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The effect of shoe inserts on the kinematics of the lower extremity and center of pressure during treadmill running

Michael George Wortley

To the Graduate Council:

I am submitting herewith a thesis written by Michael George Wortley entitled "The effect of shoe inserts on the kinematics of the lower extremity and center of pressure during treadmill running." I have examined the final electronic copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Human Performance and Sport.

Songning Zhang, Major Professor

We have read this thesis and recommend its acceptance:

Accepted for the Council:
Carolyn R. Hodges

Vice Provost and Dean of the Graduate School

(Original signatures are on file with official student records.)

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We have read this thesis
and recommend its acceptance:

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Vice Provost and Dean of Graduate Studies

The Effect of Shoe Inserts on the Kinematics of the Lower Extremity and Center of Pressure During Treadmill Running

A Thesis
Presented for the
Master of Science
Degree
The University of Tennessee, Knoxville

Michael George Wortley
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Dedication

For my parents, George and Susan, for reading *Go Dog Go* to me when I was little, and thousands of other small steps towards making me the person I am today.

For my sister, Erin, and brother, Andrew, who always made me sit in the middle of the back seat.

For Lindsey, who inspires me to be better still.

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Lastly, I would like to thank my fellow graduate students, Kei Ohmura and Kurt Clowers, for their many hours of help setting up, taking down, and testing equipment.

Abstract

Objective: The purpose of this study was to investigate the effect of different orthotic shoe inserts on kinematics of the lower extremity and center of pressure (COP) during treadmill running.

Design: The study design was an experimental investigation of five different orthotic insert conditions.

Background: Orthotic shoe inserts have been used to treat a wide variety of running injuries. Despite their high success rate, the mode of action of orthotics is not well understood. Orthotics are often prescribed to reduce overpronation that has traditionally been evaluated through tibiocalcaneal eversion. The increased use of three-dimensional (3-D) kinematic methods in recent years has created an emphasis on tibia internal rotation as a component of subtalar joint pronation. No studies have yet shown a systematic effect of a series of medial and lateral heel postings on kinematics during running.

Methods: The subtalar joint axes of seven healthy male subjects (mean age: 25.1 ± 2.3 yrs.) were measured, and the subjects performed treadmill running at 3.8 m s^{-1} in a running shoe with the factory insoles (control) and four other orthotic conditions: 5° lateral post, neutral, 5° medial post, and 10° medial post. 3-D kinematic data of the foot and leg segments were captured using a four-camera system at 120 Hz. (Vicon), and COP data were collected with an in-shoe plantar pressure system at 120 Hz. (F-Scan). A one-way repeated measures ANOVA with post hoc comparisons was used to determine significant differences between conditions.

Results: The orthotics had no significant effect on frontal plane kinematics. Tibia internal rotation was reduced with the use of a 5° medially posted orthotic. The COP was shifted significantly posteriorly with increasing height of the heel.

Conclusions: More studies are needed in order to relate anthropometrics and orthotics to predictable changes in kinematics.

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Chapter I

Introduction

The mechanisms of lower extremity injuries among runners have been a topic of much debate. James et al. [1] cited training errors as the most prevalent cause of injury, accounting for 60% of all running injuries. Many authors have implicated anatomical misalignment and improper pronation of the subtalar joint [1-8], improper timing of the subtalar and knee joints [9, 10], peak ground reaction force (GRF) [4, 8], and improper running surfaces [11] as likely culprits. Partly due to the large number of potential confounders, no study has yet found a definitive cause for common running-related injuries such as knee pain, plantar fasciitis, or shin splints.

Orthotic devices are being used to treat a wide variety of running injuries with success. Gross et al. [12] surveyed 347 runners at events sanctioned by the New York Road Runners Club who used orthotic shoe insert. The most common reasons for using an orthotic insert were knee pain (47.3%), foot pain (33.4%), ankle pain (14.4%), shin pain (14.1%), and hip pain (8.4%). Seventy-five percent reported either complete resolution of symptoms or a great improvement, and 90.8% of the runners with favorable results continued to use the inserts after their symptoms had resolved.

Some experimental studies on orthotics have also shown positive results. McClay and Manal [7] showed that runners exhibiting excessive pronation also have greater transverse plane motion in the leg. Baitch et al. [13] showed that using a 25° inverted orthotic reduced the amount of calcaneal eversion during running. Nawoczenski et al. [14] found that use of custom orthotics reduced the amount of tibial internal rotation, and

Nester et al. [15] had similar results with 10° inverted orthotics. Bates et al. [3] filmed a group of six runners who had been prescribed orthotics, and found that the use of orthotics made the period of pronation of the subjects shorter, similar to a healthy subject who did not wear orthotics.

While there is evidence for the effectiveness of orthotics, the mode of action of these devices is still not well understood. Several recent review articles [16-19] have pointed out numerous contradictions in the literature as to the effectiveness of orthoses on altering lower extremity kinematics. Stacoff et al. [20] found only small and unsystematic changes in calcaneal eversion and tibial rotation between three different orthotic conditions. Nigg et al. [21] found that the reactions of each subject to an orthotic was often not what was expected.

Problem Statement

The purpose of this study was to investigate the effect of different orthotic shoe inserts on the kinematics of the lower extremity and the center of pressure (COP) during treadmill running.

Hypotheses

1. There will be no significant difference in kinematics or COP trajectory between running in the shoe only and running in the shoe with an unmodified over-the-counter orthotic insert.
2. Running with a laterally posted orthotic will increase the amount of tibia internal rotation and shift the trajectory of the COP laterally.
3. Running with a medially posted orthotic will decrease the amount of tibia internal rotation and shift the trajectory of the COP medially.

4. Running with a high medially posted orthotic will have a more significant effect than running with a low medially posted orthotic.

Delimitations

The study was conducted within the following delimitations:

1. Five to ten healthy males. They were screened to ensure that they were reasonably homogeneous in terms of foot structure and function and had no other physical impairments of the lower extremity at the time of testing.
2. Five test conditions including treadmill running in shoes, shoes with an unmodified over-the-counter orthotic insert, shoes with an over-the-counter insert with additional lateral posting, shoes with an over-the-counter insert with minimal extra medial posting, and shoes with an over-the-counter insert with excessive medial posting at a stride frequency of 80 strides/minute at a speed of 3.8 m/s.
3. Biomechanical signals collected from ten consecutive footfalls of the right foot beginning from the second minute of a two-minute running interval.
4. Data were collected at 120 Hz from a four-camera Vicon 460 Motion Capture System and an F-scan in-shoe plantar pressure measurement system.
5. Data collection for each subject was completed in two sessions.

Limitations

The study was limited by the following factors:

1. Subjects were limited to male recreational runner between the ages of 19-29 years.
2. Inherent errors from within the motion capture system and F-scan system are always present but were considered acceptable within the specifications of the manufacturers.

Assumptions

The following assumptions were made for this study:

1. In the stance phase of the treadmill running, the talus was in synchrony in the transverse plane with the tibia and fibula, and as a result, the subtalar joint axis could be assumed to rotate in the transverse plane with the leg.
2. Flexion of the ankle joint was primarily at the talocrural joint, and the contributions of the subtalar and midtarsal joints were considered negligible.
3. Biomechanical instruments used were accurate.
4. All of the subjects were free of lower extremity injuries at the time of testing.

   The performance of the subjects was symmetrical, therefore, only the right side was assessed.

Chapter II

Review of Literature

The mechanisms of lower extremity injuries among runners have been a topic of much debate. James et al. [1] cited training errors as the most prevalent cause of injury, accounting for 60% of all running injuries. Many authors have implicated anatomical misalignment and improper pronation of the subtalar joint [1-8], improper timing of the subtalar and knee joints [9, 10], ground impact forces [4, 8], and improper running surfaces [11] as likely culprits. Partly due to the large number of potential confounding variables, no study has of yet found a definitive cause for common running-related injuries such as knee pain, plantar fasciitis, or shin splints.

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While there is epidemiological evidence for the effectiveness of foot orthotics, the mode of action of these devices is still not well understood. Several recent review articles [16-19] have pointed out numerous contradictions in the literature as to the effectiveness of orthotics on altering lower extremity biomechanics. Payne and Chuter [17] argued that future studies must do a better job reporting information about the participants, such as the subtalar joint axis location, type of orthotics used, and presence of forefoot varus. Razeghi and Batt [16] concluded that the multifactorial aetiology of running injuries, combined with the inconsistent use of definitions and methodology, have made the literature difficult to interpret.

In this chapter literature will be reviewed in the areas of the kinetics and kinematics of the lower extremity, both with and without orthotic shoe inserts. The section on kinematics will describe some common methodologies, as well as experimental results from several studies on joint ranges of motion, running mechanics, and the effects of shoes and orthotics. The section on kinetics will focus primarily on plantar pressure measurement. The final section will describe some simple but powerful tools for predicting the outcome of orthotic interventions.

Orthotics and Lower Extremity Kinematics

A variety of different methods have been used to study motions of the lower extremity during dynamic motion, including radiocinematography [22], videofluoroscopy [23], computed tomography [24], and electrogoniometry [25, 26]. Videographic analysis is the method most commonly used, and will be the main focus of this section.

The simplest method for obtaining kinematic data during gait is two-dimensional videography. Credit for the marker placement is often given to Clarke et al. [27], but the method is fairly standard and has been used by many others [7, 9, 10, 13, 28-30]. The posterior distal third of the leg and posterior calcaneus are both bisected, and two markers are placed on each point of bisection. A camera is set up posterior to the subject so that it is perpendicular to the frontal plane, and the subject is recorded. During analysis, the angle between the two bisections is the angle of calcaneal eversion or inversion. Stacoff et al.

[31] used a variation on this marker set to also determine the frontal plane angle of the forefoot relative to the rearfoot.

Two-dimensional videography has been used in a variety of studies [3, 13, 27, 29-31] to study the effects of both shoes and orthotics on lower extremity kinematics. Stacoff et al. [31] compared rearfoot motion while running in barefoot and shod conditions in nine middle-distance runners on the Swiss national team. They found that maximum pronation was greater in shod versus barefoot conditions. In an investigation of the effects of various shoe design parameters Clarke et al. [27] found that shoes with soft midsoles (25 durometer) allowed a greater maximum pronation, a greater maximum velocity of pronation, and more total rearfoot motion than shoes with hard midsoles (45 durometer.) In contrast to these findings, Bates et al. [3] found that the angle of maximum pronation was less in shod versus barefoot conditions, although the difference was not significant. Bates et al. [3] found a significant reduction in maximum pronation while wearing shoes with a custom orthotic insert versus barefoot in subjects with clinically prescribed rigid orthotics, which agreed with the results of Baitch et al. [13].

Bates et al. [3] also investigated temporal events during the stance phase of running. Subjects were shown to begin pronation significantly earlier in the support phase in barefoot versus shod conditions. Although there was no significant effect on the time to maximum pronation, the time of maximum pronation and period of pronation were significantly longer in shod versus barefoot conditions. Temporal variables for the orthotic condition were significantly different from the barefoot condition, but not from the shoe only condition.

Areblad et al. [2] showed that errors in the alignment of the camera and variations in foot placement angle can cause significant errors in the two-dimensional measurement of rearfoot angle. For this reason, much of the research conducted on the kinematics of the subtalar joint have used three-dimensional methodologies. One of the classic three-dimensional models is the one developed by Newington Children's Hospital [32], in which two spherical markers and an elongated marker wand are used to define each segment. Equations that define the joint centers for this model are already well established, but require a relatively large number of anthropometric measurements. Another option is to place rigidly constructed marker triads on each on the posterior leg and calcaneus [2, 9]. This may be the simplest of the three-dimensional methods, but the talocrural and subtalar joints must be treated as a single universal joint instead of two distinct hinge joints.

A three segment model with twelve parameters was developed by van den Bogert et al. [33] and employed by several others [14, 34, 35]. In this model, three markers are placed on the leg at the fibular head, anterior aspect of the tibia, and proximal of the lateral malleolus, and three markers are placed on the foot at posterolateral aspect of the heel, the head of the fifth metatarsal, and the navicular. Optimization is used to estimate the twelve parameters and the position of the talocrural and subtalar joint axes.

Nawoczenski et al. [14] studied the effects of semi-rigid orthotics on three-dimensional kinematics of the lower leg. Subjects were placed into two groups of ten based on their lateral calcaneal inclination, lateral talometatarsal inclination, and anterior/posterior talometatarsal angle. The "high" group had a lateral calcaneal inclination greater than 25°, and the "low" group had a lateral calcaneal inclination of less than 20°. Subjects ran on a treadmill at a preferred speed. Results indicated that there was no difference in the way the two groups responded to the orthotic. There was a statistically insignificant decrease in tibial internal rotation of 2.1°. Nawoczenski et al. [14] also calculated a ratio of tibial abduction to tibial internal rotation, which was found to be significantly greater in the orthotic condition, reflecting a decrease in tibial internal rotation. Since tibial abduction and internal rotation both reached their peak values in the first 30-50%, it was concluded that the effects of orthotic inserts occurred in the first 50% of stance phase [14, 16].

McCulloch et al. [36] studied 10 subjects demonstrating a minimum of 3° of calcaneal eversion in relaxed stance. The subjects were filmed with a four-camera Motion Analysis system during treadmill walking at 2 and 3 mph, both with and without a custom orthotic insert. Pronation was significantly reduced throughout stance phase and the duration of stance time was increased. However, the orthotic intervention did not significantly reduce the velocity of pronation during the first 20 percent of stance phase.

The mechanical link between subtalar pronation/supination and tibial internal/external rotation have long been understood, but only recently have authors suggested tibial rotation should be used as an alternative to rearfoot angle [16, 37, 38]. Cornwall and McPoil [37] argued that measurements of calcaneal inversion and eversion were limited when subjects wore shoes. For this reason, they filmed eight subjects walking down a 12-m walkway. Results showed that, although the absolute values were not comparable, the correlation between tibial rotation and rearfoot angle was r=0.953 in all 16 feet tested. Nester et al. [38] conducted a study to establish normative data for shank rotation during normal walking. Twenty-five subjects walked at a controlled cadence of

108 steps/minute. The mean tibial external rotation at touchdown was 5.4° (SD 3.7°, range –0.2° to 13.3°,) mean maximum internal rotation (corresponding to calcaneal eversion) was –7.2° (SD 4.1°, range –19.8° to 2.0°,) and mean maximum external rotation (corresponding to calcaneal inversion at toe-off) was 9.5° (SD 5.2°, range –1.9° to 18.3°.)

McPoil and Cornwall [39] sought to compare the effectiveness of both rigid and soft orthotics in controlling tibial internal rotation during walking. Using 10 subjects and standardized footwear, they found that both the soft and rigid orthotics significantly reduced the amount and velocity of tibial internal rotation compared to no orthotics, but there were no significant differences between the two. Nester et al. [15] used 12 subjects to compare anti-pronatory and anti-supinatory orthotics. Anti-pronatory orthotics significantly reduced internal rotation, the initial peak velocity of internal rotation, and the total transverse-plane range of motion of the leg. Anti-supinatory orthotics increased the range of internal rotation and the total transverse range of motion of the leg.

Reflective markers placed on the skin and shoes have been an acknowledged source of error. Reinschmidt et al. [40] studied this topic specifically by filming five subjects running with both external and bone-pinned markers placed on the subjects simultaneously. The patterns of tibiocalcaneal inversion/eversion, abduction/adduction, and plantarflexion/dorsiflexion were the same for the external markers and bone markers. However, the external markers typically overestimated the maximum values found using the bone markers. The average difference at maximum eversion was 4.2°, approximately 34.7% of the total range of inversion/eversion. Similarly, the average difference in tibiocalcaneal abduction at midstance was 3.6°, or 51.2% of the to abduction/adduction range of motion.

Stacoff et al. [20] used reflective markers attached to intracortical bone pins to study the effects of two types of cork orthotics, one designed to support the medial arch and the other designed to support the heel, during running. Five subjects were used, but no statistical differences were found between the two types of orthotics. The only significant difference found was a reduction in the total amount of tibial internal rotation when running in either orthotic versus shoes only. These results are similar to the results of Nawoczenski et al. [14], which used external markers instead of bone markers.

In summary, although there is need for better reporting of the type of orthotics used in each study, the kinematic changes observed have been mostly small and unsystematic. Further study of orthotics must use three-dimensional methodologies because two-dimensional methods are incomplete and unreliable [2]. A closer examination on effects of orthotics on tibial internal rotation is warranted [9, 10, 15, 20, 39].

Orthotics and Kinetics

There are two main modalities of kinetic measurements used in biomechanical research: force and plantar pressure. Ground reaction forces are measured using a force platform, which subjects must move across in a defined manner. Although most force platforms are capable of measuring ground reaction forces as a three-dimensional vector, the vertical component is the focus of most studies. Common variables include the vertical impact force peak [30], vertical active force peak [30], peak vertical force [10, 25], and peak vertical loading rate [30]. Ground reaction force data can be combined with kinematic data to calculate more descriptive variables, as Bellchamber and van den Bogert [34] did to quantify the moment and power of the tibia rotating about its longitudinal axis during walking and running.

The effect of footwear and orthotics on kinetic variables has not been studied as extensively as their effect on kinematic variables. De Wit et al. [30] compared peak vertical impact force, peak vertical active force, and vertical loading rate between subjects running barefoot and shod. In nine subjects, only the loading rate was found to be significantly greater in bare feet. Milani et al. [25] had 27 subjects run in eight pairs of shoes that only differed in midsole stiffness in the heel and midfoot areas. An ANOVA revealed significant differences ($P<0.05$) between shoes in the amplitude of the peak vertical force, which increased as the shoes became softer, and the loading rate of the peak vertical force, which increased as the shoes became stiffer.

