to the rearfoot could due to the flexibility of arch in absorbing shock and adapting to the ground floor surface. Therefore, the effect of sandals on the forefoot transverse plane motion could have a greater contribution to the flexibility of arch than the amount of tibia rotation.

The results of present study suggest that the frontal plane forefoot-rearfoot coupling pattern exhibit a similar trend in the out-of-phase pattern at heel-strike to the in-phase pattern at the mid-stance in asymptomatic feet. This finding could be compared with the frontal plane forefoot-rearfoot coupling pattern in symptomatic feet during running. Furthermore, variations in the forefoot transverse plane motion and the rearfoot frontal plane relationship could be related to the flexibility of arch in absorbing shock and adapting itself to the ground floor surface. Therefore, small changes in the tibial excessive motion could be expected when forefoot postings in foot orthotics change the forefoot transverse plane motion during running. In general, this finding questions the rational for the prophylactic use of forefoot posting in foot orthoses. Finally, sandal had no significant effects on rearfoot-tibia coupling motions. This result eliminates the possible confounding effects of sandals on the outcome measures of rearfoot-tibia coupling motions when they are tested with foot orthoses.

5. CONCLUSION

Significant variations in the forefoot adduction/abduction and rearfoot eversion/inversion coupling patterns could have little effect on the amount of tibial internal rotation excursion. Yet it remains to be determined whether changes in the frontal plane forefoot-rearfoot coupling patterns influence the tibia kinematics for different shoe wears or foot orthotic interventions.

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Figure 5.1 Anterior (a) and posterior (b) views of marker placements in barefoot condition. Lateral view in shod condition (c).

Figure 5.2 Angle-angle plot for rearfoot eversion/inversion versus tibial internal/external rotation from heel-strike (HS) to toe-off (TO) in the shod and barefoot conditions.

Figure 5.1

a) b)

c)

Figure 5.2

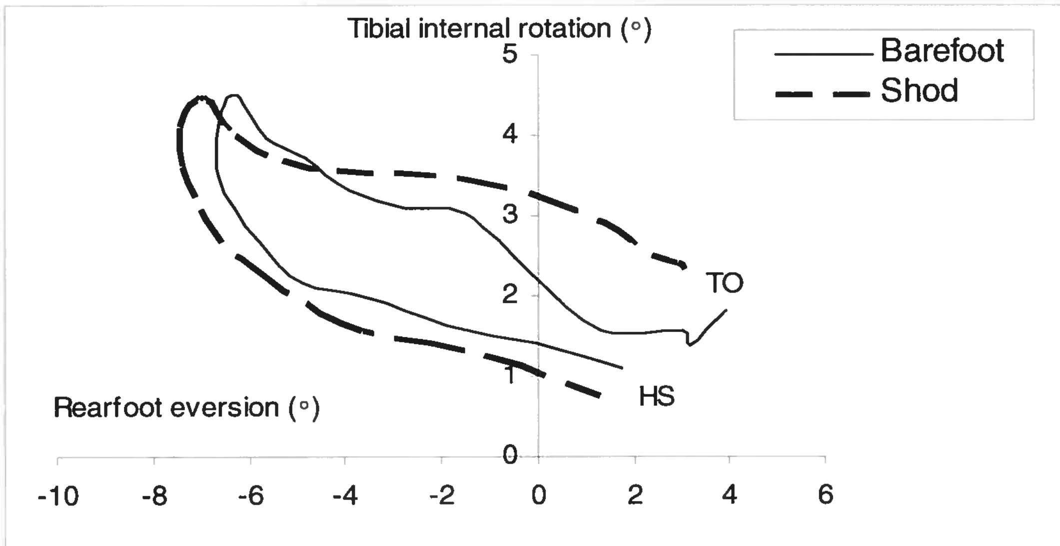
Tibial internal rotation (°)
Barefoot
Shod
TO
HS
Rearfoot eversion (°)
5
4
3
2
1
0
-10
-8
-6
-4
-2
0
2
4
6

Table 5.1 Mean values (standard deviation) of the excursion of rearfoot eversion and tibial internal rotation (degree) in barefoot and shod conditions.

Variables	Barefoot	Shod	*P* values
Rearfoot eversion excursion	-8,8(2,3)	-9,0 (4,1)	0,79
Tibial internal rotation excursion	4,1 (2,0)	4,0 (2,0)	0,89
Eversion/tibial internal rotation ratio	-1,8 (2,0)	-2,2 (1,1)	0,44

P- values compared barefoot and shod conditions.

Table 5.2 Mean (standard deviation) of the forefoot eversion/inversion and rearfoot eversion/inversion absolute relative angle (degree) across the five intervals of stance phase in barefoot versus shod running.

Intervals	Barefoot	Shod	*P* values
Heel-strike	41,3 (30,1)	31,2 (27,4)	0,28
Foot-flat	25,1 (22,7)	34,4 (23,3)	0,24
Heel-rise	20,9 (15,3)	25,0 (19,1)	0,48
Push-off	31,4 (21,3)	30,3 (28,4)	0,88
Toe-off	27,9 (29,9)	23,8 (32,0)	0,70

P- values compared barefoot and shod conditions.

Table 5.3 Mean (standard deviation) of forefoot dorsi/plantarflexion and rearfoot eversion/inversion absolute relative angle (degree) across the five intervals of stance phase in barefoot versus shod running.

Intervals	Barefoot	Shod	*P* values
Heel-strike	43,8 (32,9)	41,3 (19,5)	0,84
Foot-flat	21,7 (11,7)	27,5 (18,9)	0,42
Heel-rise	18,1 (12,8)	22,1 (15,6)	0,55
Push-off	29,4 (21,9)	34,3 (24,2)	0,55
Toe-off	43,6 (18,4)	38,6 (18,7)	0,60

P- values compared barefoot and shod conditions.

Table 5.4 Mean (standard deviation) of forefoot adduction/abduction and rearfoot eversion/inversion absolute relative angle (degree) across the five intervals in barefoot versus shod running.