Plantar pressure is closely related to ground reaction force since pressure is equal to the force divided by the area over which the force in applied. Several commercial plantar pressure measurement systems, such as F-Scan (Tekscan, Boston), PEDAR (NOVEL GmbH, Munich, Germany), and Electrodynogram (Langer Biomechanics Corp, Deer Park, New York), use a matrix of small force transducers with a known area to determine the local pressure under discrete points on the plantar foot. These systems have been used to study the effects of shoe insert orthotics [41-43]. Kimmeskamp and Hennig [44] also used this method to study the differences in foot loading patterns between Parkinson patients and controls. A Pedar insole system containing 99 transducers and sampling at 50 Hz was used on 48 subjects (24 Parkinson, 24 controls) while they walked 11m at their preferred speed. The foot was divided into 10 regions for data analysis, and the trajectory of the center of pressure was determined. Results showed that the Parkinson subjects had significantly less loading of the lateral heel and significantly more loading of the medial midfoot. Parkinson patients also had a more medially oriented trajectory of the center of pressure on the plantar foot.

The literature on the effects of orthotics on plantar pressure contains a wide range of instruments, methodologies, and variables. Most often researchers examine the distribution of pressure [41, 43], the duration of distinct sub-phases of gait [42], and the correlation of localized heel pressure to other variables of gait or perception [25, 30]. One thing that is widely accepted among researchers in the need to distinguish between the effect of the shoe on the foot from the effect of the shoe and orthosis together [16].

Redmond et al. [43] compared the effects of a non-cast orthosis and a Root orthosis with a 6° rearfoot varus post in 22 subjects exhibiting excessive pronation. Using the PEDAR in-shoe measurement system, they divided the foot into six regions: heel, midfoot, lateral forefoot, medial forefoot, hallux, and the lateral digits. They found no significant differences between the non-cast orthosis and shoe-only conditions. In the heel, the Root orthosis significantly reduced the maximum force, peak pressure, and mean pressure, and increased the contact area. In the midfoot, maximum force and contact area were increased using the Root orthosis, resulting in a smaller peak and mean pressures. In the lateral forefoot, maximum force and mean pressure were significantly reduced using the Root orthosis. There was no significant difference in contact areas or peak pressure. There were no other significant effects on the other regions of the foot, except for a subgroup of participants who saw a sharp increase in peak pressure under the hallux as large as 125%. Further research is being conducted to determine why these subjects demonstrated that effect.

Brown et al. [41] did a similar study using several types of orthotics, including custom molded Plastizote (Zotefoam Inc.) orthotics, Spenco arch supports (Kimberly Clark Inc.), and off-the-shelf cork and plastic orthotics. Using the F-Scan system (Tekscan, Boston, MA), Brown et al.[41] divided the foot into eight regions: the great toe, forefoot, midfoot, heel, 2nd through 5th metatarsal heads, 1st metatarsal head, the lesser toes, and the whole foot. The Plastizote orthotics significantly reduced pressure under the heel and 2nd-5th metatarsal heads. The Spenco, cork, and plastic orthotics significantly increased the pressure in the midfoot. The cork orthotics reduced the pressure under the heel, and the plastic orthotics reduced pressure under the forefoot.

Reed and Bennett [42] investigated the effects of the Root and Blake orthotics on the temporal events of stance phase using an Electrodynogram system (Langer Biomechanics Corp, Deer Park, New York.) Pressure sensors were placed directly beneath the medial and lateral heel as well as beneath the 1st, 2nd, 3rd, and 5th metatarsal heads, and the hallux of the subject. Four sub-phases of stance phase were defined: load acceptance, load support, foot flat, and propulsion. Results indicated that the Root and Blake orthotics were not significantly different from each other, but both had a significant effect compared to the shoe-only condition. Specifically, both orthotics shortened the load support phase and prolonged the propulsive and foot-flat phases. The overall changes in the percentage of stance time spent in each phase was similar to that found by Bates et al. [3] in their kinematic study.

A major criticism of commercially available insole plantar pressure measurement systems is that they are unable to measure shear forces. A group in the United Kingdom has developed an in-shoe transducer capable of measuring longitudinal and transverse shear force [45-47]. Studies using this device have been mainly limited to studies of the diabetic foot, but the magnitude of the shear forces in the normal foot have been found to be much less than the magnitude of the vertical force.

In summary, whether soft or rigid, many shoe inserts increase the pressure in the midfoot region and reduce pressure in the heel and lateral forefoot. Only small differences have been found between non-cast orthotics and shoes, indicating that some customization of the shoe inserts may be necessary in order to elicit a statistically significant change in plantar pressure related variables.

Estimation of the Subtalar Joint Axis and Predicting the Effects of Orthotics

All motion about joints is caused by either external forces (such as gravity and ground reaction forces) or muscle forces, and the range of motion about a joint is limited by the bony architecture and ligaments [48]. Forces applied lateral to the subtalar joint axis cause pronation moments, while forces applied medial to the subtalar joint axis cause supination moments. Rotational equilibrium occurs when the sums of the moments acting in both directions are exactly equal (net moment = 0) [5, 6, 49], resulting in no motion of about the joint axis.

Orthotics are used to exert a supination moment about the subtalar joint axis of feet that are maximally pronated in relaxed stance position [49]. Many studies have shown that orthotics have a significant effect on either the distribution of plantar pressure [41, 43] or the center of pressure [21] during gait. The moment across the subtalar joint can be changed by altering the distance of the center of pressure from the subtalar joint axis [50]. In order to successfully predict how an orthotic is going to impact foot function, first the orientation of the subtalar joint and the range of motion of the ankle must be known.

The neutral position of the subtalar joint is the common reference point from which the majority of foot and ankle measurements are taken. The basic definition of subtalar neutral is "that position of the joint in which the foot is neither pronated nor supinated." [51]. There are several ways to find this position, including finding the position in which there is twice as much available supination as pronation [51, 52], observing when the lateral aspect of the calcaneus is parallel to the lateral aspect of the leg [53], observing when the skin lines over the sinus tarsi are neither stretched or wrinkled [53], and palpating the medial and lateral aspects of the talar head to find when it is maximally congruent with the navicular [52-54]. Once the neutral position of the subtalar joint is found, motions of the calcaneus relative to the lower leg can be measured by drawing lines bisecting the posterior calcaneus and the posterior leg and measuring the angle between them with a protractor or goniometer.

Åström and Arvidson [54] performed measurements on 121 healthy men and women from ages 20-50 years. With subjects lying prone while being measured with a goniometer, the mean subtalar neutral position was found to be 2° (± 3°) of valgus. The mean calcaneal eversion and inversion were 10° (± 4°) and 28° (± 6°), respectively. The mean forefoot varus was 7° (± 4°) for men and 6° (± 5°) for women. Ankle dorsiflexion and plantarflexion were 36° (± 6°) and 49° (± 7°), respectively. These values are only slightly different from the values found by Siegler et al. [55] using pneumatic actuators on fifteen cadaver feet, particularly tibiocalcaneal dorsiflexion (24.68° ± 3.25°), plantarflexion (40.92° ± 4.32°), inversion (22.41° ± 9.08°), and eversion (11.85° ± 10.34°).

Several weight-bearing measures have also proved useful to clinicians. Åström and Arvidson [54] measured the relaxed bilateral stance position of the calcaneus, finding a mean value of 7° (± 4°) of valgus. They also measured the mean tibial angle during stance to be 6° (± 2°) of varus. Torburn et al.[56] found that calcaneal eversion measured from single-leg standing was more reliable than maximum calcaneal eversion measured by passive positioning (intraclass correlation coefficients of 0.92 and 0.39, respectively,) and that eversion during single-legged stance was not significantly different from maximal eversion during fast walking.

Several methods have been proposed for determining the location and orientation of the subtalar joint axis. Morris and Jones[57] recommended drawing a circle of dots and the anterior superior aspect of the talar head and neck region, moving the subtalar joint through its range of motion, and observing the spot in that region in which no motion occurred. After repeating the same process on the lateral posterior aspect of the calcaneus, those exit points of the subtalar joint axis can be measured in the sagittal and transverse planes. A much simpler method proposed by Kirby[6] is known as the palpation method. In the palpation method, the subtalar joint is placed in neutral with the subject lying prone. The plantar surface of the foot is palpated, beginning at the posterior heel and moving towards the anterior foot in 1 cm increments. While the foot is being palpated one hand, the other hand is used to detect motions about the subtalar joint. Points that can be palpated without causing motion are assumed to be on the subtalar joint axis.

Phillips and Lidtke[58] incorporated the palpation into a much more complicated method of locating the subtalar joint axis. By measuring the rearfoot angle in neutral, maximally inverted, and maximally everted positions, as well as measuring the inferior calcaneal angles in the same positions, the authors developed a series of mathematical equations that define the subtalar joint as a line in three-dimensional space relative to the most inferoposterior point of the heel. This method makes it possible to report both the transverse and sagittal orientations of the subtalar joint axis, measurements commonly missing from the literature.

Once the orientation of the subtalar joint axis in neutral is known, mechanical principles can be applied to make a prediction about the effect of an orthotic insert. It is important to remember that the subtalar joint is a tri-planar joint, so the resultant change in moments about its axis could be observed in all three planes of motion. Although these measurements are all taken in a static situation, they offer a good beginning to understanding how the foot will move in a dynamic situation.

Summary

Although the mode of action of orthotic interventions has yet to be fully understood, they have had considerable clinical success, and as a result are widely prescribed by practitioners. Much of the research has focused on the effects of orthotics on kinematics, but results thus far have been inconclusive, owing largely to the wide variety of orthotics and experimental methods used. Several recent studies [21, 41-43] have focused on the effects of orthotics on plantar pressure, and have found that orthotics do have a significant impact on the distribution of pressure. Work still needs to be done to relate plantar pressure to kinematics and the anatomical differences between subjects in a dynamic situation.

Chapter III

Methods

The purpose of this study was to determine the effects of a series of different orthotic inserts on lower extremity kinematics and the center of pressure during running. The results of this research may be useful in determining the effectiveness of an orthotic intervention, and also provide some insight into the mode of action of orthotics. Subjects were screened to ensure that they were a homogeneous group. Each subject performed a two-minute run on a treadmill under five different footwear conditions while kinematic and plantar pressure data were recorded.

Nomenclature

STJ : subtalar joint.
A, B, and C: reflective markers on the heel triad. [2]
D, E, and F: reflective markers on the leg triad. [2]
P_0: the most posterioinferior point of he heel. [58]
P_{i0} and P_{i30}: points where the STJ axis intersects the planes z = 0 mm and z = 30 mm, respectively. [58]
α_n: orientation of the STJ axis in the sagittal plane relative to the, in the neutral position.
B_n: orientation of the STJ axis in the transverse plane relative to the ground longitudinal axis of the foot, in the neutral position.
RCS: room coordinate system. [2]
LCS: leg coordinate system. [2]
FCS: foot coordinate system. [2]
γ: frontal plane angle of the heel. [2]
λ: frontal plane angle of the leg. [2]
δ: sagittal plane angle of the heel. [2]
η: sagittal plane angle of the leg. [2]
ι: tranverse plane angle of the heel. [2]
κ: transverse plane angle of the leg. [2]
ζ: ankle plantar/dorsiflexion. [2]
μ: ankle eversion/inversion. [2]
θ: ankle abduction/adduction. [2]

Subjects

Potential subjects were recruited from the student population at The University of Tennessee and members of the Knoxville Track Club. All the potential subjects were males (ages 18-40) who had no recent history of severe lower extremity injury, and read and signed an informed consent form approved by the Institutional Review Board at The University of Tennessee prior to their participation in the study. All subjects participated in a screening session in order to determine STJ neutral, maximum inversion and eversion angles, relaxed bipedal calcaneal stance position, single-leg calcaneal stance position, STJ orientation (α_n and B_n), and forefoot varus. A subgroup of subjects who were the most similar in terms of B_n were selected to participate in the experimental testing session.

Instrumentation

All testing for this study took place in the Biomechanics/Sports Medicine Lab, Room 135 in the Health, Physical Education, and Recreation Building at The University of Tennessee. While subjects ran on a MedTrack ST Programmable treadmill (Quinton), simultaneous recording of 3D kinematic data and plantar pressure data were conducted. Standardized testing shoes, socks, and orthotic inserts were used as well.

Kinematics

A four-camera high-speed three-dimensional (3D) motion capture system was used to collect kinematic data (120 Hz, Vicon 460, Vicon, UK). The system was calibrated using a standard L-frame and a 500mm wand. The L-frame was oriented such that the y-axis was parallel to the direction of running on the treadmill, with the forward direction being positive.

Three 14-mm retroreflective spherical markers were mounted on a piece of vinyl in an equilateral triangle shape to form a single marker triad. One triad was attached to the posterior leg, and another was attached to the posterior heel counter of the shoe.[2]

The 3-D coordinates of each reflective marker were reconstructed using Vicon Workstation software running on a personal computer. The 3-D coordinate data were further processed using customized Matlab software to determine 3-D movements of the leg and foot, the ankle joint, and the STJ axis.

Plantar Pressure

Plantar pressure data were collected using the F-Scan In-Shoe Plantar Pressure Measurement System (Tekscan, Boston). The F-Scan sensors were prepared by trimming them to the size of the shoe, and then laminating them with EasySeal repositionable laminating sheets (GBC Office Products Group, Skokie, IL). This was done in order to protect the sensor from wrinkling and prolong the life of the sensor.

The plantar pressure data was collected using F-Scan software. The data, including the trajectories of the center of pressure, were used to compute were analyzed using custom software written in Matlab.

Synchronization

A flexible foot switch (MA-153, Motion Lab Systems, Baton Rouge, LA) was used to temporally synchronize the kinematic and plantar pressure data. The foot switch was adhered to the plantar surface of the heel using double-sided tape. Output from the footswitch was fed through the analog/digital converter of the Vicon system, and was used to indicate heel strike.

Treadmill

The treadmill used in this study was a MedTrack ST55 Programmable treadmill (Quinton, Bothel, WA.) There were no side-rails on the treadmill that could interfere with the cameras. Prior to testing, a calibration was performed to ensure the accuracy of the belt speed. The length of the belt was measured, and a flat circular retro-reflective marker was placed in the center of the belt. The treadmill was set to 13.6 km hr^{-1}, and filmed with a digital camera (JVC, 120 Hz) for 10 seconds with no one running on the treadmill. The belt speed was found to be 3.82 m s^{-1}. Then the procedure was repeated with a 70 kg man running on the treadmill, and the belt speed was found to be 3.79 m s^{-1}, a 1% decrease in belt speed.

Footwear

Subjects wore a standard lab shoe (Novetto, adidas, and Supernova Cushion, adidas) for all experimental conditions. The orthotics used were commercially available over-the-counter orthotic inserts (Powerstep, Stable Step Inc.) which had been modified for each condition. All modifications were performed by a certified orthopedic clinical specialist with many years of experience grinding orthotics clinically. Posting was made from SolFlex firm white ¼' (Sole Tech, Salem, MA), a blend of ethylene vinyl acetate (EVA) and styrene-butadiene rubber (SBR) with a Shore A 60-65 durometer. Wedges of SolFlex were glued to the heel of the orthotic insert from the most posterior heel to the beginning of the medial arch and then ground into the appropriate shape- either 5° medial, 10° medial, or 5° lateral.

Experimental Protocol

Subjects participated in two testing sessions. In the first session, basic anthropometric and ankle complex measurements were taken and subjects became familiar with the testing protocol by practicing running on the treadmill while matching their strides to a metronome set to 80 Hz. The second session was the primary data collection session.

The first session began by recording the subject's height and weight, followed by a series of measurements to characterize the subject's foot. All measurements were taken on the subject's right foot. With the subject lying prone, STJ was put into its neutral position by palpating for talonavicular congruency and the rearfoot angle was measured using a goniometer. Maximum calcaneal inversion and eversion were also measured with the subject lying prone. Forefoot varus was measured with subject lying supine. The foot was placed in the STJ neutral position as above, and the forefoot varus was measured as the angle between the plantar surfaces of the subject's forefoot and heel. Calcaneal eversion was also measured with the subjects standing in a relaxed bilateral stance position and a unilateral stance position.

The orientation of the subject's STJ axis in the neutral position was determined using the methods of Phillips and Lidtke [58]. A felt pen was used to mark the bisections of the distal third of the posterior leg, the posterior heel, and the inferior heel. The most inferior-posterior aspect of the heel was marked and labeled point P_0. With the subject lying prone and the foot placed in STJ neutral, a goniometer was used to measure the frontal plane angle between the leg and the calcaneus. This angle was also measured with the foot in maximum supination and maximum pronation, and the medial-lateral displacement of point P_0 was measured. The angle of the inferior heel bisection in maximum supination and maximum pronation was also measured. Finally, the transverse plane projection of the subtalar joint axis was drawn on the plantar surface of the foot using the palpation method [6]. The foot was traced on a sheet of paper, and the lines representing the inferior calcaneal bisection and the subtalar joint axis were transferred to the paper.

Using the equations developed by Phillips and Lidtke [58], the data from the first testing session was used to define two points on the subtalar joint axis: P_{i0}, where the subtalar joint axis intersects the plane z = 0 mm, and P_{i30}, where the subtalar joint axis intersects the plane z = 30 mm. Using these points, the angular deviation of the STJ axis from vertical, α_n, and the deviation of the STJ axis from the longitudinal axis of the foot, B_n, were found using equations (1) and (2).

$$\alpha_n = \tan^{-1}\left(\frac{P_{i30}^z - P_{i0}^z}{P_{i30}^x - P_{i0}^x}\right) \qquad (1)$$

$$\beta_n = \sin^{-1}\left(\frac{P_{i30}^y - P_{i0}^y}{\left\| P_{i30} - P_{i0} \right\|}\right) \qquad (2)$$

To begin the second data collection session, subjects performed a standard warm-up of treadmill running at five miles-per-hour for five minutes, followed by stretching. A rigid triad of markers (A, B, and C) was placed on the heel counter of the subject's right shoe, and another rigid triad of markers (D, E, and F) was placed on the posterior right leg. The subject's right leg was then positioned in a rigid fixation device such that the center of the knee was directly above the center of the ankle and the left leg was parallel to the right leg. This was filmed for two second to use as a reference for joint and segment angle calculations [2]. The fixation device was removed prior to data collection.

The footswitch was secured to the plantar surface of the subject's heel using double-sided tape, and an F-Scan sensor was fitted to the subject's shoe. Five experimental conditions included C1- shoe only, C2- unmodified over-the counter orthotic insert, C3- 5° medial post, C4- 10° medial post, and C5- 5° lateral post. The order in which these conditions were performed was randomly chosen for each subject. For each condition, the subject ran on the treadmill for two minutes at 3.8 m s^{-1}. During the first minute, the subject synchronized the footstrikes of his right foot with a metronome set to 80 Hz. During the second minute, ten consecutive footstrikes were recorded. The subject was given five minutes to recover before the next condition began, during which the footwear and markers were changed and a new static reference was filmed.

Data Processing and Analysis

The kinematic data were smoothed in the Vicon workstation using a Woltering filter with a predicted mean standard error of 5 mm^2. All kinematic, footswitch, and plantar pressure data were then imported to Matlab where it was synchronized, trimmed into individual steps, and then processed separately. All C3D files were imported into Matlab using public domain software, ReadC3D written by Alan Morris in 1998 and revised by Jaap Harlaar in 2002 (www.C3D.org).

The methods of Areblad et al.[2] were used to compute lower extremity kinematics (See Appendix A for a complete account of the method). The frontal, sagittal, and transverse plane angles were computed for both the foot and leg separately (see

Nomenclature section above), and plantarflexion/dorsiflection (ζ), eversion/inversion (μ), and abduction/adduction (θ) were computed for the ankle joint. The angular velocities of all of the above angles were calculated using the finite difference method, as were the velocities of the center of pressure in both the medial-lateral and anterior-posterior directions.

The analysis of kinematic variables was limited to four motions: relative frontal plane motion of the calcaneus (relative to the tibia and abbreviated as RE) with 0° being neutral, inversion being positive, and eversion being negative; absolute frontal plane motion of the calcaneus (relative to the fixed room coordinate system, abbreviated as FE) with 90° being neutral, inversion being greater than 90°, and eversion being less than 90°); absolute frontal plane motion of the tibia (abbreviated as LE) with 90° being neutral, varus being greater than 90°, and valgus being less than 90°; and absolute transverse plane motion of the tibia (abbreviated as LR) with 0° being neutral, external rotation being positive, and internal rotation being negative. For a complete list of variables in these four motions, see appendix E.

The trajectory of the center of pressure (COP) was broken up into its medial-lateral (CX) and anterior posterior (CY) directions, with the origin at the most medial-posterior corner of the sensor. Therefore, when CX is increasing the COP was moving laterally, and when CY was increasing the COP was moving anteriorly.

Statistical Analysis

The means and standard deviations were calculated for several events of interest (Appendix E.) A one-way repeated measures analysis of variance (ANOVA) was performed on selected variables and T-tests in post hoc comparisons to find differences between experimental conditions ($p < 0.05$).

Chapter IV

Results

Subject Selection

A total of 14 subjects were screened, and seven subjects (age: 25.1 ± 2.3 yrs, height: 178.0 ± 5.2 cm, body mass: 71.2 ± 4.6 kg) were selected to participate in the study based on the transverse plane orientation of their subtalar joint axis, β_n (16.6° ± 1.6°). Complete details of the anthropometric measurements of ankle complex for the seven experimental subjects are listed in Tables 1 and 2. Full details of all fourteen subjects are given in Appendix D.

Kinematics

Analysis of the kinematics began with heel strike. The initial positions of the ankle (RE_0), calcaneus (FE_0), and tibia (LE_0 and LR_0) for all five experimental conditions are listed in Table 3. The ANOVA did not reveal any significant changes at the heel strike for those variables.

Immediately following heel strike, the calcaneus everted with a peak velocity vFE_1 while the tibia adducted and internally rotated with peak velocities of vLE_1 and vLR_1, respectively. The ANOVA results demonstrated a significant omnibus effect ($F = 3.530$, $P = 0.021$) on the time to peak internal tibia rotation ($tvLR_1$). The post hoc comparison showed that there were significant differences between the neutral orthotic condition and the 10° medial post condition, as well as between the 5° lateral post and 10° medial post conditions (Table 4).

Table 1. Basic Subject Information (Experimental Subjects Only)

Subject	Age (yrs)	Height (cm)	Weight (kg)	Days of Running per Week
1	27	171.0	77.3	3.0
2	23	170.2	65.1	3.0
8	25	180.3	70.5	6.0
9	24	180.3	69.5	4.5
11	22	180.3	74.5	3.5
13	27	182.9	66.4	4.0
14	28	181.0	75.0	4.0
mean	**25.1**	**178.0**	**71.2**	**4.0**
SD	**2.3**	**5.2**	**4.6**	**1.0**

Table 2. Subject Screening Data (Experimental Subjects Only)

Subject	STJ neutral (deg)	Maximum Inversion (deg)	Maximum Eversion (deg)	Relaxed Bipedal Calcaneal angle (deg)	Single Leg Calcaneal angle (deg)	α_n (deg)	β_n (deg)	Forefoot Varus (deg)
1	0.0	24.7	-14.0	-9.0	-13.3	20.0	14.9	4.0
2	0.3	35.0	-15.7	-9.7	-11.0	15.9	16.2	3.0
8	0.3	25.3	-14.3	-12.0	-15.7	24.9	14.0	5.0
9	2.0	44.7	-8.3	-9.0	-9.3	20.8	17.7	16.0
11	1.7	31.0	-10.7	-11.7	-13.7	21.1	18.6	7.0
13	8.0	34.7	-13.3	-5.0	-8.0	23.4	17.3	0.0
14	1.7	39.7	-11.0	-5.7	-10.3	20.3	17.4	8.0
mean	**2.0**	**33.6**	**-12.5**	**-8.9**	**-11.6**	**20.9**	**16.6**	**6.1**
SD	**2.8**	**7.3**	**2.6**	**2.7**	**2.7**	**2.8**	**1.6**	**5.1**

Table 3. Mean (SD) Lower Extremity Angles at Heel Strike.

	Control	Neutral Orthotic	5° Medial Post	10° Medial Post	5° Lateral Post
RE_0	6.4 (6.5)	6.2 (7.5)	5.5 (6.8)	5.0 (7.9)	6.2 (7.1)
FE_0	102.7 (5.8)	103.2 (6.7)	102.5 (5.8)	102.8 (6.4)	101.9 (6.2)
LE_0	96.4 (2.7)	96.6 (3.3)	96.8 (2.8)	97.0 (3.3)	96.4 (3.6)
LR_0	5.5 (5.0)	4.7 (5.8)	4.8 (4.4)	5.2 (5.7)	4.5 (6.1)

Table 4. Mean (SD) Peak Velocities During Early Stance Phase.

	Control	Neutral Orthotic	5° Medial Post	10° Medial Post	5° Lateral Post
VFE_1	-301.0 (117.5)	-280.3 (130.1)	-281.8 (128.8)	-267.3 (141.6)	-291.5 (130.3)
VLE_1	59.9 (38.5)	61.0 (37.7)	59.9 (44.4)	68.1 (38.3)	61.8 (33.1)
VLR_1	-148.0 (56.6)	-144.5 (77.6)	-138.7 (86.9)	-167.6 (71.1)	-143.4 (86.3)
$tvLR_1$	0.038 (0.021)	0.042 (0.015)	0.044 (0.019)	0.049 (0.014)	0.036[d] (0.019)

d- Significantly different from 10° medial post ($F = 10.440$, $P = 0.018$)

After the foot and ankle reached maximum eversion (FE_1 and RE_1, respectively) the tibia externally rotated and abducted with peak velocities vLR_2 and vLE_2, respectively. These values are listed in Tables 5 and 6.

Center of Pressure

The trajectory of the center of pressure (COP) was broken up into its medial-lateral (CX) and anterior posterior (CY) directions, with the origin at the most medial-posterior corner of the sensor. Therefore, when CX is increasing the COP was moving laterally, and when CY was increasing the COP was moving anteriorly.

In the medial-lateral direction, key events were the peak lateral velocity (vCX_1), most lateral position (CX_1), peak medial velocity (vCX_2), and most medial position (CX_2). These values are given in table 7.

Table 5. Mean (SD) Maximum Eversion.

	Control	Neutral Orthotic	5° Medial Post	10° Medial Post	5° Lateral Post
RE_1	-10.3 (3.3)	-9.2 (3.6)	-10.2 (3.0)	-9.9 (3.7)	-10.5 (3.6)
tRE_1	0.132 (0.030)	0.118 (0.020)	0.128 (0.025)	0.123 (0.020)	0.126 (0.027)
FE_1	87.6 (2.3)	88.4 (2.7)	87.6 (2.8)	88.0 (2.7)	87.2 (3.0)
tFE_1	0.132 (0.027)	0.125 (0.022)	0.132 (0.026)	0.139 (0.017)	0.126 (0.028)

Table 6. Mean (SD) Leg External Rotation and Adduction During Late Stance Phase

	Control	Neutral Orthotic	5° Medial Post	10° Medial Post	5° Lateral Post
vLR_2	53.9 (133.5)	118.0 (67.9)	78.2 (113.2)	124.2 (106.3)	81.3 (90.6)
$tvLR_2$	0.189 (0.039)	0.211 (0.012)	0.195 (0.034)	0.203 (0.027)	0.203 (0.032)
vLE_2	-84.3 (61.5)	-94.8 (39.9)	-83.0 (51.9)	-102.6 (61.0)	-79.9 (41.1)
$tvLE_2$	0.216 (0.011)	0.215 (0.017)	0.211 (0.017)	0.215 (0.013)	0.220 (0.014)

Table 7. Mean (SD) Medial-Lateral Center of Pressure Variables

	Control	Neutral Orthotic	5° Medial Post	10° Medial Post	5° Lateral Post
vCX_1	12.7 (10.4)	17.2 (20.6)	21.0 (11.1)	26.8 (12.5)	15.5 (13.0)
CX_1	5.7 (0.4)	5.5 (0.5)	5.5 (0.4)	5.5 (0.3)	5.5 (0.4)
vCX_2	-17.7 (10.0)	-15.3 (5.5)	-20.0 (13.0)	-17.5 (9.2)	-13.4 (7.5)
CX_2	4.5 (0.5)	4.4 (0.4)	4.3 (0.5)	4.4 (0.4)	4.3 (0.5)

In the anterior-posterior direction, key events were the most posterior point (CY_1), the peak anterior velocity (vCY1), and the most anterior point (CY_2). A significant effect was found for CY_2 ($F = 4.481$, $P = 0.005$). These values are listed in table 8.

Table 8. Anterior-Posterior Center of Pressure Variables

	Control	**Neutral Orthotic**	**5° Medial Post**	**10° Medial Post**	**5° Lateral Post**
CY_1	7.3 (2.4)	7.2 (1.8)	6.9 (1.5)	6.5 (1.4)	6.7 (1.8)
vCY_1	232.2 (69.3)	197.6 (33.4)	220.1 (48.7)	211.7 (40.8)	215.6 (44.8)
CY_2	21.8 (1.3)	21.4^{a} (1.2)	21.4 (1.4)	21.3^{a} (1.3)	21.5^{d} (1.5)

a- Significantly different from control ($P < 0.05$)
d- Significantly different from 10° medial post ($P < 0.05$)

Chapter V

Discussion

The purpose of this study was to investigate the effects of different orthotic inserts on kinematics of the lower extremity and the center of pressure (COP) during treadmill running. The following discussion will address this purpose by comparing each orthotic condition to the control (no orthotic) condition. Subject screening data and the limitations of this study will also briefly be discussed. Whenever they are available, relevant results from the literature will be included. However, it is important to remember that the relevance of literature may be limited since most studies have been conducted on over-ground running.

Screening of Subjects

In two recent review articles [16, 17], the authors expressed a need for researchers to better describe their subjects. Since the location and orientation of the subtalar joint axis is believed to have an impact on how an orthotic will affect kinematics [5, 35], it is possible that the results of studies can vary depending on the characteristics of the subjects. For this reason, the subjects in this study were screened using eight anthropometric variables, all of which are reported in Appendix D.

While it is undeniably important that subjects be similar to each other, attempting to normalize too many subject variables can be problematic. Assuming that all eight variables used in this study were normally distributed and independent of one another, the probability of a subject falling within one standard deviation (SD) of the population mean was 66%. Therefore, in order to find 10 subjects who were within one SD of the mean of any single variable, it would likely have been necessary to screen 15 candidates. Normalizing subjects for any two variables would have required the screening of approximately 23 people before finding 10 suitable subjects. (The probability of a subject falling within one SD of the means of two variables is 66% x 66%, or about 44%.) By following the same logic to the extreme case, one could reasonably guess that you would need to perform 278 screenings in order to find 10 who were within one standard deviation of all eight criteria!

For this study the transverse plane angle of subtalar joint axis (β_n) was used as the lone selection criterion. Of the 14 subjects who were screened, nine subjects fell within the desired range of β_n. Two of those nine were unable to participate, one because of an injury, and the other because we did not have the proper lab shoe size to fit him.

Validity of Kinematic Data

The methods used during this study [2] have not been widely used in the literature. However, the results are comparable to the results of other studies done on similar populations. Hamil et al. [9] found that maximum eversion during treadmill running ranged from 8.2° (2.9°) to 14.7° (3.9°) depending on the stiffness of the shoes. The current study found a mean value of 10.3° (3.3°) for maximum eversion in the control condition. McClay and Manal [7] found a maximum eversion of –11.2° (2.7°) and a range of tibia internal rotation of 8.9° (2.6°) in normal subjects during treadmill running at 3.35 m s^{-1}. The current study found a range of 9.1° (4.1°) for tibia internal rotation during the control condition. Similar values for tibia internal rotation during running are also reported in Nawoczenski et al. [14].

Neutral Orthotics

Based on the results of several previous studies [3, 21, 31, 39], it was hypothesized that the use of an unmodified non-cast over-the-counter orthotic would not significantly alter the kinematics or the COP compared to the control (shoe only). This appears to be the case for the kinematic variables for this study, since the ANOVA did not show any significant differences between these two conditions. The neutral orthotic did have a significant effect on the most anterior point of the COP path, CY_2, when compared to the control condition. This effect, however, may be due to the difference in heel height between the different conditions, as shown in Figure 1.

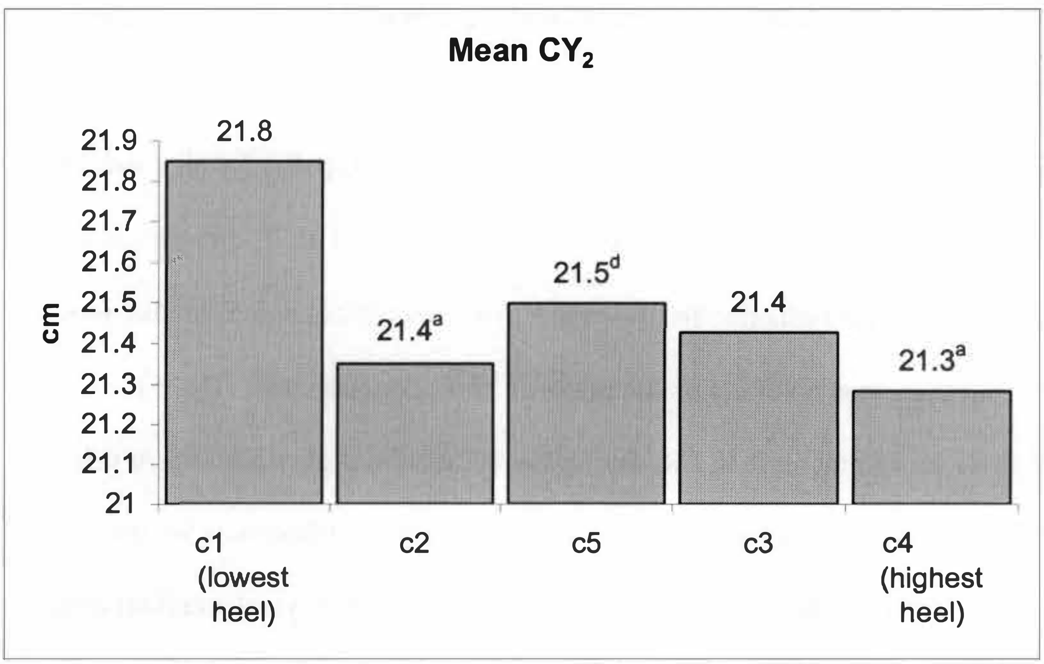

Figure 1: Most anterior point of COP trajectory.

a- Significantly different from c1 ($P < 0.05$)

d- Significantly different from c4 ($P < 0.05$)

Medially Posted Orthotics

The traditional view was that subtalar joint pronation was primarily a frontal plane event. Orthotic studies primarily focused on calcaneal eversion and/or tibiocalcaneal eversion. The data from this study suggests that medially posted orthotics did not alter maximum tibiocalcaneal eversion, RE_1 (Fig. 2), but there was a downward trend in the peak velocity of tibiocalcaneal eversion, vRE_1 (Fig. 3).