Intervals	Barefoot	Shod	*P* values
Heel-strike	71,5 (45,4)	34,5 (28,2)	0,01*
Foot-flat	21,2 (18,2) [Heel strike]	20,7 (15,7)	0,93
Heel-rise	17,9 (14,5) [Heel strike]	16,0 (13,5)	0,67
Push-off	33,2 (21,1)	42,1(23,7) [Heel rise]	0,32
Toe-off	39,9 (27,1) [Heel rise]	41,0 (25,1) [Foot flat]	0,91

Significant differences between intervals in each condition are shown by superscript (P<0,05). P- values compared barefoot and shod conditions

Chapter 6

6. MANUSCRIPT 3

Title: **Effect of foot orthoses on amplitude and timing of rearfoot and tibial motions, ground reaction force and knee moment during running**

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Journal: **Submitted to Journal of Science and Medicine in Sport**

Key Words: **Foot orthoses; rearfoot eversion; tibial internal rotation; ground reaction force; knee moment; running**

ABSTRACT

Background: Changes in amplitude and timing of rearfoot eversion and tibial internal rotation by foot orthoses and their contributions to ground reaction forces and joint moments are not well understood. The objectives of this study are to test if orthoses modify the amplitude and time to peak of rearfoot eversion, tibial internal rotation, active ground reaction force and knee adduction moment, and determine if rearfoot eversion, tibial internal rotation amplitudes are correlated to peak active ground reaction force and knee adduction moment during the first 60% stance phase of running.

Method: Eleven healthy men ran at 170 steps per minute in a shod and a shod with foot orthosis conditions. Video and force-plate data were collected simultaneously to calculate motions, forces and moments. Paired t tests, two repeated factor ANOVAs and Pearson correlation were performed to test the hypotheses (P<0,05).

Findings: Wearing semi-rigid foot orthoses reduced significantly rearfoot eversion and peak active ground reaction force. No significant time differences occurred among the peak rearfoot eversion, tibial internal rotation and peak active ground reaction force in both conditions. A positive different from zero correlation was observed between peak knee adduction moment and the amplitude of rearfoot eversion during running in both conditions.

Interpretation: Findings imply that modifying rearfoot frontal plane motion could be related to a reduction of excessive knee adduction moment but not to a cushioning of the vertical ground reaction force. However, the cushioning characteristics of the orthoses were found to reduce the vertical ground reaction force during running.

INTRODUCTION

Excessive rearfoot eversion (REV) and tibial internal rotation (TIR) during repetitive motions such as in running could lead to Achilles tendonitis, medial tibial stress syndrome (Clement et al., 1981; Viitasalo & Kvist, 1983), patellofemoral pain syndrome and knee injuries (van Mechelen, 1992; Stacoff et al., 2000). Foot orthoses are prescribed to align the rearfoot and limit its motion relative to the tibia. Previous studies reported that foot orthoses reduced either REV (MacLean et al., 2006) or TIR (Nawoczenski et al., 1995; Stacoff et al., 2000). Because the timing and amplitude of REV and TIR attenuate ground reaction force and dissipate stress, their reduction may not be the only function of foot orthoses (Hreljac et al., 2000; Nester et al., 2003; Stergiou and Bates, 1997; Tiberio, 1987). Limiting foot and leg movement by means of foot orthoses could also reduce muscle force and modify lever arms (Nigg et al., 1999). These changes in the components of joint moment could affect load distribution at the knees. Generally, the effect of reducing REV and TIR amplitudes with the use of foot orthoses on ground reaction forces and muscle moments is not well understood during running.

Timing of peak REV, TIR and knee flexion must be synchronized during early stance phase of gait for cushioning external force and absorbing shock (Bates et al., 1978; Tiberio, 1987). If REV and TIR peak continue beyond mid-stance during knee extension, this leads to an antagonist motion and knee injury (Tiberio, 1987). Subotnick (1985) reported that peak REV and TIR must occur before mid-stance to attenuate the peak vertical reaction force and the foot must supinate to become rigid at push-off. Little evidence exists on the timing among the peak REV, TIR and vertical ground reaction force during the stance phase of running. To our knowledge, no research addressed whether foot orthoses could change these timing events or not.

Vertical ground reaction forces were proposed as indications of loading conditions (Andriacchi 1994; Cole et al., 1995; Perry & Lafortune 1995). They could increase when normal foot and tibia motions are restricted or exaggerated (Perry & Lafortune, 1995). Perry and Lafortune (1995) reported that the active vertical ground reaction force (AVGRF) increased with the use of medially posted orthoses when the peak REV angle was reduced by 6,7° during running. In contrast, no significant change in peak AVGRF was observed when REV decreased by 3° (Mündermann et al., 2003) or when the rearfoot was everted excessively (Perry & Lafortune, 1995). These findings suggest that the amplitude of lower-limb motion could contribute to the vertical ground reaction force during running. Though foot orthoses control rearfoot eversion and tibial rotation (without regard to knee flexion), its efficacy at cushioning the peak vertical ground reaction force is still unclear.

Knee adduction moment (KAM) is thought to increase load in the medial aspect of the tibial plateau and femoral condyle, thereby causing knee pain in runners (Hurwitz, 2000; Hunt et al., 2006). To our knowledge no investigation was carried out to determine if changes in the REV and TIR amplitudes caused by foot orthoses could be associated to peak KAM during running. Because REV and TIR as well as peak AVGRF and KAM occur during the first 60 % stance phase of running, we we hypothesized that foot orthoses perturb their relationships during this period of stance phase. This study aimed to test if foot orthoses modify the amplitude and time to peak of REV, TIR, peak AVGRF and KAM, and determine if REV and TIR amplitudes are correlated to peak AVGRF and KAM during the first 60 % stance phase of running.

METHOD

Eleven able-bodied men having an average age of 27,9 (SD 4,5) years, weight of 86,1 (SD 7,0) kg and height of 179,0 (SD 5,9) cm volunteered for this research. The number of subjects was based on a α of 0,05 and a β of 0,20 according to Erdfelder et al. (1996). None of the subjects had any orthopedic or neurological ailments affecting their running gait. The experimentation procedures, approved by the Hospital Ethics Committee, were explained to all participants.