With three-dimensional motion analysis becoming more common, researchers are focusing more on the transverse plane component of subtalar joint motion, primarily through tibia rotation [15, 37-39]. Nawoczenski et al. [14] characterized subtalar pronation as a combination of tibia internal rotation and tibia adduction, which is similar to calcaneal eversion. Using custom neutral-cast orthotics, they found a consistent, although statistically insignificant, decrease in tibia internal rotation of 2.1° when running in orthotics.

Figure 4 shows the mean range of internal tibia rotation (LR_1) for all conditions from the current study. Similar to the results of Nawoczenski et al. [14], there was a statistically insignificant reduction in LR_1 of 0.9° from the control and 5° medial post conditions, but there was no effect on the range of tibia adduction, LE_1 (Fig. 5). However, there was an increase in LR_1 during the 10° medial post conditions of 0.1° versus the control condition (Fig. 4). There are two possible explanations for this: the changes in LR_1 are random and orthotics have no effect, or that beyond a certain amount of medial posting additional posting causes an increase in LR_1. There is insufficient data both in this study and in the literature to critically evaluate the second possibility.

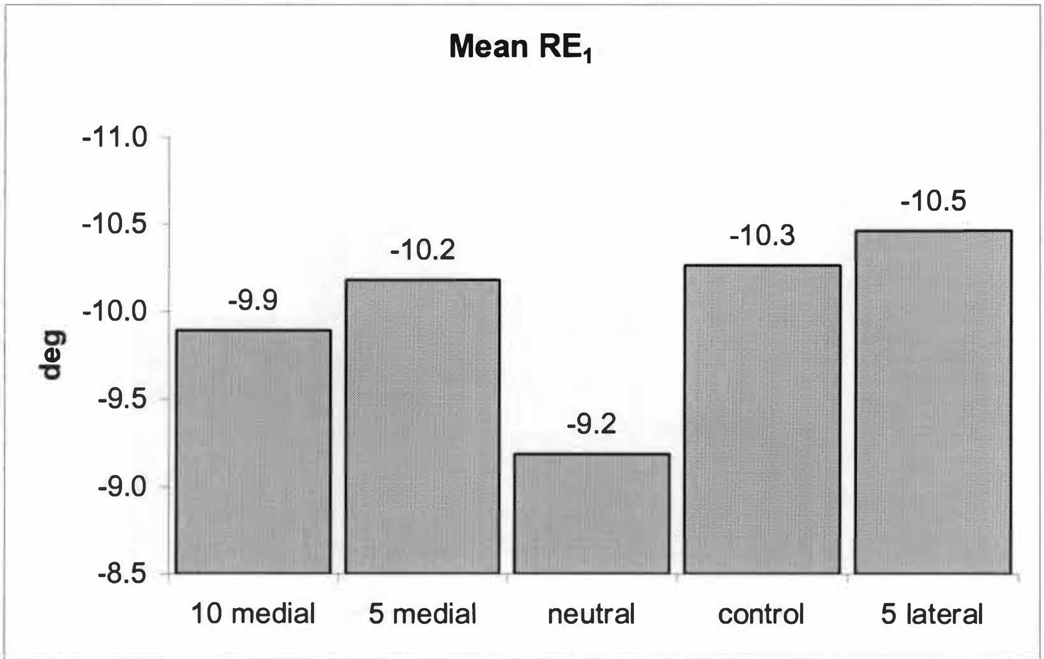

Figure 2: Mean maximum tibiocalcaneal eversion.

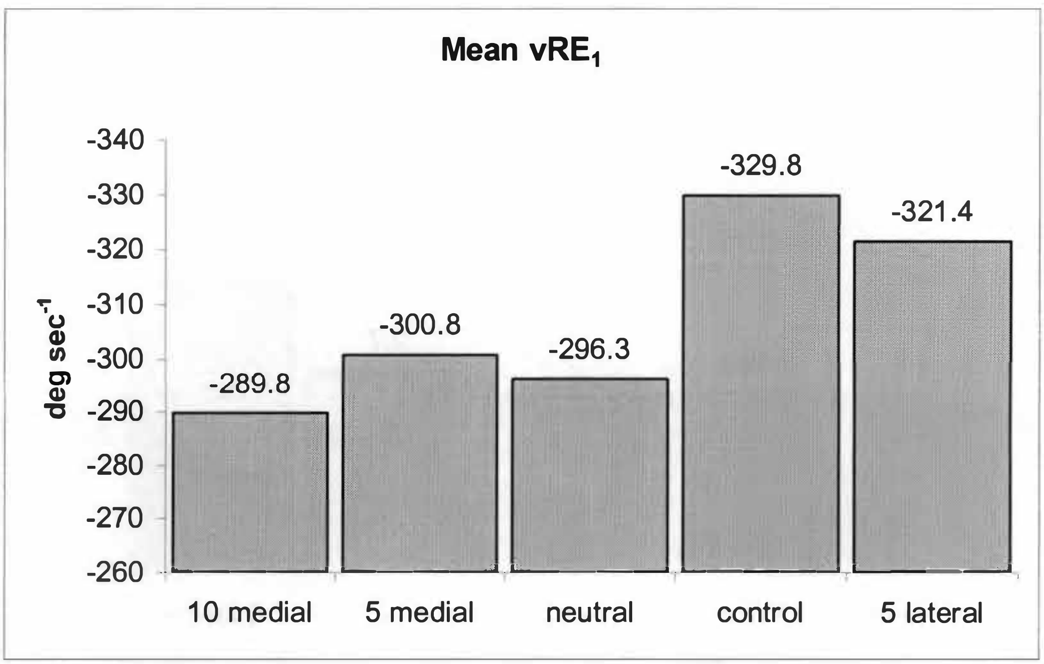

Figure 3: Mean peak velocity of tibiocalcaneal eversion.

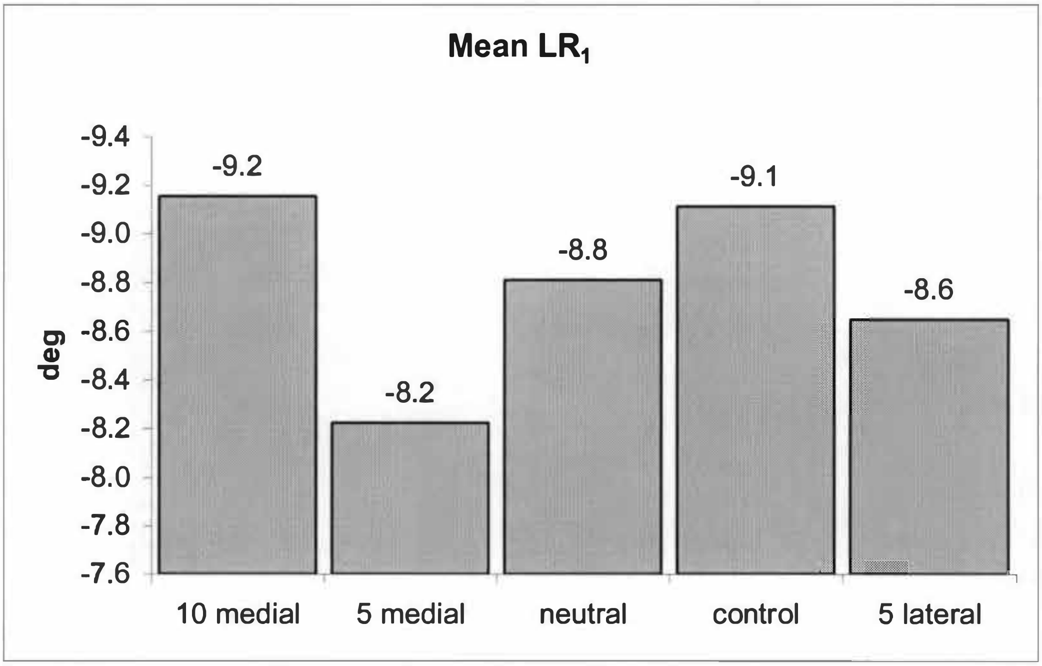

Figure 4: Mean range of tibia internal rotation.

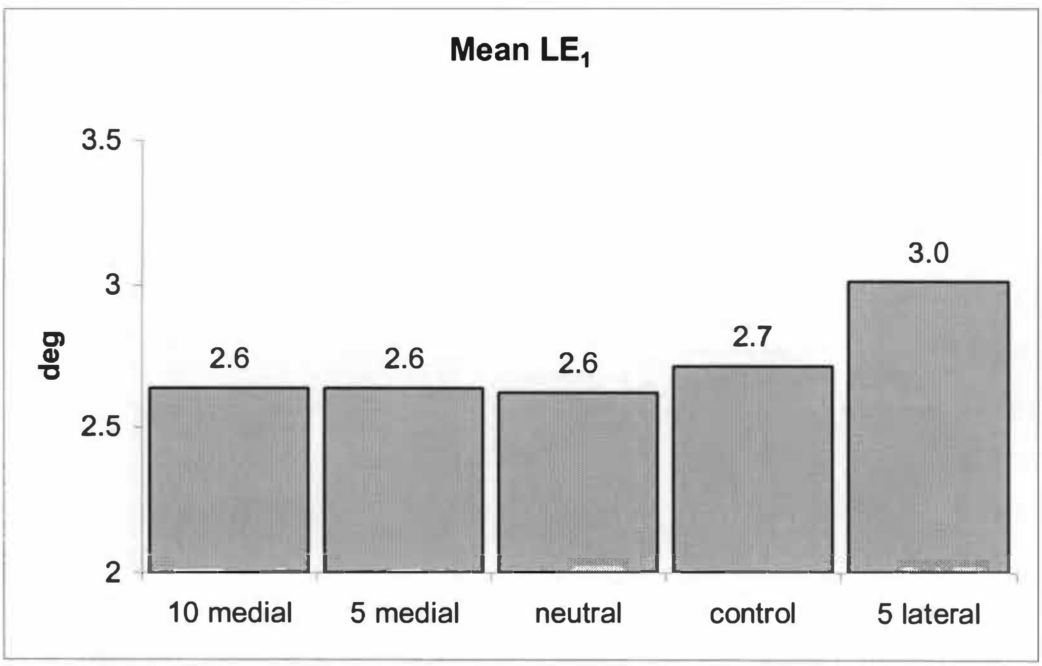

Figure 5: Mean range of tibia adduction.

Medially posted orthotics consistently and systematically increased the medial-lateral peak velocities of the COP (Fig. 6 and Fig. 7). Increased peak velocities have generally been regarded as indicative of instability in the literature. However, the kinematic data from this study does not support this conclusion.

Laterally Posted Orthotics

While there was one statistically significant effect for the laterally posted condition, it generally tended to have the same effect as the control condition. Nester et al. [15] found that during treadmill walking, anti-supinatory orthotics significantly increased the range of initial internal tibia rotation. During the treadmill running of this study, the lateral post did not show that effect. The only statistically significant kinematic difference was in the time to the peak velocity of internal tibia rotation ($tvLR_1$), indicating that the period of subtalar joint pronation may have lasted longer.

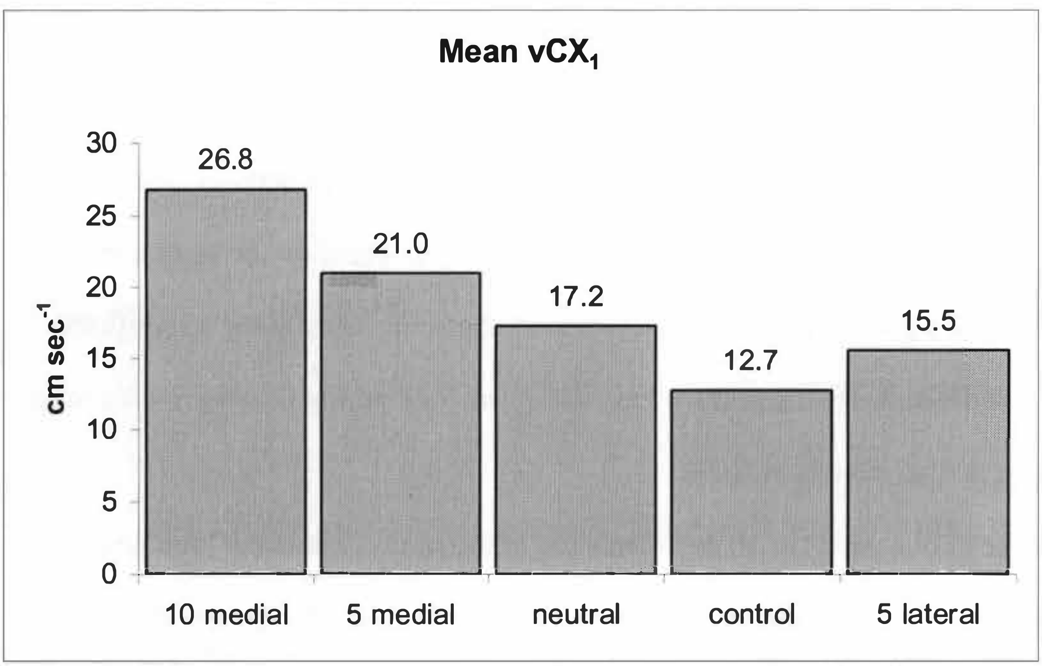

Figure 6: Peak lateral velocity of the COP during the first 50% of stance phase.

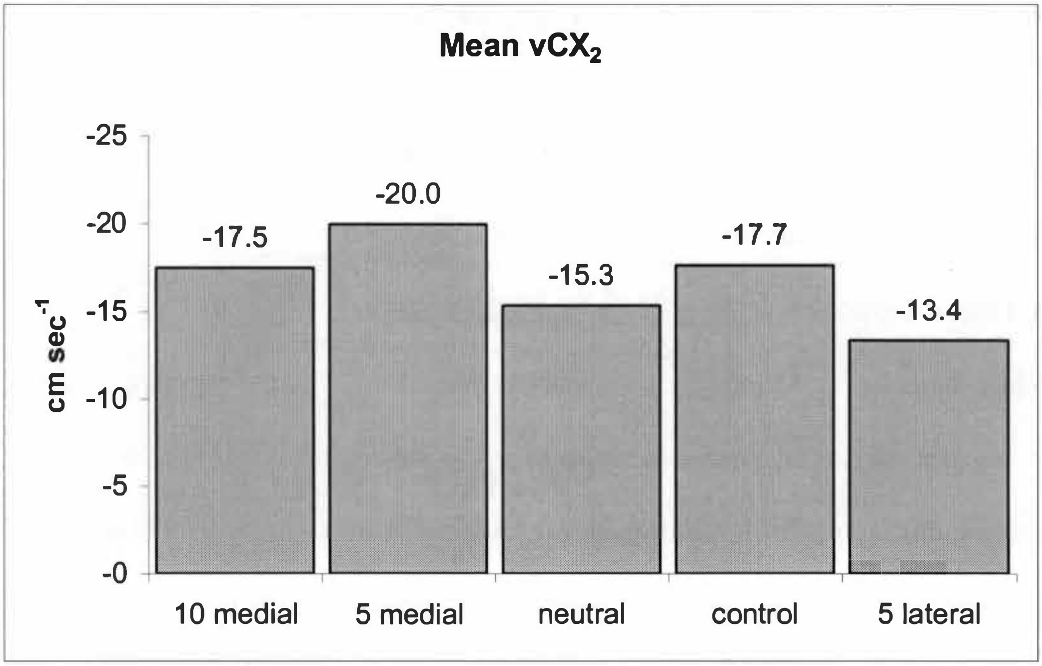

Figure 7: Peak medial velocity of the COP during the last 50% of stance phase.

The Relationship Between Frontal Plane and Transverse Plane Motion

The true purpose of the subtalar joint is believed to be to convert vertical ground reaction force into transverse plane rotation of the leg [48, 59]. Nigg et al. [35] used the ratio of tibiocalcaneal eversion to tibia internal rotation, called the transfer coefficient, to quantify the relationship between frontal plane motion and transverse plane motion. Nawoczenski et al. [14] computed the ratio of leg adduction to leg rotation, based on the assumptions that leg adduction is a good indicator of tibiocalcaneal eversion. They found a significant increase in the transfer coefficient with the use of orthotics during running, owing mainly to reduction in tibia internal rotation.

In the current study, both the orientation of the subject's subtalar joint and computed complete kinematics of the foot and leg were measured. The transfer coefficient, as calculated by Nawoczenski et al. [14] is equivalent to the ratio of LE_1 to LR_1.

Plotting the transfer coefficient from the control condition against the sagittal plane orientation of the subtalar joint, α_n, can lend some insight into how these variables are related (Fig. 8). While Figure 8 does support the hypothesis that the smaller α_n is the greater the LE_1 to LR_1 ratio will be, a definitive conclusion cannot be made from only seven subjects.

Evaluation of the Study

The major limitations of this study were the number of subjects. Having a sufficient number of subjects is critical to finding significant differences if they exist, and it is also essential to being confident that significant differences do not exist. Given more subjects, there may have been a better chance to find significant differences between the 5° and 10° medially posted conditions, as well as between the medial posted conditions and the control.

There was an intentional trade-off made in using the treadmill and the F-Scan instead of over-ground running and the force platform. The force platform would have undoubtedly improved the accuracy of the COP measurements. However, using the treadmill made it possible to control the running speed exactly, and also made it much simpler to control the stride frequency. The treadmill also made it possible to record several consecutive steps, allowing data to be collected more efficiently.

Given these results, as well as the current literature, further studies in this area are warranted. There are currently no studies in the literature done to compare several different amounts of medial heel posting during controlled walking or running trials. The results could determine if there is an amount of medial posting beyond which subtalar joint pronation increases. Another potential future study would be to further explore the

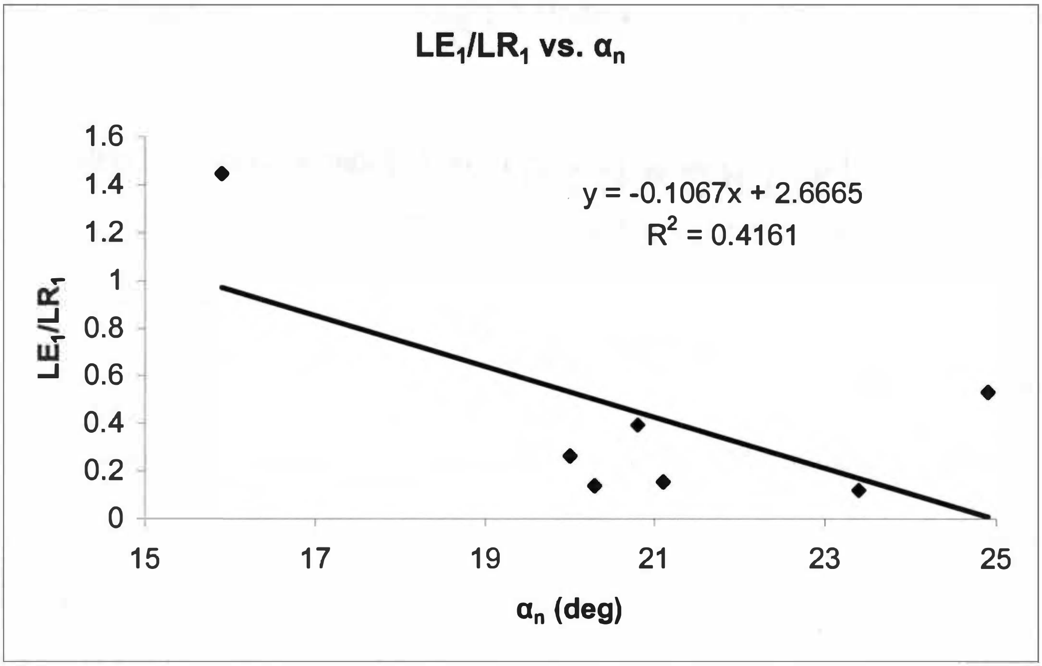

Figure 8: The relationship between the transfer coefficient and α_n.

relationship between subtalar joint orientation and the amounts of motion both in the transverse and frontal planes. Many have speculated that a lower sagittal plane orientation of the subtalar joint axis would result in less motion in the transverse plane [48, 59], and Nawoczenski et al. [14] computed the ratio of leg adduction to leg rotation. Both the orientation of the subject's subtalar joint and the angular kinematics of the foot and leg were measured in the current study. However, there were an insufficient number of subjects to draw any conclusions about the relationship between these variables.

Summary

Although not statistically significant, there were several trends in the kinematic data. The orthotics had no effect on tibiocalcaneal eversion or leg adduction. There was a reduction in the velocity of tibiocalcaneal eversion with both the neutral and medially posted orthotics versus the control. The 5° medially posted orthotic reduced leg internal rotation, but the 10° medially posted orthotic did not. The laterally posted orthotic condition did not have a significant impact on lower extremity kinematics during treadmill running as it did in the treadmill walking study performed by Nester et al. [15].

The orthotic conditions in this study produced a significantly systematic effect on vCX_1 and vCX_2, with medial orthotics increasing these velocities, and lateral orthotics decreasing them. It is unclear what these changes mean to the function of the lower extremity.

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Appendix A

Summary of "Three-dimensional measurement of rearfoot motion during running."[2]

Areblad et al.[2] describes one of the simplest methods of measuring three-dimensional motions. This method is only able to provide angular kinematics, but its simplicity makes it easy to apply to almost any body segment. This appendix will summarize the methodology used by Areblad et al.[2] to determine the 3-D angular kinematics from a series of marker triads, using the foot and leg segments as an example.

Notation

$\hat{X}$, $\hat{Y}$, $\hat{Z}$: Orthogonal unit vectors that create the fixed room coordinate system (RCS)

$\hat{i}$, $\hat{j}$, $\hat{k}$: Orthogonal unit vectors that create the coordinate system of a segment. The segment will be denoted by subscripts.

$A \cdot B$: The scalar, or dot, product of vectors A and B. The result of this operation is always a scalar. $A \cdot B = \|A\|\|B\|\cos(\theta_{AB})$ where θ_{AB} is the angle between vectors A and B. This relationship allows us to easily compute the angle between two vectors.

$A \times B$: The vector, or cross, product of A and B. The result of this operation is always a vector that is orthogonal to both A and B.

$\|A\|$: The norm of vector A. The norm is the scalar length of the vector.

Marker Triads

A marker triad is a set of three markers that are placed on a body segment so that they are non-collinear, and they do not move relative to each other. Often it is best to adhere the three markers onto a rigid triangle, and than fasten the triangle onto the body segment. The three markers will be used to define a segment coordinate system. For the purpose of this example, markers A, B, and C will constitute the marker triad on the right heel, and markers D, E, and F will constitute the marker triad on the right leg (Fig. 9).

Determining a Provisional Coordinate System from a Marker Triad

Three non-collinear points can be used to define a plane, and a 3-D coordinate system can be defined from the two axes of the plane and the cross product of those two axes. In the current example, equations 1, 2, and 3 can be used to determine the provisional coordinate system of the foot (PCF).

$$\hat{i}'_f = \frac{(B-A)}{\|B-A\|} \tag{1}$$

$$\hat{j}'_f = \frac{(C-A)\times\hat{i}'_f}{\|(C-A)\times\hat{i}'_f\|} \tag{2}$$

$$\hat{k}'_f = \hat{i}'_f \times \hat{j}'_f \tag{3}$$

$$[PCF] = \left[\hat{i}'_f, \hat{j}'_f, \hat{k}'_f\right] \tag{4}$$

Transformation Matrices

By definition, when a segment is in its neutral position, the three angles defining that segment's orientation should all equal 0. Since all segmental angles are defined relative to the RCS, when a segment is in neutral the segment's coordinate system should be equal to the RCS. Since this rarely occurs in practice, data collection should always begin by filming a static trial of the subject in neutral position. A transformation matrix can be determined from the static trail. When the provisional coordinate system is multiplied by the provisional coordinate system, the result is a segment coordinate system that is equal to the RCS when the segment is in its neutral position.

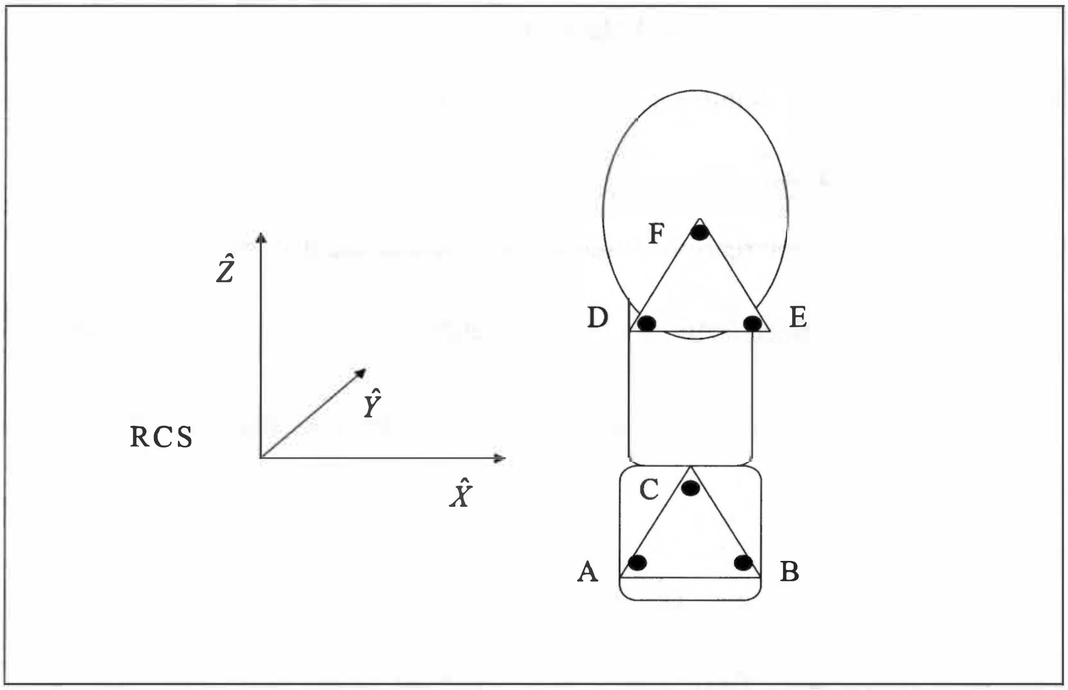

Figure 9: The placement of marker triads on the posterior heel and leg.

From the foot example, assuming that PCF has already been computed from the static trial, the transformation matrix of the foot (TMF) can now be found using equation 5.

$$[TMF] = [PCF]^{-1} * [RCS] \tag{5}$$

For all subsequent trials, a true foot coordinate system (FCS) can now be found but first computing PCF, and then using equation 6.

$$[FCS] = [PCF] * [TMF] \tag{6}$$

Computing the Absolute Angles of a Segment

After equations 1-6 have been applied to a segment, trigonomic functions can be used to determine the orientation of the segment relative to the RCS. These equations can be set up any number of ways, depending on the desired outcome. The equations used to determine the absolute angles of the foot are provided in equations 7, 8, and 9, with an explanation of each equation.

Equation 7 provides the rearfoot angle, or the angle of the heel in the frontal plane. Rearfoot angle is defined as the angle of $\hat{k}_f$ relative to the $\hat{X}\hat{Y}$ plane measured in the $\hat{i}_f\hat{k}_f$ plane. The result of $(\hat{Z} \times \hat{j}_f)$ is a vector that is in the $\hat{i}_f\hat{k}_f$ plane and parallel to the $\hat{X}\hat{Y}$ plane.

$$\text{Rearfoot Angle} = \cos^{-1}(\hat{k}_f \cdot (\hat{Z} \times \hat{j}_f)) \qquad (7)$$

Equation 8 provides the flexion angle of the rearfoot, or the angle of the heel in the sagittal plane. This angle is defined as the angle of $\hat{k}_f$ relative to the $\hat{X}\hat{Y}$ plane measured in the $\hat{j}_f\hat{k}_f$ plane. The result of $(\hat{i}_f \times \hat{Z})$ is a vector that is in the $\hat{j}_f\hat{k}_f$ plane and parallel to the $\hat{X}\hat{Y}$ plane.

$$\text{Flexion Angle} = \cos^{-1}(\hat{k}_f \cdot (\hat{i}_f \times \hat{Z})) \qquad (8)$$

Equation 9 provides the abduction angle of the rearfoot, or the angle of the heel in the transverse plane. This angle is defined as the angle of $\hat{j}_f$ relative to the $\hat{Y}\hat{Z}$ plane measured in the $\hat{j}_f\hat{X}$ plane. Since $\hat{X}$ is always perpendicular to the $\hat{Y}\hat{Z}$ plane, the result of $\cos^{-1}(\hat{j}_f \cdot \hat{X})$ is subtracted from 90°.

$$\text{Abduction Angle} = 90° - \cos^{-1}(\hat{j}_f \cdot \hat{X}) \qquad (9)$$

Computing Relative Angles Between the Foot and Leg

The coordinate system of the leg (LCS) can also be computed using the same methods as for the FCS. Once both coordinate systems have been determined, equations 10, 11, and 12 can be used to determine the relative angles between the two segments. As with the absolute angles, there are several different ways the relative angles can be computed based on the desired axis of rotation and which direction of positive motion.

For this example, dorsi/plantar flexion is defined as rotation about $\hat{i}_l$ with dorsiflexion being positive. The result of $(\hat{k}_f \times \hat{i}_l)$ is a vector that is perpendicular to the $\hat{i}_l\hat{k}_f$ plane.

$$\text{Dorsi/Plantar Flexion} = 90° - \cos^{-1}(\hat{k}_l \cdot (\hat{k}_f \times \hat{i}_l)) \qquad (10)$$

Eversion/inversion is defined as rotation about $(\hat{i}_l \times \hat{k}_f)$ with inversion being positive. Since $(\hat{i}_l \times \hat{k}_f)$ is the axis of rotation, $\hat{k}_f$ and $\hat{i}_l$ are all that is needed to 5determine the angle of eversion or inversion.

$$\text{Eversion/Inversion} = 90° - \cos^{-1}(\hat{k}_f \cdot \hat{i}_l) \qquad (11)$$

Internal/external rotation is defined as rotation about $\hat{k}_f$ with external rotation being positive. This is similar to dorsi/plantar flexion, except this is the angle of $\hat{i}_f$ relative to the $\hat{i}_l\hat{k}_f$ plane.

$$\text{Internal/External Rotation} = 90° - \cos^{-1}(\hat{i}_f \cdot (\hat{k}_f \times \hat{i}_l)) \qquad (12)$$

Summary

The methods of Areblad et al. [2] can easily be applied to any number of segments using variations on the above equations. There are four basic steps to this method: (1) calculate the transformation matrices from a static trial of the subject in the neutral position using equations 1-5, (2) compute the coordinate system of each segment frame by frame using equations 1-4 and 6, (3) compute the absolute angles of each segment using variations of equations 7-9, and (4) compute the relative angles between any two segments using variations of equations 10-12.

Appendix B

Summary of "Clinical determination of the linear equation for the subtalar joint axis"[58]

There have been several methods proposed for determining the orientation of the subtalar joint (STJ) axis in subjects without resorting to radiography. Kirby [49] describes a method he calls the palpation method, in which the plantar surface of the foot is palpated in order to find points that do not cause rotation. These points are assumed to be on the STJ axis. The palpation method produces a 2-D representation of the STJ axis. Morris and Jones [57] proposed a method to measure the three-dimensional orientation of the STJ axis by drawing a group of dots on the skin near the two ends of the axis. When the foot is moved through its range of motion about the STJ, the dots that do not move will represent the ends of the STJ axis.

Phillips and Lidtke [58] developed a method for determining and equation for the STJ axis relative to anatomical landmarks that can be found on any subject. While this method of determining the position of the STJ axis is not as direct as Morris and Jones' method [57], the measurements involved are straightforward, and although the resulting equation represents the STJ in a static situation, it is potentially a key stepping stone to representing the STJ dynamically.

The purpose of this appendix is to summarize the method of Phillips and Lidtke [58]. Much of the analysis will be left out of this summary to conserve space.

The Coordinate System

The origin of the coordinate system used in this method is the most inferior posterior aspect of the calcaneus, henceforth referred to as P_0. The x-axis runs from P_0 along the bisection of the plantar heel. Directly vertical from P_0 defines the z-axis, and the y-axis is the cross product of the z and x axes. Therefore, all points anterior, medial, and superior of P_0 are positive, and all points posterior, lateral, and inferior of P_0 are negative.

Measurements

There are 9 measurements in all that need to be taken. Eight of these can be directly measured with a ruler or an ergometer. In the frontal plane, these include the angle of the calcaneus relative to the tibia in STJ neutral (F_n), maximum supination (F_s), and maximum pronation (F_p). The linear displacement of P_0 along the y-axis must also be measured. This can be accomplished by recording the medial-lateral position of P_0 in STJ neutral (y_0), maximum supination (y_1), and maximum pronation (y_2), and then computing the linear displacement during supination as $y_s = y_1 - y_0$, and the linear displacement during pronation as $y_p = y_2 - y_0$. In the transverse plane, the angle of adduction of the x-axis during maximum supination (T_s) and the angle of abduction during maximum pronation (T_p) need to be measured.

Before the final measurement can be taken, the Kirby's palpation technique [49] must be performed on the subject. With the foot in neutral, begin palpating at the posterior heel until a point is found that causes no rotation about the STJ axis. Mark that point with a felt marker, and then palpated 1 to 2 cm distally on the plantar foot until another point is found. Continue this until 6 points have been marked, and then draw the best line between the 6 points. This line, along with point P_0, the x-axis, and an outline of the neutral foot can be transferred to a sheet of paper representing the plane z = 0. The intersection of STJ axis with the x-axis is x_a, and the intersection of the STJ axis with the y-axis is y_a. The angle of the STJ axis from the sagittal plane is $\alpha = \tan^{-1}(\frac{y_a}{x_a})$, which is the final measurement.

Calculation of the Equation of the STJ Axis

In addition to analytical equation solving, the errors in the angular and linear measurements must be estimated in order to determine an accurate equation for the STJ axis. While most of the following equations are written to produce the desired outcome, the error estimation equations need to be solved, and this is no small feat. In fact, it is far easier to estimate the solution using a search algorithm than to solve these equations directly, and that is the approach that was used in the current study.