Six cameras (Motion Analysis Corporation, Santa Rosa, CA, USA) were arranged in two arcs of 120° positioned on the left and right sides of a force-plate (AMTI, Watertown, MA, USA, 960 Hz) located in the middle of a 10 m runway. The capture volume (0,5 m in length × 0,5 m in width × 0,75 m in height) covered the right lower-limb motions during the running trials. Nine reflective skin markers, 16 mm in diameter, were attached to the right foot and tibia as shown in Figure 6.1. Three of them were fixed over the posterior calcaneus, medial and lateral sides of the calcaneus to define a rearfoot coordinate system according to the ISB recommendations (Wu et al., 2002). An additional marker was located on the extremity of the second toe to calculate the ankle moment. Five other markers were placed over the tibial tubercle, head of fibula, anterior middle aspect of the tibia, medial malleolus, and lateral malleolus to define the tibial coordinate system. During the running trials, three markers namely, the extremity of the second toe, medial malleolus and tibial tubercle, were removed and calculated as virtual markers. This was done to avoid marker dropout, skin movement artifacts and hidden markers which may occur for landmarks on the medial side of the foot and tibia during running trials.

Subjects were tested in two running conditions. The shod condition consisted of sandals where three straps covered the posterior side of the calcaneus, the tarso-metatarsal joints and the metatarso-phalangeal joints. Sandals allowed an easy tracking of the markers during running trials, and were previously used to evaluate the effects of foot orthoses on the rearfoot and tibial motions (Eslami et al., 2007: Branthwaite et al., 2004; Nawoczenski et al., 1995). In the shod/orthoses condition, subjects were fitted with semi-rigid foot orthoses. The orthoses were fabricated from a ductile polypropylene plastic material (3 mm in thickness) designed to provide rearfoot stabilization and arch support. They were fixed in the sandals by means of a double-sided adhesive tape.

Ten running trials were performed in the shod and shod/orthoses conditions in a block random order. In each experimental condition, subjects ran in a comfortable pace at a cadence of 170 steps per minute controlled by means of a metronome. The video cameras (60 Hz) were synchronized with the force plate measurements (960 Hz). All kinematic data were filtered at 8 Hz with a low-pass zero phase shift fourth-order Butterworth filter. Three-dimensional joint rotations were calculated according to Grood and Suntay (1983). The sequences of rotations were plantar/dorsiflexion about the medio-lateral axis of the proximal segment, abduction/adduction about the floating axis, and inversion/eversion about the anterior/posterior axis of the distal segment. TIR was measured as a transverse plane motion of the foot with respect to the tibia. The segment inertial parameters were obtained from the adjustments to Zatsiorsky-Seluyanov's parameters as outlined by De Leva (1996). A Newton-Euler inverse dynamics approach was applied to calculate knee moments. The knee moments were normalized with respect to the subject's body mass.

There were four dependant variables. Rearfoot eversion and tibial internal rotation amplitudes were calculated by determining the difference between their respective values at heel-strike minus their maximum value occurring during the first 60% of the stance phase. Peak AVGRF and peak KAM were determined during that period of stance phase. Furthermore, time to peak for each variable was identified and reported as a percentage of the stance phase. The average of the amplitude and normalized time to peak values was taken over the ten running trials for each dependent variable and for the shod and shod/orthoses conditions. Kolmogorov-Smirnov tests performed on the averages of the dependent variables showed no significant differences with a normal distribution ($0{,}46<Z<0{,}95$; $0{,}32<P<0{,}99$).

Paired t tests were performed to compare the mean amplitude of the four dependent variables between shod and shod/orthoses conditions. A two-repeated factor ANOVA (4 variables × 2 conditions) tested the difference on normalized time to peak data for all four dependent variables between the shod and shod/orthoses conditions. Protected t tests were used if a main effect was found to be significant. Pearson's correlations were performed to determine if the amplitude of REV was correlated with the peak AVGRF and KAM and verify if the amplitude of TIR was correlated to the peak AVGRF and KAM in the shod and shod/orthoses conditions. The level of significance was set at P<0,05 for all tests.

RESULTS

Figure 6.2 shows the mean amplitudes of the four dependant variables for the shod and shod/orthoses conditions during running. With semi-rigid orthoses, REV amplitude and peak AVGRF decreased respectively by an average of 4,1° (P=0,001) and 6% (P=0,008), when compared with the shod condition. No statistical difference was found in the mean TIR amplitude (P=0,06) and peak KAM (P=0,19).

For the normalized time to peak values, a significant main effect of variables was observed ($F_{3,30}$ = 5,6; P=0,003). The protected t post-hoc tests revealed that significant timing differences were observed between peak KAM (30,4%) and peak TIR (45,6%) (P=0,02) as well as peak KAM and peak AVGRF (39,7%) (P=0,03). No significant difference was observed between time to peak REV (38,4%) and time to peak for the other variables (0,06<P<0,35). No main effect was noted for the conditions ($F_{1,10}$=1,04; P=0,33), nor for the interaction effect of variables and conditions ($F_{3,30}$ =1,35; P=0,27). These findings imply similar time sequences for the four variables in the shod and shod/orthoses conditions.

Table 6.1 presents Pearson's product moment correlation coefficients between variables in the shod and shod/orthoses conditions. The average coefficient of correlation r was relatively low at 0,34. Statically significant correlations were observed between the REV amplitude and peak KAM in both shod (r = 0,59) and shod/orthoses (r = 0,65) conditions.

DISCUSSION

The results of the present study suggest that wearing a semi-rigid foot orthosis reduces significantly REV amplitude and active ground reaction force with no significant decrease on TIR amplitude and peak KAM. Regarding the REV and TIR amplitudes, the findings are in accordance with MacLean et al. (2006), but in contrast to Nawoczenski et al. (1995), Nester et al. (2003) and Stacoff et al. (2000). Results revealed that a reduction of REV was not accompanied with an equal reduction in TIR during running. Variability in the movement pattern in the lower-limb segments in individuals could be the factor in response to foot orthoses. Bellchamber and van den Bogert (2000) found high inter-individual differences in lower-limb segments movement pattern. They found that during running, movement transfer was mainly from tibia to rearfoot, nonetheless, some subjects showed an inverse movement pattern. This movement transfer was suggested depending on the flexion position of the foot (plantar/dorsiflexion), loading of the ankle joint complex, fusion of selected joints and integrity of the ligaments (Hintermann & Nigg 1998). The observed disparity in the results of this study compared with previous studies for these variables could be attributed to individual differences in response to orthoses as well as the type of foot orthoses utilized.