To begin, the equation for the line representing the STJ axis in the plane z = 0 is given in equation 1. When the foot moves about the STJ axis, the anatomical landmarks that define the coordinate system move relative the STJ axis. Equation 2 represents the STJ axis in the plane z = 0 when the foot is fully supinated, and equation 3 represents the STJ axis when the foot is fully pronated.

$$y = x\tan(\alpha) + y_a \qquad (1)$$

$$y = x\tan(\alpha - T_s) + (y_a - y_s) \qquad (2)$$

$$y = x\tan(\alpha - T_p) + (y_a - y_p) \qquad (3)$$

These three lines intersect at three points: the intersection of the neutral axis with the supinated axis, $P_{ns} = (x_{ns}, y_{ns}, 0)$; the intersection of the neutral axis with the pronated axis, $P_{np} = (x_{np}, y_{np}, 0)$; and the intersection of the supinated axis with the pronated axis, $P_{sp} = (x_{sp}, y_{sp}, 0)$. Equations 4 – 9 define these three points of intersection, determined by rewriting equations 1-3.

$$x_{ns} = \frac{y_s}{\tan(\alpha - T_s) - \tan\alpha} \tag{4}$$

$$y_{ns} = \frac{y_s \tan\alpha}{\tan(\alpha - T_s) - \tan\alpha} + y_a \tag{5}$$

$$x_{np} = \frac{y_p}{\tan(\alpha - T_p) - \tan\alpha} \tag{6}$$

$$y_{np} = \frac{y_p \tan\alpha}{\tan(\alpha - T_p) - \tan\alpha} + y_a \tag{7}$$

$$x_{sp} = \frac{(y_s - y_p)}{\tan(\alpha - T_s) - \tan(\alpha - T_p)} \tag{8}$$

$$y_{sp} = \frac{(y_s - y_p)\tan(\alpha - T_s)}{\tan(\alpha - T_s) - \tan(\alpha - T_p)} + (y_a - y_s) \tag{9}$$

If the STJ axis remained fixed in space while the foot was moved about it (a key assumption of this method), then the three lines in equations 1 – 3 should intersect a single point. They typically do not, however, due to errors in the measurement of T_s, T_p, y_s, and y_p. If point P_{sp} were moved so that it was on the line given by equation 1, that would provide a good estimate of $P_{i0} = (x_{i0}, y_{i0}, 0)$, the true point where the STJ axis crosses the plane z = 0. This estimate is given in equations 10 and 11.

$$x_{i0} = \frac{x_{sp} + (y_{sp} - y_a)\tan(\alpha)}{1 + \tan^2 \alpha} \quad (10)$$

$$y_{i0} = x_{i0} \tan \alpha + y_a \quad (11)$$

Let q_1, q_2, r_1, and r_2 represent the errors in T_s, T_p, y_s, and y_p, respectively. By adding in the error terms and setting equations 4 and 6 equal to x_{i0}, the relationships between q_1 and r_1, and q_2 and r_2 can be found. These are given in equations 12 and 13.

$$r_1 = f(q_1) = x_{i0}[\tan(\alpha - T_s - q_1) - \tan \alpha] - y_s \quad (12)$$

$$r_2 = f(q_2) = x_{i0}[\tan(\alpha - T_p - q_2) - \tan \alpha] - y_p \quad (13)$$

There are an infinite number of solutions to equations 12 and 13, but it can be safely assumed that the best solutions are the ones that minimize q_1, q_2, r_1, and r_2. Without getting into the details of the derivation, the optimal solutions of equations 12 and 13 occur when equations 14 and 15 are true.

$$q_1 = -f(q_1)f'(q_1) \quad (14)$$

$$q_2 = -f(q_2)f'(q_2) \quad (15)$$

Once the optimal values of q_1, q_2, r_1, and r_2 have been found, P_{i0} can be found using equations 16 and 11.

$$x_{i0} = \frac{y_s + r_1}{\tan(\alpha - T_s + q_1) - \tan \alpha} = \frac{y_p + r_2}{\tan(\alpha - T_p + q_2) - \tan \alpha} \quad (16)$$

In order to define an equation for the STJ axis, two points on the axis must be found. P_{i0} represents the point where the axis intersects the plane $z = 0$. For the second point, Phillips and Lidtke [58] arbitrarily chose to find point P_{i30}, where the STJ axis intersects the plane $z = 30$. The steps to accomplishing this are similar to finding P_{i0}.

The first step is to find the equivalent values of y_s and y_p in the plane z = 30, y_{30s} and y_{30p}. These are given in equations 17 and 18.

$$y_{30s} = (y_s + r_1) + 30\sin F_n - 30\sin F_s \tag{17}$$

$$y_{30p} = (y_p + r_2) + 30\sin F_n - 30\sin F_p \tag{18}$$

Once y_{30s} and y_{30p} have been found, equations for the three lines representing the STJ axis in STJ neutral, maximum supination, and maximum pronation can be found, given in equations 19, 20, and 21, respectively.

$$y = x\tan(\alpha) + y_a \tag{19}$$

$$y = x\tan(\alpha - T_s - q_1) + (y_a - y_{30s}) \tag{20}$$

$$y = x\tan(\alpha - T_p - q_2) + (y_a - y_{30p}) \tag{21}$$

From equations 19 – 21, the three points of intersection, $P_{30ns} = (x_{30ns}, y_{30ns}, 30)$, $P_{30np} = (x_{30np}, y_{30np}, 30)$, and $P_{30sp} = (x_{30sp}, y_{30sp}, 30)$ can be found using equations 22 – 27.

$$x_{30ns} = \frac{y_{30s}}{\tan(\alpha - T_s - q_1) - \tan\alpha} \tag{22}$$

$$y_{30ns} = \frac{y_{30s}\tan\alpha}{\tan(\alpha - T_s - q_1) - \tan\alpha} + y_a \tag{23}$$

$$x_{30np} = \frac{y_{30p}}{\tan(\alpha - T_p - q_2) - \tan\alpha} \tag{24}$$

$$y_{30np} = \frac{y_{30p}\tan\alpha}{\tan(\alpha - T_p - q_2) - \tan\alpha} + y_a \tag{25}$$

$$x_{30sp} = \frac{(y_{30s} - y_{30p})}{\tan(\alpha - T_s - q_1) - \tan(\alpha - T_p - q_2)} \tag{26}$$

$$y_{30sp} = \frac{(y_{30s} - y_{30p})\tan(\alpha - T_s - q_1)}{\tan(\alpha - T_s - q_1) - \tan(\alpha - T_p - q_2)} + (y_a - y_{30s}) \qquad (27)$$

As was the case in the plane z = 0, if all of the measurements were errorless then P_{30ns}, P_{30np}, and P_{30sp} would be equal. Since the errors in T_s, T_p, y_s, and y_p have all been estimated, the errors in the plane z = 30 can be attributed to errors in F_s and F_p, which will be denoted as m_1 and m_2, respectively. By setting P_{30ns} equal to P_{30np} and adding in the error terms, we can express m_2 as a function of m_1. This is given in equation 28.

$$m_2 = f(m_1) = -F_p + \sin^{-1}\left\{ \begin{array}{l} \sin F_n + \dfrac{y_p + r_2}{30} - \\ \dfrac{y_s + r_1 + 30\sin(F_s + m_1)}{30[\tan(\alpha - T_p - q_1) - \tan\alpha]}[\tan(\alpha - T_p - q_2) - \tan\alpha] \end{array} \right\} \qquad (28)$$

The optimal solution of equation 28 is given in equation 29.

$$m_1 = -f(m_1)f'(m_1) \qquad (29)$$

Once m_1 and m_2 are found, y_{30s} and y_{30p} can be recalculated, and P_{i30}, the intersection of STJ axis with the plane z = 30, can be found using equations 22 and 23.

Once two points on the STJ axis have been established, a simple linear equation can be used to represent all points that fall on the axis, given in equation 30.

$$\frac{x - x_{i0}}{A} = \frac{y - y_{i0}}{B} = \frac{z - z_{i0}}{C} \qquad (30)$$

A, B, and C are the direction cosines of the STJ axis, and are determined from the points P_{i0} and P_{i30}.

$$A = (x_{i0} - x_{i30})\sqrt{(x_{i0} - x_{i30})^2 + (y_{i0} - y_{i30})^2 + (z_{i0} - z_{i30})^2} \qquad (31)$$

$$B = (y_{i0} - y_{i30})\sqrt{(x_{i0} - x_{i30})^2 + (y_{i0} - y_{i30})^2 + (z_{i0} - z_{i30})^2} \qquad (32)$$

$$C = (z_{i0} - z_{i30})\sqrt{(x_{i0} - x_{i30})^2 + (y_{i0} - y_{i30})^2 + (z_{i0} - z_{i30})^2} \quad (33)$$

Summary

The major drawback of this method is that it relies on so many measurements, increasing the vulnerability to measurement error. The possibility of error, however, is directly addressed in the calculation of the equation for the STJ axis, hopefully reducing the effects of measurement errors. The resulting equation provides more information than just the orientation of the STJ axis. The mediolateral deviation of the STJ axis could also be estimated from the equation, which, according to Kirby[5,6], could have a greater impact on function that changes in the orientation.

Appendix C

Informed Consent Form

Investigator: Michael Wortley
Address: Exercise Science and Sport Management
The University of Tennessee
1914 Andy Holt Avenue
Knoxville, TN 37996
Phone: (865) 386-7283

You are invited to participate in a research study on running shoes entitled, "The effect of a shoe-insert on the plantar pressure distribution and lower extremity kinematics during treadmill running" which examines changes in biomechanical measurements with the use of shoe-inserts.

You are aware that you should be a healthy male recreational runner, and have no major injuries to your lower extremity within the past year. If you are qualified and decide to participate, you will be asked to complete these tasks: 1) attend one screening session, and 2) attend one test session. The screening session is to measure the arch index of your foot, and to determine the position of your subtalar joint through passive measurements of your lower extremity. During the test session, you will run on a treadmill at 7 minutes-per-mile pace for three trials of 2-minutes duration each.

Please wear loose shorts, a comfortable short-sleeved shirt or tank top when you report to the lab for the testing sessions. The test session will take approximately about 30 minutes. You will begin with a standard warm-up by using a stationary bike for 5 minutes and stretching. You will perform one bout (2 minutes) of level running at a required speed (3.8 m/s) on the treadmill, in each of five testing conditions: in shoes, in shoes with an over-the-counter orthotic insole, in shoes with an orthotic insole with a lateral post, in shoes with an orthotic insole with a light medial post, and in shoes with an orthotic insert with a heavy medial post. During the test, biomechanics instruments will be used to make measurements. Some of these instruments will be placed/fixed on your body. None of the instruments will impede your ability to engage in normal and effective motions during the test. If you have any further questions, interests or concerns about any instrumentation, please feel free to contact the investigator.

The potential risks include an ankle sprain from foot contact in an unbalanced fashion and muscular strain in lower extremity. Every effort will be made to reduce these risks through proper warm-up, sufficient practice before the test, and use of spotters. You will be encouraged to warm-up actively prior to each testing session so that you feel physically prepared to perform effectively and thus minimize any chance for injury. All tests will be conducted and the equipment will be handled by the qualified research personnel in the Biomechanics/Sports Medicine Lab, who will sign a confidentiality statement. The Biomechanics/Sports Medicine Lab has tested more than 200 subjects in various research projects involving dynamic activities such as jumping, landing, and running in several research projects over the past seven years. None of them were injured in any fashion during the test sessions.

Should any injury occur during the course of testing, standard first aid procedures would be administered as necessary. At least one researcher with a basic knowledge of athletic training and/or first aid procedures will be present at each test session. In the event of physical injury is suffered as a result of participation in this study, the University of Tennessee does not automatically provide reimbursement for medical care or other compensation.

Your benefits include assessment of your performance and biomechanics of running. You are welcome to make an appointment to review the data from your tests. In addition, if you wish to have a copy of the results of the study, please let me know.

Your participation is entirely voluntary and your decision whether or not to participate will involve no penalty or loss of benefits to which you are otherwise entitled. Your identity as a subject will be held in strict confidence and any description of your data will be referred to by a subject number only. Any information that are obtained in connection with this study and that can be identified with you will remain confidential and will be disclosed only with your permission.

Once you have read this informed consent form and all of your questions have been answered, you are required to sign and date the form below and the attached form that lists individual subject requirements. Your signature indicates that you have read and understand the information provided above, that you willingly agree to participate, that you may withdraw your consent at any time and discontinue participation at any time without penalty or loss of benefits to which you are otherwise entitled.

Subject Name: Signature: Date:

_________________________ _________________________________

Investigator: Date:

_________________________ ___________________

Appendix D

Subject Information

Table 9. Basic Subject Information (All Subjects).

Subject	Age (yrs)	Height (cm)	Weight (kg)	Days of Running per Week
1	27	171.0	77.3	3.0
2	23	170.2	65.1	3.0
3	20	185.4	65.0	3.5
5	24	175.3	77.3	4.5
6	23	193.0	115.9	4.0
7	29	190.5	138.6	3.0
8	25	180.3	70.5	6.0
9	24	180.3	69.5	4.5
10	36	172.7	68.2	7.0
11	22	180.3	74.5	3.5
12	27	175.3	75.9	0.0
13	27	182.9	66.4	4.0
14	28	181.0	75.0	4.0
mean	**25.8**	**179.9**	**79.9**	**3.8**
st. dev.	**4.0**	**7.0**	**22.0**	**1.7**

Table 10. Subject Screening Data (All Subjects).

Subject	STJ neutral (deg)	Maximum Inversion (deg)	Maximum Eversion (deg)	Relaxed Bipedal Calcaneal angle (deg)	Single Leg Calcaneal angle (deg)	alpha (deg)	beta (deg)	Forefoot Varus (deg)
1	0.0	24.7	-14.0	-9.0	-13.3	20.0	14.9	4.0
2	0.3	35.0	-15.7	-9.7	-11.0	15.9	16.2	3.0
3	1.3	26.0	-7.7	-7.3	-10.3	29.6	23.5	7.0
5	-0.3	39.0	-13.0	-9.7	-13.0	23.3	26.3	5.0
6	1.0	25.0	-12.3	-7.7	-10.0	22.1	13.3	6.0
7	-2.3	32.3	-18.0	-5.7	-14.7	24.4	11.9	11.0
8	0.3	25.3	-14.3	-12.0	-15.7	24.9	14.0	5.0
9	2.0	44.7	-8.3	-9.0	-9.3	20.8	17.7	16.0
10	0.0	38.3	-14.0	-4.3	-9.0	24.8	9.5	5.0
11	1.7	31.0	-10.7	-11.7	-13.7	21.1	18.6	7.0
12	-2.7	23.0	-20.3	-18.0	-20.0	23.0	13.2	6.0
13	8.0	34.7	-13.3	-5.0	-8.0	23.4	17.3	0.0
14	1.7	39.7	-11.0	-5.7	-10.3	20.3	17.4	8.0
mean	**0.8**	**32.2**	**-13.3**	**-8.8**	**-12.2**	**22.6**	**16.5**	**6.4**
st. dev.	**2.6**	**7.0**	**3.5**	**3.7**	**3.3**	**3.2**	**4.6**	**3.9**

Appendix E

Illustrations of Kinematic and Center of Pressure Variables

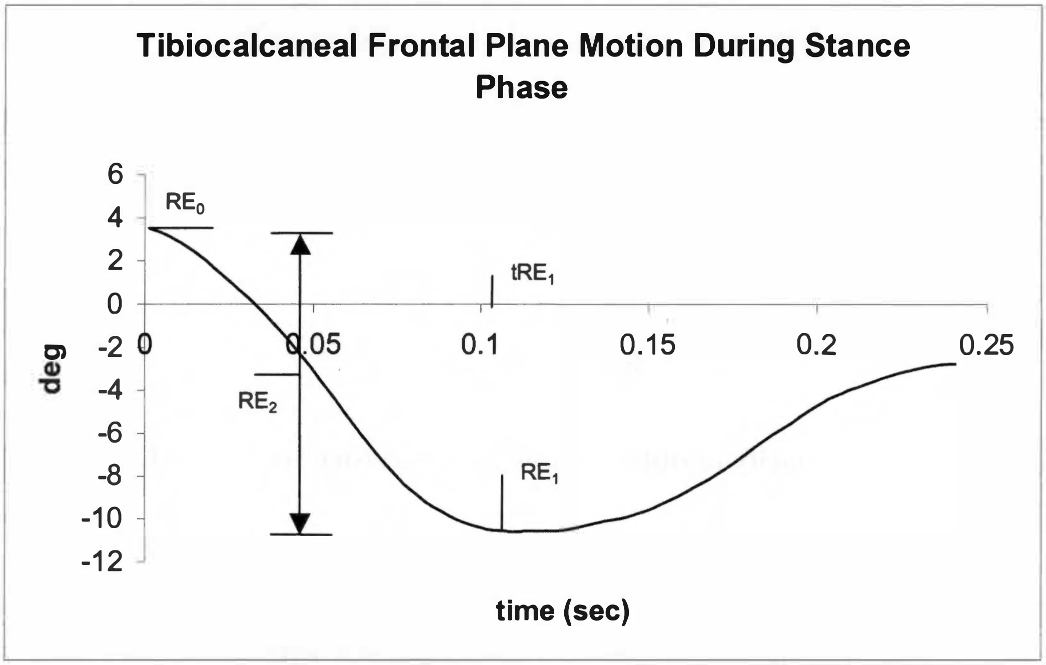

Figure 10. RE_0: Tibiocalcaneal inversion at heel-strike (deg). RE_1: Maximum tibiocalcaneal eversion (deg). tRE_1: Time of RE_1 (sec). RE_2: ROM of tibiocalcaneal eversion (deg).

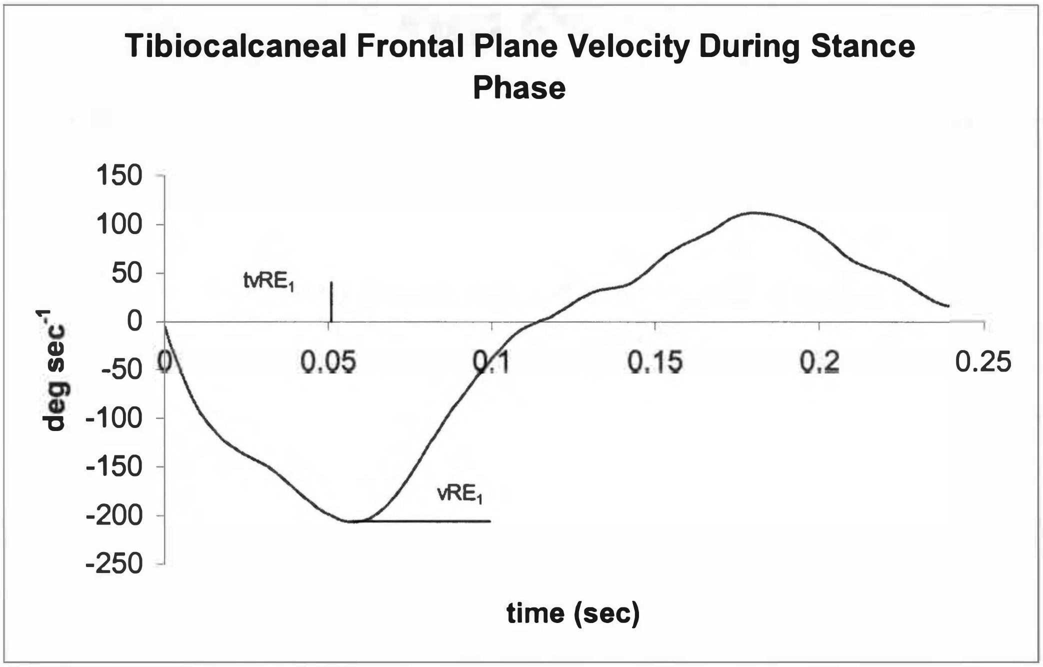

Figure 11. vRE_1: Peak velocity of tibiocalcaneal eversion (deg sec^{-1}). $tvRE_1$: Time of vRE_1 (sec).

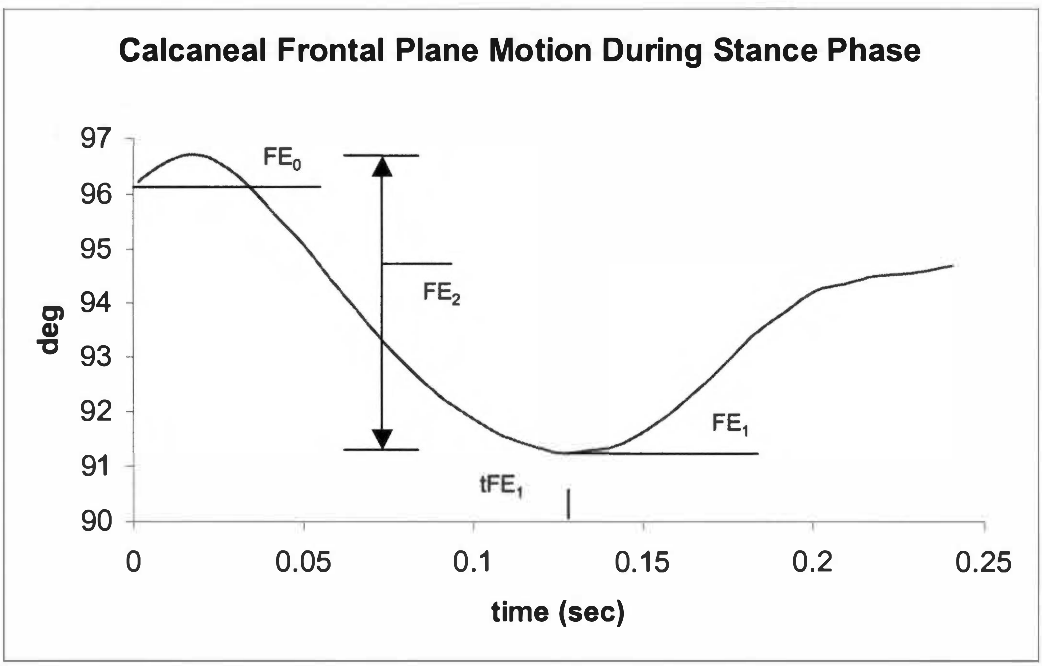

Figure 12. FE_0: Heel inversion at heel-strike (deg). FE_1: Maximum heel eversion (deg). tFE_1: Time of FE_1 (sec). FE_2: ROM of heel eversion (deg).

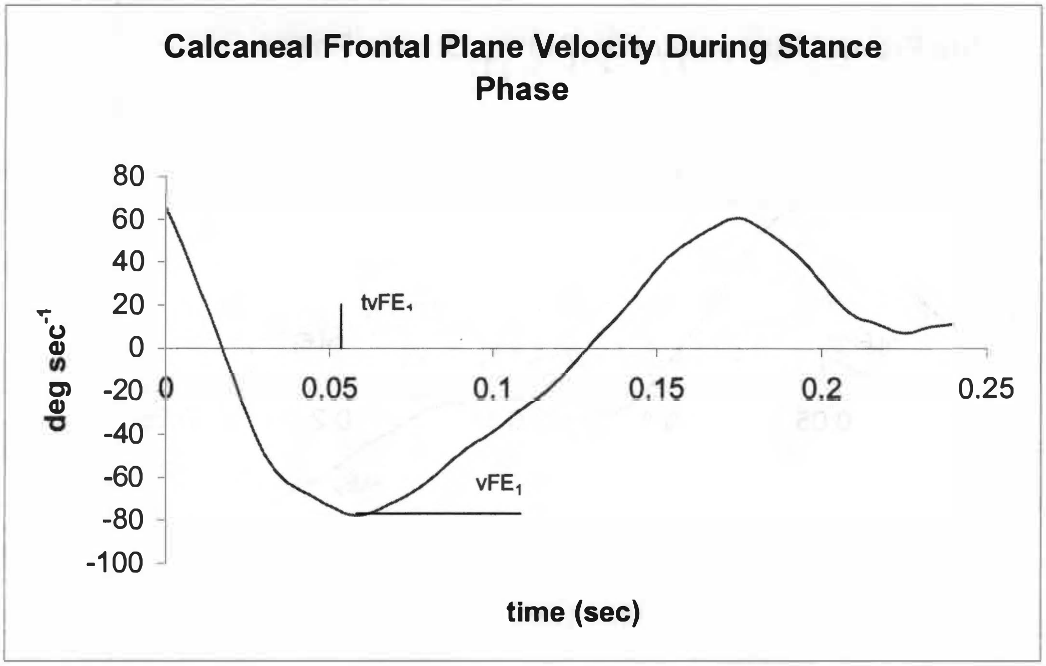

Figure 13. vFE_1: Peak velocity of heel eversion (deg sec^{-1}). $tvFE_1$: Time of vFE_1 (sec).

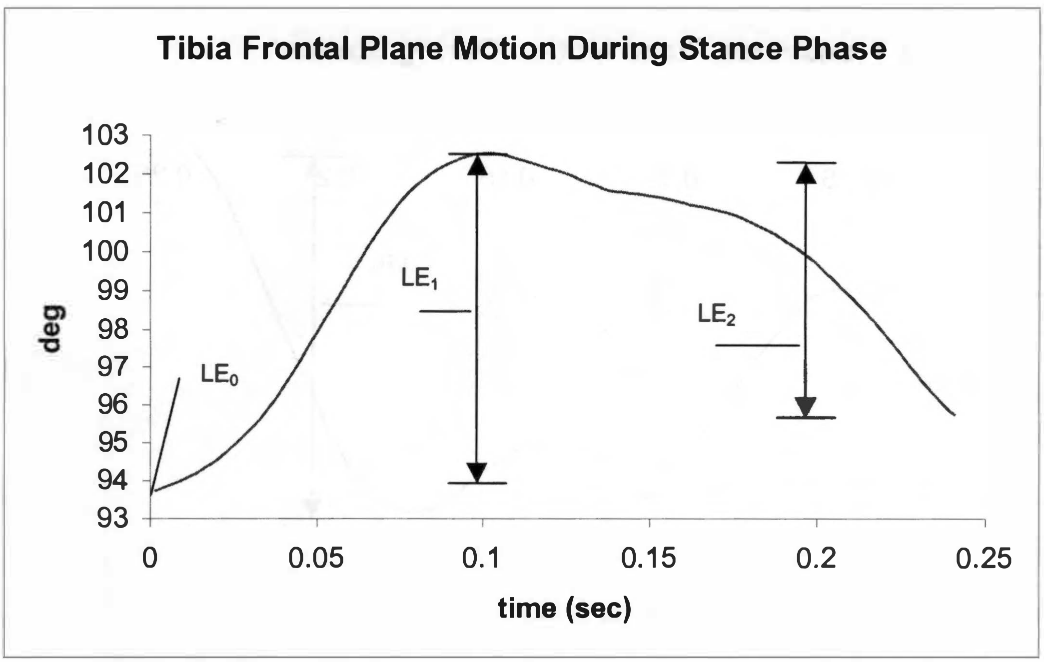

Figure 14. LE_0: Tibia angle in the frontal plane at heel-strike (deg). LE_1: ROM of tibia varus (deg). LE_2: ROM of tibia valgus (deg).

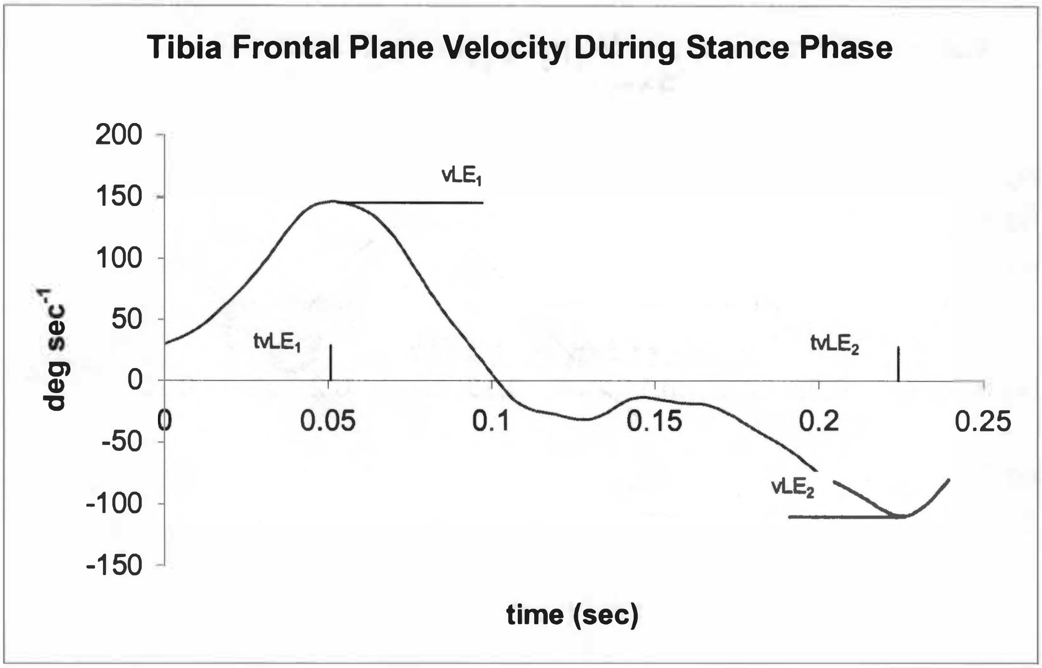

Figure 15. vLE_1: Peak velocity of tibia varus (deg sec^{-1}). $tvLE_1$: Time of vLE1 (sec). vLE_2: Peak velocity of tibia valgus (deg sec^{-1}). $tvLE_2$: Time of vLE_2 (sec).

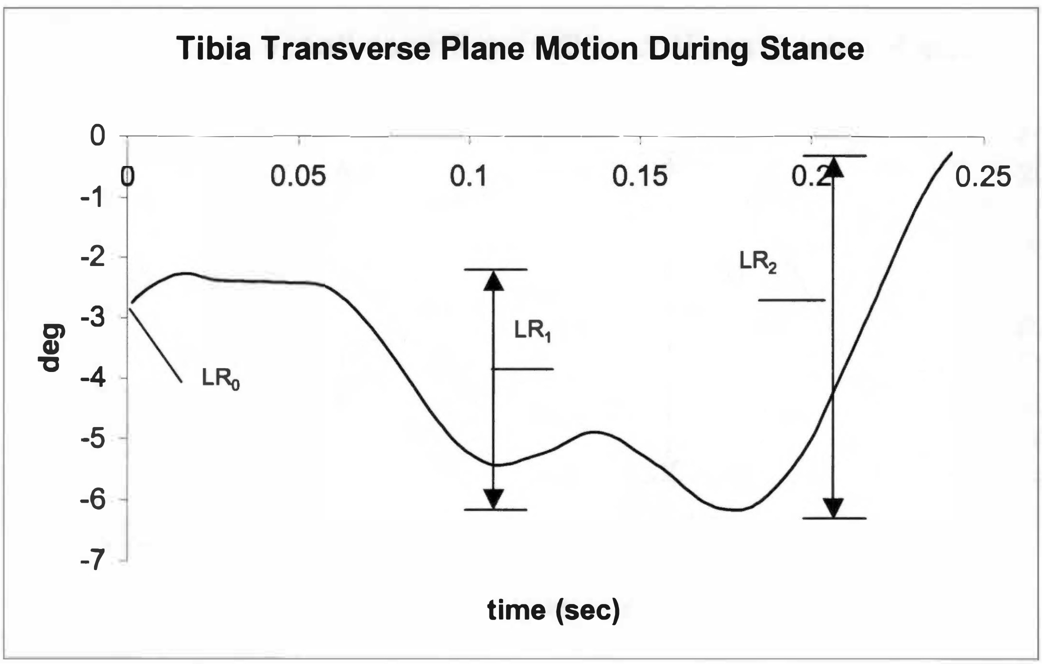

Figure 16. LR_0: Tibia angle in the transverse plane at heel-strike (deg). LR_1: ROM of tibia internal rotation (deg). LR_2: ROM of tibia external rotation (deg).

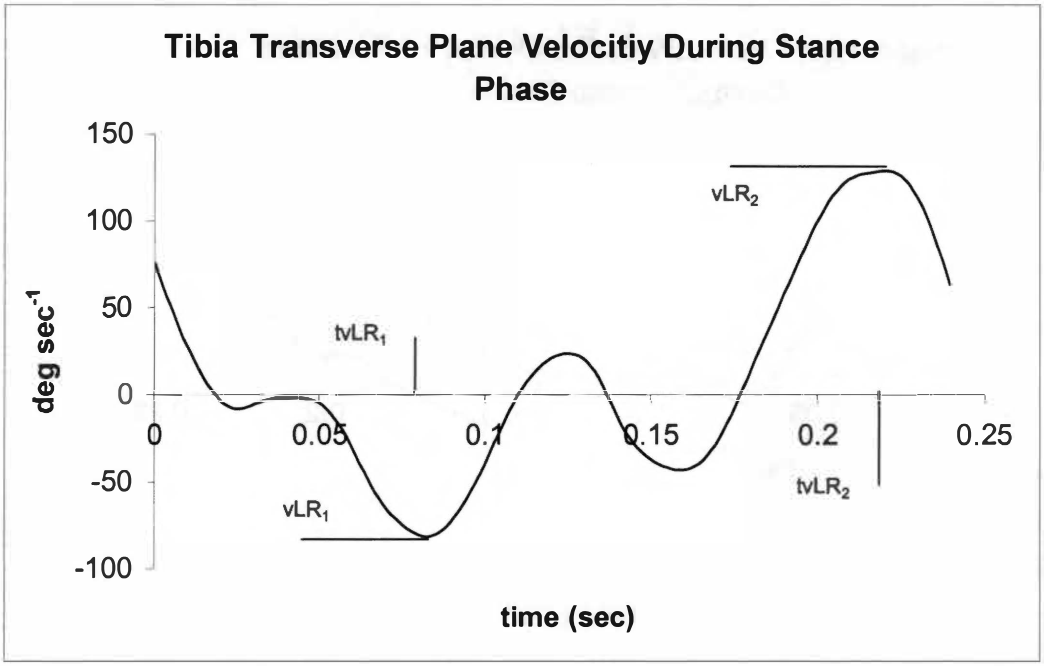

Figure 17. **vLR_1: Peak velocity of tibia internal rotation (deg sec^{-1}). $tvLR_1$: Time of vLR_1 (sec). vLR_2: Peak velocity of tibia external rotation (deg sec^{-1}). $tvLR_2$: Time of vLR_2 (sec).**