With the use of orthoses, a reduction of 10-20% in peak AVGRF was reported in the literature. This amount of reduction was considered insufficient to prevent injuries (Nigg et al., 1999). In this study, the AVGRF was decreased by an average of 5,5%. This small reduction could be related to the flexibility of the semi-rigid orthoses. Peak AVGRF was proposed as a significant discriminator between groups of injured and uninjured runners with stress fractures (Messier et al., 1988; Grimston et al., 1994). It is speculated that ground reaction forces occurring during physical activities such as normal running might not be a major factor in the development of injuries in running. It is unknown to which extent the peak AVGRF could be related to the risk of lower limb injuries during running.

The absence of a significant effect of the semi-rigid orthoses on KAM is supported by Maly et al. (2002) though these results are different from those presented by Kakihana et al. (2005), Mündermann et al. (2003) and Nester et al. (2003). A large KAM could increase the risk of overloading of the medial structures of the knee contributing to the iliotibial band friction (Andriacchi et al., 1985) and patellofemoral pain syndromes (Stefanyshyn, 2006) in runners. Wedged foot orthoses were used as a treatment in order to change load distributions at the knee. Yasuda and Sasaki (1987) found that a laterally wedged insole reduced the load in the medial compartments of the knee in standing. Keating et al. (1993) reported that a laterally wedged insole might be effective to reduce knee pain in osteoarthritis patients during the stance phase of walking. In our study, a semi-rigid orthosis did not change the peak KAM during running.

This study was the first to report the outcome of the use of foot orthoses on time to peak of REV, TIR, AVGRF, and KAM during the stance phase of running. Results suggest that the peak value for the KAM occurred earlier than peak AVGRF and TIR during the early stance in both conditions. This finding shows that the use of orthoses could not change the observed time differences. Hunt et al. (2006) reported similar finding although they tested only the frontal plane component of the ground reaction force. The lack of significant timing differences among peak REV, TIR and AVGRF in normal individuals indicates that the timing of these events does not differ during the first 60% stance phase of running. Abnormalities in foot structure or the misalignment of foot and leg may result in a disruption of these timing events which could be synchronized by wedged foot orthoses. This issue will be addressed in a future study.

In this study, the peak AVGRF was not correlated with the amplitude of REV and TIR in shod and shod/orthoses conditions during running. Mündermann et al. (2003) also reported no significant change in the peak active ground force while REV was decreased. In contrast, Perry and Lafortune (1995) noted that AVGRF was increased when REV was reduced by 6,7°. We suggest that, small and non-consistent reductions in the amplitudes of rearfoot eversion (without regard to knee flexion) by means of foot orthoses could not be associated with the observed reduction in peak AVGRF. It is speculated that timing and amplitude of the ankle and knee movements in the sagittal plane as well as shock absorbing characteristics of orthoses could be more effective in cushioning the peak AVGRF than foot pronation itself.

A significant positive correlation between REV amplitude and peak KAM was observed in shod and shod/orthoses conditions during running in healthy subjects. Keating et al. (1995) and Kerrigan et al. (2002) reported, during walking, a laterally wedged insole increased the eversion angle of the subtalar joint, therefore, reducing adduction moment at the knee. Kakihana et al. (2004) observed that wearing a laterally wedge during walking, decreased KAM as a result of more laterally shifted location of the COP. This can be attributed to a reduced knee moment arm length (Kakihana et al., 2005). Nigg et al. (2003) did not find significant change in the average shift of COP by the use of a medially wedge during running. They also reported no correlation between the COP location and knee moment during running. On the other hand, Nawoczenski and Ludewig (1999) indicted that a reduction of EMG activity for the biceps femoris with the use of orthoses during running. The decrease in EMG activity in the biceps femoris may be a response to the decreased requirements of this muscle for controlling tibial internal rotation when orthotics are worn (Nawoczenski & Ludewig 1999).These recent findings imply a likelihood of interaction between the foot and leg kinematic and muscle activity during running. To reduce peak KAM, muscle activity could be more affected than lever arm when the foot orthoses control rearfoot eversion during running. This could be possible with a greater reduction of REV amplitude. Findings of this study suggest that patients with excessive rearfoot eversion could have a greater response to the treatment of excessive loading in the medial compartment of the knee during running with medial wedged foot orthoses.

CONCLUSION

Wearing semi-rigid foot orthoses could reduce rearfoot eversion and cushion the active ground reaction force. It appears that the shod/orthoses condition did not change time to peak values for rearfoot eversion, tibial internal rotation and ground reaction force in able bodied subjects. Rearfoot eversion was associated with the knee adduction moment during running while rearfoot eversion and tibia internal rotation were not related to peak ground reaction force. These findings imply that modifying rearfoot frontal plane motion could be related to a reduction of excessive knee adduction moment but not to a cushioning of the ground reaction force. The cushioning characteristics of the orthoses were found to reduce the vertical ground reaction force during running

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Figure 6.1 Anterior and lateral views of marker placement in the barefoot and shod conditions

Figure 6.2 Mean amplitude of (a) rearfoot eversion, (b) tibial internal rotation, (c) peak knee adduction moment, and (d) peak ground reaction force. (*) indicates statistical differences between shod ☐ and shod/orthoses ■ conditions for $P < 0.05$.

Figure 6.1

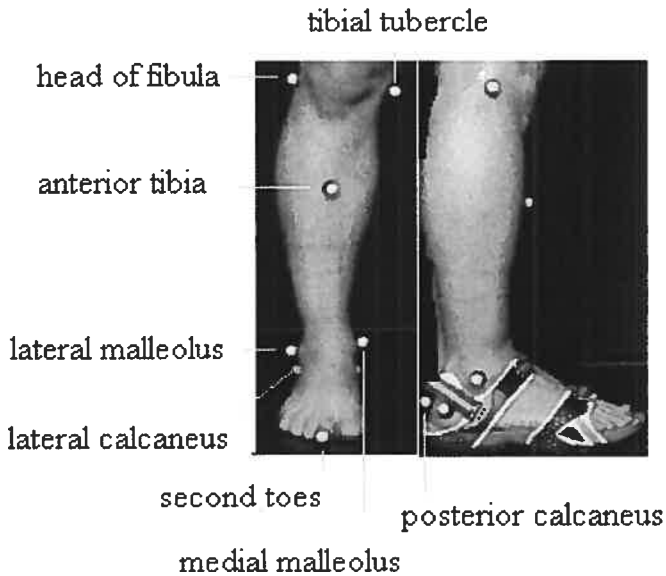
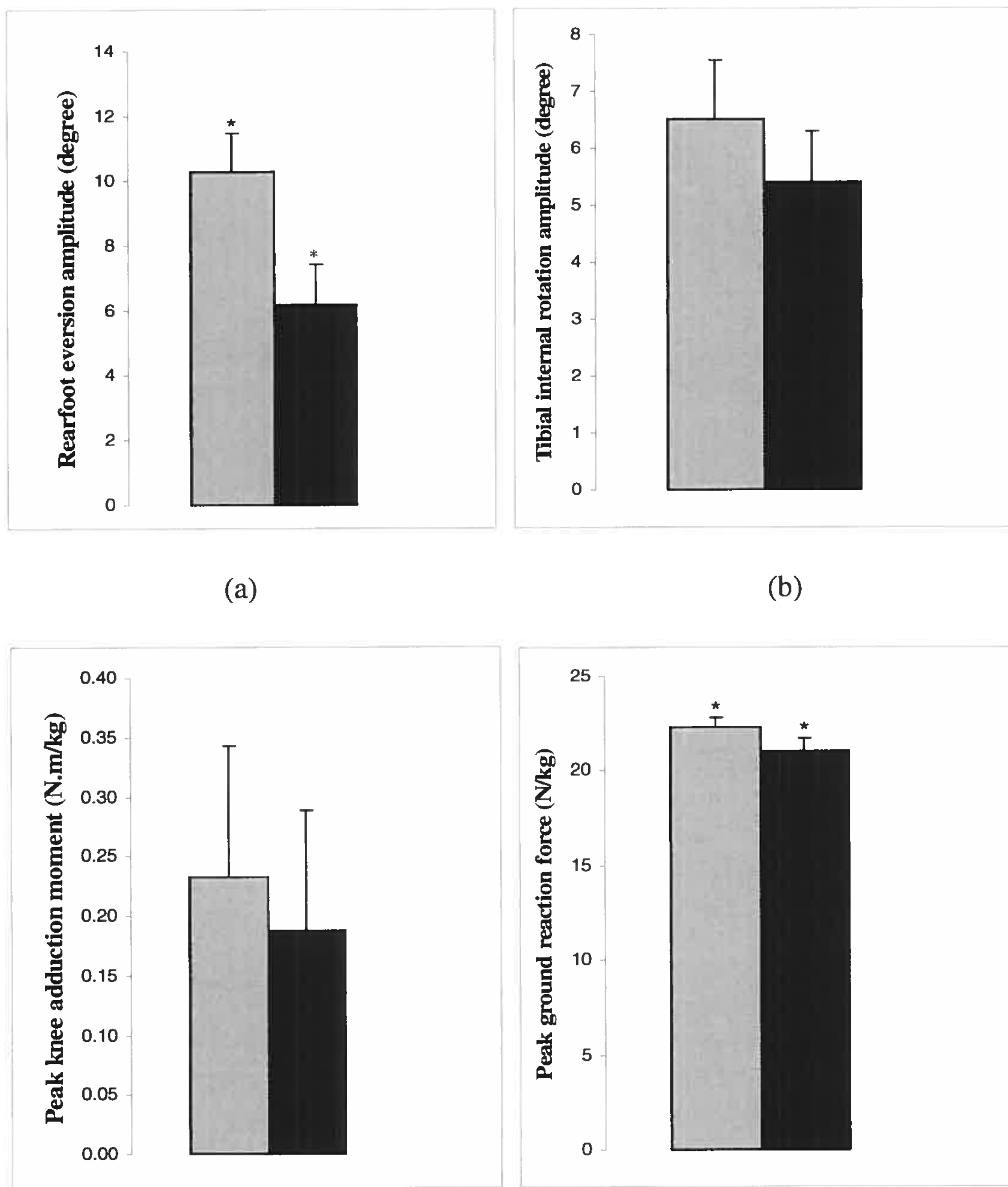

tibial tubercle
head of fibula
anterior tibia
lateral malleolus
lateral calcaneus
second toes
medial malleolus
posterior calcaneus

Figure 6.2



(a) (b) (c) (d)

Table 6.1. Correlation coefficients (r) and P values between rearfoot eversion (REV) and peak active ground reaction forces (AVGRF) and knee adduction moment (KAM) and between tibial internal rotation (TIR) and AVGRF and KAM in shod and shod/orthoses conditions during running (P<0,05).

	Shod		**Shod/Orthoses**	
	r	***P***	***r***	***P***
REV-AVGRF	**0,17**	**0,61**	**0,32**	**0,32**
REV-KAM	**0,59**	**0,04**	**0,65**	**0,02**
TIR-AVGRF	**0,48**	**0,12**	**0,36**	**0,27**
TIR-KAM	**0,13**	**0,69**	**0,03**	**0,90**

Chapter 7

DISCUSSION

The general objective of this research project was to determine the effect of foot-angle changes on the kinematics of foot and its upper joints and segments, and their relationships with ground reaction force and knee adduction moment. This chapter will first argue that the body moves as a multilink structure during single-limb standing posture. Next, foot motion patterns and the tibial internal rotations will be described during the stance phase of running. Then, rearfoot eversion and tibial internal motions, and their role in reducing knee moment and ground reaction forces will be discussed in relation to the use of foot orthoses. Finally, the limitations of this study, and a look at some of the implications for further investigation will close this chapter.

7.1 Foot Angle Changes and Variability of Body Joints and Segments during Single-Limb Stance

The first specific objective of this research was to see how single-limb standing posture is affected by the lower-limb joints, pelvis and trunk, when a wedge orients the foot in any of four directions. Because postural control during single-limb stance is often evaluated by measuring the amplitude and the velocity of centre of pressure (Baier and Hopf, 1998; Hertel et al., 2002), the contribution of body joints and segments has remained unknown. This study is the first study to explore this territory in a single-limb stance test.

Single-limb stance was selected because it increases the challenge of maintaining balance requiring reorganization of the center of mass over a small and narrow base of support (Riemann et al., 2003). Changes in the foot-angle during single-limb stance could increase the tendency toward more postural movements at particular joints or segments. Therefore, the contribution of the joints and segments to the maintenance of posture could be displayed when foot is oriented in a given direction.

It was shown that, during single-limb stance, there was greater angular variability in all the joints and segments for all wedge positions than in the no wedge condition. This finding supports the idea that, during single-limb stance, the body maintains posture as a multilink structure moving in different planes of movement when foot-angle changes. The choice of a postural strategy to offset instability depends on the available appropriate sensory information (Nashner & McCollum, 1985), the internal central commands, and/or the reflexive responses in terms of a stretch reflex (Gatev et al., 1999). Interestingly, orienting the foot to any given directions increased the variability of joint and segment located in the same plane of movement in a similar trend. When the foot was tilted laterally, the frontal plane angle variability for the subtalar joint and hip were greatly increased. It is suggested that orienting the foot laterally could change the Achilles tendon orientation and decrease sagittal orientation of the subtalar joint axis. The hip frontal plane motion variability could, in turn, modify the variability of the subtalar joint to reorganize the center of mass over a narrow base of support, during single-limb stance.

When the foot was oriented to the medial directions, the variability of the ankle, knee and hip angles in the sagittal plane, was significantly high. Angle variability of pelvis and trunk increased about 2 to 3 times higher than in the no wedge condition when the foot was oriented to a posterior direction. These larger angles of variability were observed when the subtalar joint and hip had relatively low variability in the frontal plane. This result is in consistent with the idea that, when postural control can no longer be adequately corrected by the subtalar joint motion, the body reacts as a multi-segmental chain with large correction movements of the proximal segments in different planes of movement (Tropp & Odenrick, 1988).

The variability of the proximal joints and segments were increased almost equally in controlling single-limb posture when the foot was oriented in the anterior or posterior attitudes. Anterior/posterior orientation of the foot could increase variability of the sagittal plane motion of the joints and segments during single stance-limb stance. High rotation of pelvis and trunk in the transverse plane could increase stability by minimizing the sagittal plane displacement of the centre of mass. The trunk with its comparatively great mass could quickly adjust posture during single-limb stance (Tropp & Odenrick, 1988). In general, changes in foot-angle by means of a wedge, could serve to maintain standing posture by targeting the response of specific body joints or segments response in their respective plane of movement. Orienting the foot in a given direction could either, bring the subtalar joint and ankle in their normal positions, or misalign the foot joints. This could serve to define an appropriate postural control strategy by influencing CNS data on both proprioception and mechanical properties (Nigg et al., 1999).

The foot-angle changes in different plane of movement primarily affect tibial rotation. Stubblefield et al. (2001) reported a strong cross correlation between the foot and tibial rotations in different planes of movement during single-limb stance. Excessive motion of the rearfoot and tibia was reported to cause various lower-limb injuries (Stacoff et al., 2000). These injuries often have appeared during repetitive motion such as running. Various types of footwear have been used to control these excessive motions, even though there has little assessment of the inter-relationships among the forefoot, rearfoot and tibia. The second study described the underlying mechanisms of the forefoot motion pattern with respect to the rearfoot and their effect on tibial rotations during the stance phase of running.

7.2 Forefoot-Rearfoot Coupling Patterns and Tibial Internal Rotation in Running

The goal of second study was to determine the motion patterns of the forefoot with respect to those of the rearfoot, and to examine their effect on tibial rotation. This study dealt with as asymptomatic feet in the stance phase of running. It is the first to introduce forefoot-rearfoot coupling motion patterns by determining the relative phase angle phase. Relative phase angle demonstrates an estimation of the in-phase or out-of-phase relationships of two predominantly sinusoidal oscillators (Peters et al., 2003; Hamill et al., 1999). Hamill et al. (1999) reported a relative phase angle of 45°

for EV/TIR at foot strike. This quickly changed to a more in-phase relationship (10°) that was maintained throughout the reminder of stance phase. They also reported an out-of-phase relationship at foot-strike for knee flexion-TIR, knee adduction-TIR and femoral internal rotation-TIR. Our findings showed an out-of-phase relationship between forefoot and rearfoot, at heel-strike, which transitioned to an in-phase relationship by mid-stance. It then reverted to its out-of-phase state, from mid-stance to toe-off. Because relative phase angle is typically derived from the position-velocity phase planes of two oscillating segments (DeLeo et al., 2004), a more out-of-phase relationship is likely to occur in the segments with less mass in the forefoot-rearfoot than in the rearfoot-tibial coupling motion.

Data from this investigation support the idea that variations in the horizontal plane motion of forefoot with respect to the rearfoot, have little effect on tibial internal rotation. This questions the rationale the use of prophylactic forefoot posting in foot orthoses to prevent excessive tibial internal rotation. Nordin and Frankel (2001) suggested that the forefoot motion, could contribute to shock absorption by increasing the flexibility of foot's arch. It is believed that the amount of tibial rotation could be affected by variations in the frontal plane coupling patterns of forefoot-rearfoot. This idea is based on the twisted-plate (Hunt et al., 2001) and mitered-hinge models (Inman & Mann, 1978) of the foot and ankle. Understanding frontal plane forefoot-rearfoot motion patterns and their contributions to tibial rotation is crucial because the forefoot posting is often used to reduce tibial rotation (Johanson et al., 1994; Pohl et al., 2006). The effects on tibial rotation brought on by changes in frontal plane forefoot-rearfoot motion patterns, are better observed when either, various foot interventions are chosen, or when faster running are performed. The third and final study of this thesis attempted to focus on the first 60% of the stance phase during running and provide some insight into the effect of foot orthoses in perturbing the contribution of rearfoot eversion and tibial rotation to peak knee moment and ground reaction force.

7.3 Rearfoot and Tibial Rotations in Relation to Ground Reaction Force and Knee Moment during Stance Phase of Running

The third specific objective was to determine the contributions of rearfoot eversion and tibial internal rotation to peak knee adduction moment and ground reaction force, during the stance phase of running. Findings from this study suggest that a decreased peak adduction knee moment could be associated with the reduction of frontal plane rearfoot eversion during the first half of the stance phase of running. This shows that, controlling the rearfoot frontal plane motions with respect to the tibia, could decrease the load which is placed on the medial part of the knee. To date, there has been no report on the contributions of rearfoot frontal plane motion on the peak adduction moment by using foot orthoses during running. Yasuda and Sasaki (1990) found that, in the standing position, the laterally-wedged insole increased the eversion angle of the rearfoot and thereby reduced the load in the medial compartment of the knee. During the stance phase of walking, an increase in the rearfoot eversion angle, brought about by a laterally wedged insole, might be effective for the reduction of knee pain in osteoarthritis patients (Keating et al., 1993). Because the increase in load distribution could contribute to syndromes affecting the knee, such as iliotibial band friction syndrome (Keating et al., 1993; Yasuda & Saski, 1990), it is suggested that using wedged foot orthoses could change load distribution at the knee during running. An appropriate alignment of the rearfoot segment with respect to the tibia and minimization of muscle activity, are thought to be factors in reducing knee adduction moment during running.

Because rearfoot eversion and tibial rotation are considered to be key components of lower-limb shock absorption during the first-half of stance (Hreljac et al., 2000; Naster et al., 2003), it was expected that reducing these motions would increase peak active ground reaction force. To test this assumption, a semi-rigid foot orthosis was used to control rearfoot and tibia rotation. In the third study, the use of foot orthoses brought about a significant reduction of peak active reaction force. This reduction, however, did not contribute to rearfoot eversion and tibial rotation. MacLean et al. (2006) and Mündermann et al. (2003) showed a significant reduction in rearfoot eversion, while they found no significant changes in the peak active reaction force. In this study, the reduction in the peak active reaction force could be attributed to the characteristic of foot orthoses utilized.

In summary, the three studies forming the core of this thesis attempt to provide a better understanding of foot motion, as it influences the proximal joints and segments. The major finding of this study is that foot-angle changes can affect both the kinematics and moment of particular proximal joints and segments during single-limb standing and running. The contribution of each body joint and segment to standing posture depends on the direction of foot angle. In addition, there is little support for the claim that variations in the frontal plane motion pattern of the forefoot, with respect to rearfoot, could affect the amount of tibial rotation during the stance phase of running. It was observed, however, that rearfoot motion during running contributes to the amount of frontal plane moment at the knee.

The results of this research could be used to estimate the proper foot alignment by ascertaining postural dysfunctions in symptomatic feet. Postural dysfunctions adversely affect the ability to control the body joints over a narrow base of support. It is possible that these dysfunctions change the strategy of contribution of body joints and segments in maintaining posture. For example, muscle weakness at the ankle result in large compensatory motions used by the hip and trunk motions to correct standing imbalance during standing posture (Horak & Nashner, 1986). Another result of our study was that a more out-of-phase relationship, for forefoot and rearfoot was observed in the strike-phase of running. It is unknown, however, that whether lower-limb injuries in runners could be prevented by foot interventions which change this out-of-phase relationship to one that is more in-phase. Furthermore, this coupling pattern may be different in symptomatic feet. Finally, findings show that excessive rearfoot eversion could be a risk factor because of an increased load at the knee.

7.4 Limitations

This work has some limitations and thus, cautions must be exercise in the interpretation of the findings of this research. First of all, the model representing the foot, utilized only two segments moving about the mid-foot. Though this model is more complex than many previous experimental models, it continues to lack some recognition of major segments and articulations of the ankle and foot. It has been shown that the first metatarsal may move relative to the navicular (Cornwall & McPoil, 2002) and thus, the measurements cannot be construed to strictly represent the mid-foot joints. To minimize the effect of this limitation, three-dimensional motion of the forefoot was considered to represent motion in the all joints in the forefoot with respect to rearfoot.

Another limitation is the difficulty to extrapolate meaningful results from a 170-step-per–minute speed running trial. It must be pointed out that, higher running speeds or sideward cutting movements could change the joint motion patterns. For example, an EV/TIR ratio of 1,53 was reported when normal subject ran at 3,35 m/s (McClay and Manal, 1997), while those who ran at 4 m/s showed an average of 1,32 (Nigg et al., 1993). Although many factors could account for these differences, it is currently unknown whether joint motion patterns change under different testing conditions (for example, sprinting or walking). Because changes in running speed could have confounding effects on the segments' coupling motion, this was controlled at 170-step-per–minute speed running trial by a metronome across all trials. Thus, the interpretation of observed joint motion patterns is limited to that running speed.

Regarding to foot orthoses effects on lower-limb, findings of third study are limited to the immediate effect foot orthoses. In fact, gait adaptations to foot orthoses can be divided into short-term and long-term adaptations categories. Short-term adaptation is defined as the immediate adjustment of the body's gait mechanics to a modification of footwear. Nawoczenski et al. (1995) acknowledge the distinction between short-term and longer-term adaptations and allowed for accommodation to footwear modifications before the collection of biomechanical data. Fisher et al. (2003) showed that, under experimental condition, the effects of footwear modification on the knee adduction moment slightly increased over a one-week wear period. Although, subjects in our study were given adequate time to adjust to foot orthoses, results indicate immediate effects or short-term adaptations.

Use of the EV/TIR ratios and CRP values has been limited to the interpretation of the motion related to running injuries. It is important to establish a healthy EV/TIR ratio and CRP benchmarks, nevertheless, the boundaries of normal joint coupling motion haven't yet been defined during running. This is because most of the findings have been based on studies with a relatively low number of subjects. Since a small change in EV/TIR ratio and CRP may considerably reduce the risk of injuries, one must be cautious in extrapolating how relevant differences in joint coupling might be used to predict running injuries. In the second and third study of this thesis, the examination of rearfoot motion, with respect to the tibia, shared the same model as previous studies. This enables researchers to make meaningful comparisons to data from running-task studies.

7.5 Future Studies

The first study attempted to determine the contributions of the body joints and segments in single-limb standing posture when the foot is oriented in different positions. High angle variability was observed in the effects of foot-angle changes. Because the purpose of using foot orthotics is to improve stability by aligning the foot with its proximal segments, it remains unknown whether high joint angle variability is related to better control of posture. Combining force plate data with body joints and segments variability, could explain in part the role of body joints and segments as a source of improved stability in single-limb stance.

In the clinical setting, practitioners must be cautious when comparing different foot types with, either postural control strategies or joint coupling motions. The results of previous studies indicate that healthy individuals with cavus feet have a significantly larger centre of pressure area during single-leg stance than do individuals with rectus feet (Hertel et al., 2002). Moreover, Nawoczenki et al. (1995) reported that, by using foot orthoses, there was a significant decrease in tibial internal rotation in individuals with cavus feet. To the contrary, no change was observed in the low arch group. In the research setting, subjects should be tested with different type of feet in postural control strategies and joint coupling motions. This provides the information that if the type of feet could be taken into account while using a single, specific foot modification during standing and running.

Muscular activity is thought to modulate joint movement patterns during gait (Zernicke and Smith, 1996) which may produce changes in subtalar joint kinematics. This is consistent with the idea that pronation during running is increased over time, as muscle fatigue increase (Fromme et al., 1997). It is possible that an orthosis increases the afferent feedback emanating from cutaneous receptors. This, in turn, can lead to decreased eversion due to muscular contraction of invertor muscles (Stacoff et al., 2000). Future studies should attempt to isolate muscle activity during running under barefoot, shod and orthoses conditions.

The foot joints included tibial-femoral and hip–knee coupling motion related to running injuries may not be presented prior to injuries. Prospective studies are needed to provide insight into relationships between joint coupling and injury prevalence. Information from these studies will provide a foundation upon which joint coupling motions can be developed. Examining these coupling motions in relation to lower-limb motions during running, could lead to effective approaches to injury prevention. Furthermore, studies of pathology-specific populations will help elucidate the mechanisms behind the inter-relationships of lower limb joints, as well as the injuries that result from these inter-relationships. Findings will provide clinicians additional evidence for deciding the most appropriate treatment for lower-limb injuries.

To sum up, future areas of research should include the development of normative studies with larger subject numbers and different foot types to further define the normal bounds of joint coupling. Also necessary, are more in-depth assessments of other joint coupling relationships in injury prevalence in prospective studies. Information from these studies will provide a solid foundation for effective injury-reducing intervention strategies.

Chapter 8

CONCLUSION

This research project investigated the effect of foot-angle changes on its proximal joints and segments during standing and running. One aspect of this research was to determine angle variability during single-limb stance, as it was manifested in the subtalar joint and its proximal joints and segments, when the foot was oriented in four directions. The findings suggest similar contributions for the joints and segments which are in the same plane of movement. Compared to no wedge condition, variability of the subtalar joint was 6 times larger in the frontal plane when foot was oriented laterally. Angle variability was 3,5 times greater in the ankle in the sagittal plane, and 2,5 times greater in pelvis and trunk in the transverse plane when the foot was in the anterior or posterior attitude. According to these findings, orienting the foot in a given direction by means of a wedge changed the contributions of body joints and segments in single-limb standing posture.

Another important contribution of this thesis was to document the variation of forefoot-rearfoot coupling patterns, and determine their effect on tibial internal rotation during the stance phase of running. In asymptomatic feet, forefoot-rearfoot coupling was in a more out-of-phase at heel-strike. This transitioned into an in-phase relationship by mid-stance. From mid-stance to toe-off, this coupling pattern transitioned back to an out-of-phase relationship. This could be compared with the forefoot-rearfoot coupling pattern in symptomatic feet during running. Compared to barefoot running, shod running demonstrated a more in-phase relationship of forefoot adduction/abduction and rearfoot eversion/inversion. It was further shown that, in respect to the rearfoot, the amount of tibial rotation at heel-strike would not be affected by the variation of forefoot motion in the horizontal plane. Yet, the amount of tibial rotation remains unknown when the rearfoot and forefoot coupling motion is manipulated in the frontal plane. Clinically, if forefoot posting in foot orthoses controls abduction/adduction movement of the forefoot, this may not effectively control tibial internal rotation. It is speculated that, forefoot motions contribute more by increasing the arch flexibility than tibial rotation in absorbing the shock.

A notable finding was that the amplitude of rearfoot eversion was positively correlated with peak knee adduction moment and a weak association of rearfoot eversion and tibial internal rotation with peak vertical ground reaction force. Foot orthoses could contribute to the reduction of excessive knee moment and, thus, decrease high load at the knee through the control of rearfoot eversion/inversion. The shock absorbing characteristic of orthoses could be more effective in cushioning the peak active vertical ground reaction force than foot pronation itself. This characteristic of orthoses could be crucial in activities, such as landing, which have high loading conditions.

Regarding the general objective of this research project, the findings provide a foundation upon the importance of foot segment in relation to their proximal joints and segments during standing and running. An appropriate foot orthosis could modify the angle variability and amplitude of the subtalar joint and ankle in improving standing posture and reducing excessive knee adduction moment during running. The findings also question the rational for the prophylactic use of forefoot posting in foot orthoses in the reduction of excessive tibial rotation. It is anticipated that this work can help clinicians to find their way towards better solutions for wedge/orthotic fittings.

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