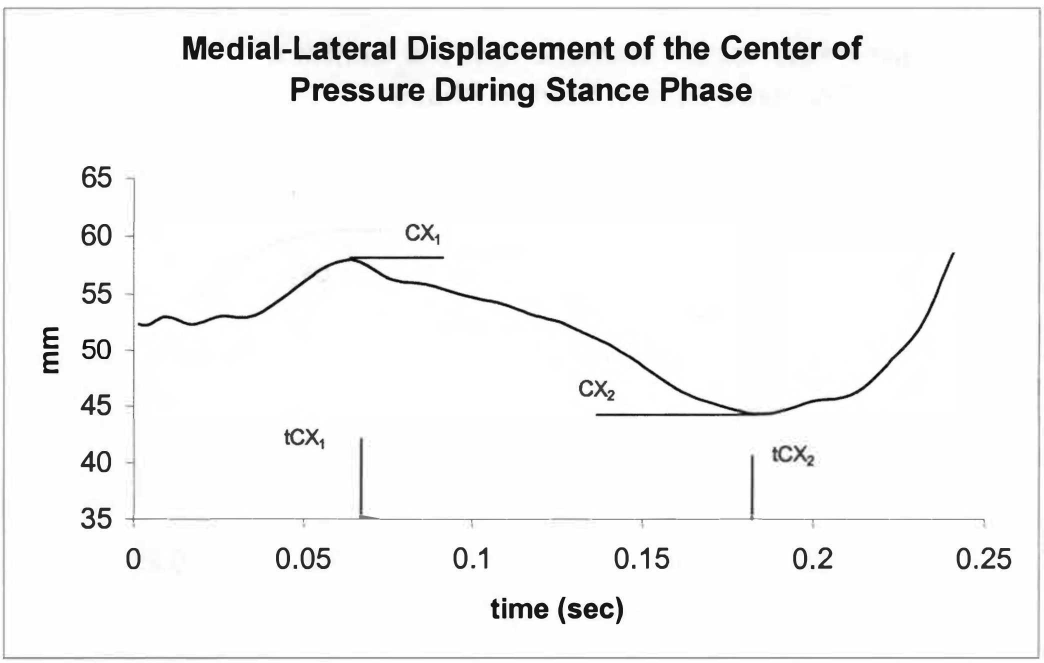

Figure 18. **CX_1: Most lateral point of the path of the COP (mm). tCX_1: Time of CX_1 (sec). CX_2: Most medial point of COP (mm). tCX_2: Time of CX_2 (sec).**

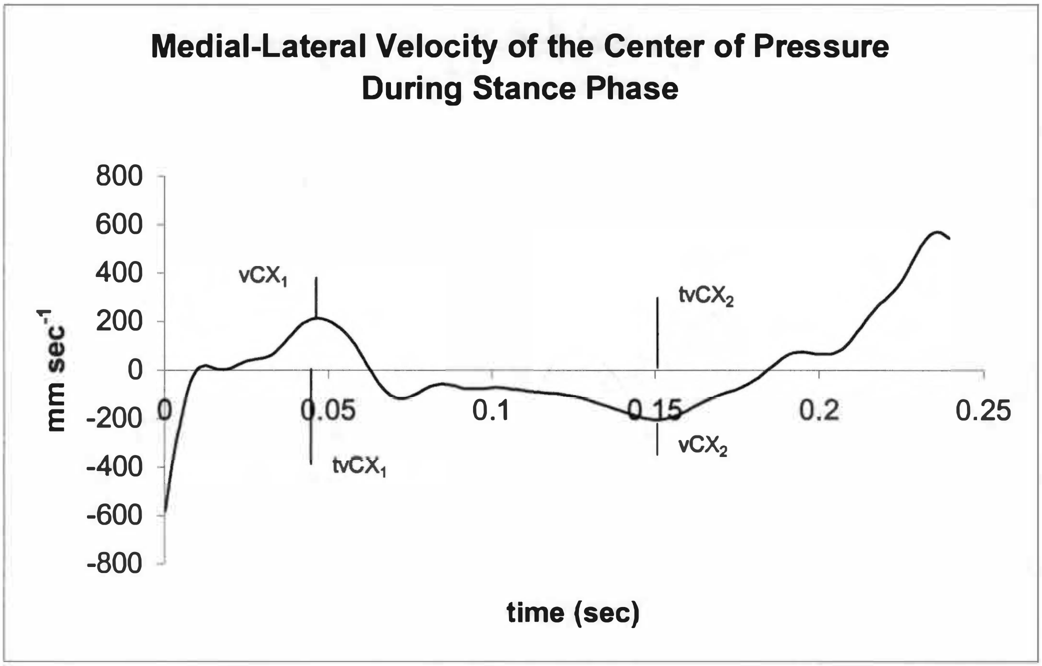

Figure 19. vCX_1: Peak lateral velocity of COP (mm sec^{-1}). $tvCX_1$: Time of vCX_1 (sec). vCX_2: Peak medial velocity of COP (mm sec^{-1}). $tvCX_2$: Time of vCX_2 (sec).

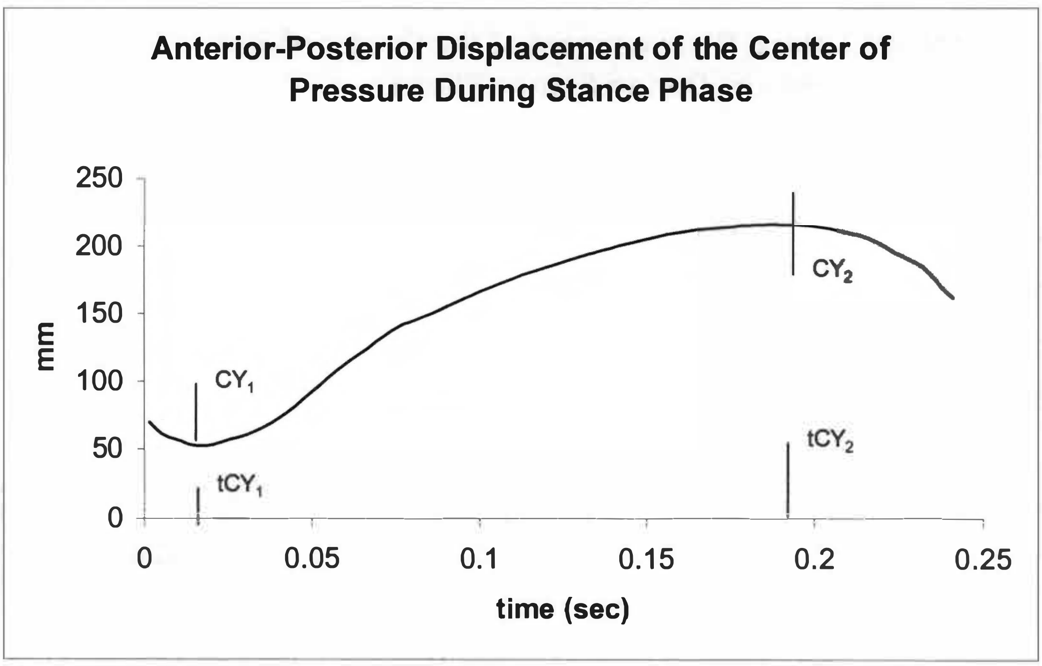

Figure 20. CY_1: Most posterior point of COP (mm). tCY_1: Time of CY_1 (sec). CY_2: Most anterior point of COP (mm). tCY_2: Time of CY_2 (sec).

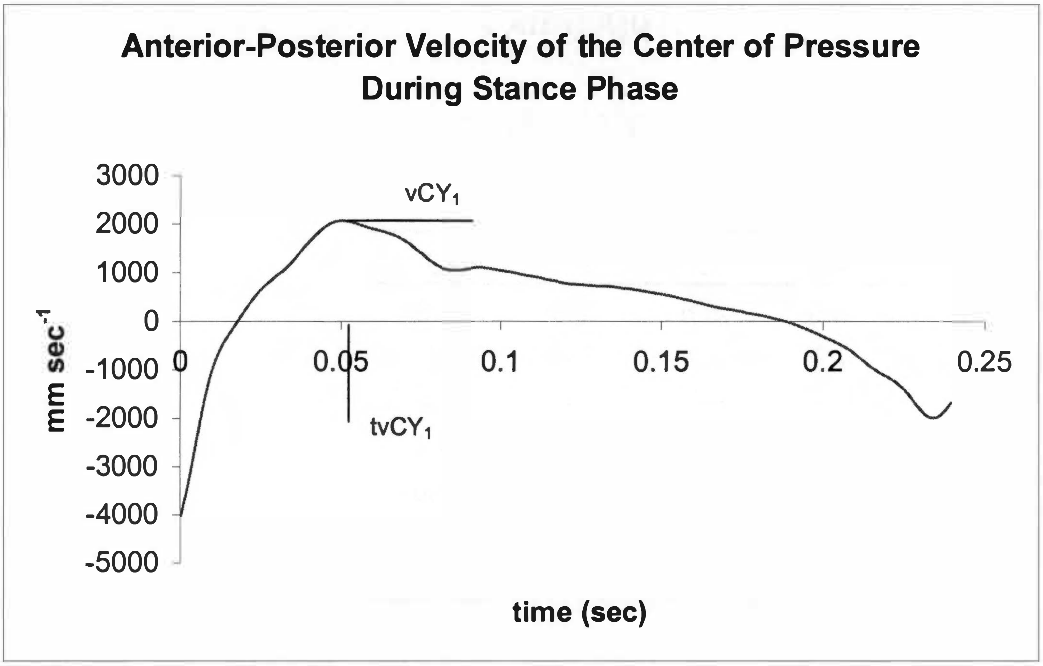

Figure 21. vCY_1: Peak anterior velocity of COP (mm sec^{-1}). $tvCY_1$: Time of vCY_1 (sec).

Appendix F

Complete Results

Table 11. Values of RE_0 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-5.3	-6.0	-6.6	-7.8	-5.8
2	6.1	4.9	6.2	3.4	4.0
8	7.0	6.3	6.3	6.6	8.4
9	14.9	17.1	13.3	10.7	13.1
11	12.1	12.0	12.8	13.8	13.6
13	6.3	8.3	4.4	11.2	9.8
14	3.4	0.7	2.0	-2.7	0.5
mean	**6.4**	**6.2**	**5.5**	**5.0**	**6.2**
s.d.	**6.5**	**7.5**	**6.8**	**7.9**	**7.1**

Table 12. Values of RE_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-11.4	-10.8	-9.5	-11.0	-11.7
2	-11.3	-8.9	-8.8	-9.1	-8.9
8	-6.8	-5.9	-8.3	-6.6	-7.6
9	-6.4	-5.3	-6.8	-5.1	-6.4
11	-7.8	-6.2	-9.4	-8.4	-8.4
13	-14.6	-13.7	-15.3	-14.9	-14.7
14	-13.5	-13.6	-13.3	-14.1	-15.5
mean	**-10.3**	**-9.2**	**-10.2**	**-9.9**	**-10.5**
s.d.	**3.3**	**3.6**	**3.0**	**3.7**	**3.6**

Table 13. Values of tRE_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.178	0.118	0.157	0.143	0.161
2	0.107	0.117	0.117	0.121	0.120
8	0.133	0.112	0.143	0.106	0.119
9	0.102	0.088	0.095	0.091	0.087
11	0.118	0.123	0.104	0.121	0.114
13	0.171	0.156	0.155	0.148	0.163
14	0.118	0.113	0.123	0.130	0.119
mean	**0.132**	**0.118**	**0.128**	**0.123**	**0.126**
s.d.	**0.030**	**0.020**	**0.025**	**0.020**	**0.027**

Table 14. Values of RE_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-7.1	-6.8	-6.7	-7.1	-6.7
2	-10.4	-8.1	-10.2	-8.7	-6.5
8	-15.1	-13.7	-14.4	-16.2	-17.2
9	-26.5	-27.4	-21.4	-21.6	-21.9
11	-20.0	-21.6	-24.0	-24.6	-25.3
13	-22.3	-20.9	-22.2	-29.4	-27.6
14	-14.1	-14.2	-17.4	-11.4	-12.6
mean	**-16.5**	**-16.1**	**-16.6**	**-17.0**	**-16.8**
s.d.	**6.8**	**7.5**	**6.5**	**8.5**	**8.6**

Table 15. Values of vRE_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-123.0	-154.4	-111.1	-188.7	-143.3
2	-261.1	-220.1	-253.8	-223.4	-210.1
8	-276.1	-248.7	-279.5	-234.8	-294.0
9	-496.3	-459.8	-356.8	-260.9	-385.2
11	-419.6	-373.5	-437.3	-395.0	-414.0
13	-394.6	-371.6	-365.0	-487.9	-519.8
14	-337.8	-245.9	-301.9	-237.9	-283.4
mean	**-329.8**	**-296.3**	**-300.8**	**-289.8**	**-321.4**
s.d.	**122.8**	**107.3**	**103.7**	**109.2**	**128.0**

Table 16. Values of $tvRE_1$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.086	0.051	0.074	0.066	0.087
2	0.047	0.056	0.060	0.062	0.058
8	0.024	0.023	0.026	0.027	0.022
9	0.051	0.032	0.037	0.031	0.027
11	0.024	0.025	0.022	0.019	0.021
13	0.062	0.041	0.048	0.051	0.046
14	0.033	0.025	0.053	0.068	0.029
mean	**0.047**	**0.036**	**0.046**	**0.046**	**0.041**
s.d.	**0.022**	**0.013**	**0.019**	**0.020**	**0.024**

Table 17. Values of FE_0 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	92.2	93.3	91.1	92.3	91.4
2	99.4	97.9	100.3	97.7	97.3
8	100.3	100.8	101.3	101.5	101.7
9	109.6	111.3	107.3	104.6	107.6
11	106.5	110.1	108.5	107.7	106.4
13	105.6	107.7	104.6	111.9	108.7
14	105.1	101.5	104.3	103.7	100.2
mean	**102.7**	**103.2**	**102.5**	**102.8**	**101.9**
s.d.	**5.8**	**6.7**	**5.8**	**6.4**	**6.2**

Table 18. Values of FE_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	86.0	87.7	87.4	86.9	86.2
2	91.4	93.3	92.7	92.7	92.9
8	87.3	88.5	86.5	88.3	87.3
9	89.7	90.5	89.5	89.8	89.4
11	86.0	86.1	85.1	86.0	84.5
13	85.0	85.6	84.4	84.5	85.1
14	88.0	87.2	87.7	88.1	85.1
mean	**87.6**	**88.4**	**87.6**	**88.0**	**87.2**
s.d.	**2.3**	**2.7**	**2.8**	**2.7**	**3.0**

Table 19. Values of tFE_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.160	0.113	0.146	0.147	0.150
2	0.114	0.114	0.135	0.128	0.115
8	0.136	0.150	0.158	0.141	0.133
9	0.108	0.106	0.109	0.141	0.100
11	0.128	0.133	0.110	0.130	0.126
13	0.174	0.158	0.167	0.170	0.169
14	0.104	0.102	0.101	0.117	0.090
mean	**0.132**	**0.125**	**0.132**	**0.139**	**0.126**
s.d.	**0.027**	**0.022**	**0.026**	**0.017**	**0.028**

Table 20. Values of FE_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-8.5	-6.1	-5.1	-6.3	-8.2
2	-8.4	-6.3	-8.9	-6.9	-5.7
8	-12.9	-12.3	-14.7	-13.5	-10.8
9	-21.0	-20.8	-17.8	-14.7	-18.1
11	-20.4	-23.9	-23.4	-21.7	-21.8
13	-22.9	-22.1	-23.2	-31.4	-24.6
14	-16.6	-14.3	-17.4	-16.7	-15.1
mean	**-15.8**	**-15.1**	**-15.8**	**-15.9**	**-14.9**
s.d.	**6.0**	**7.3**	**6.9**	**8.7**	**7.0**

Table 21. Values of vFE_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-146.3	-119.2	-88.6	-109.9	-142.7
2	-153.4	-122.6	-143.8	-114.8	-110.5
8	-247.8	-231.4	-258.2	-235.2	-266.5
9	-419.1	-396.6	-302.2	-240.5	-348.7
11	-391.7	-436.6	-449.1	-353.9	-383.5
13	-400.5	-382.1	-377.4	-516.1	-476.1
14	-348.4	-273.4	-353.4	-300.7	-312.6
mean	**-301.0**	**-280.3**	**-281.8**	**-267.3**	**-291.5**
s.d.	**117.5**	**130.1**	**128.8**	**141.6**	**130.3**

Table 22. Values of $tvFE_1$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.080	0.049	0.074	0.070	0.078
2	0.047	0.056	0.060	0.065	0.053
8	0.030	0.032	0.033	0.035	0.029
9	0.052	0.034	0.039	0.033	0.029
11	0.028	0.027	0.026	0.028	0.027
13	0.066	0.044	0.054	0.053	0.052
14	0.034	0.032	0.037	0.042	0.034
mean	**0.048**	**0.039**	**0.046**	**0.047**	**0.043**
s.d.	**0.020**	**0.011**	**0.017**	**0.016**	**0.019**

Table 23. Values of LE_0 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	96.2	98.2	95.5	98.4	96.2
2	94.9	94.1	95.8	94.6	94.0
8	93.4	94.3	94.8	94.7	93.8
9	94.6	94.2	94.2	93.6	94.4
11	95.4	94.2	95.7	95.1	93.6
13	99.0	99.2	100.0	100.5	101.1
14	101.1	102.2	101.7	102.1	101.9
mean	**96.4**	**96.6**	**96.8**	**97.0**	**96.4**
s.d.	**2.7**	**3.3**	**2.8**	**3.3**	**3.6**

Table 24. Values of LE_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	2.1	1.9	2.5	3.6	2.8
2	8.3	8.1	6.9	7.9	8.0
8	2.6	2.2	2.1	1.9	2.2
9	3.1	3.1	3.0	2.9	2.7
11	-1.2	0.0	-1.4	0.7	1.0
13	1.6	1.7	1.3	0.1	0.5
14	2.2	1.6	4.0	2.5	4.1
mean	**2.7**	**2.7**	**2.6**	**2.8**	**3.0**
s.d.	**2.8**	**2.6**	**2.6**	**2.5**	**2.5**

Table 25. Values of LE_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	2.2	3.5	2.4	-4.1	1.9
2	-6.6	-6.2	-5.2	-5.4	-4.9
8	-2.0	-2.1	-2.2	-2.3	-2.8
9	-3.7	-4.2	-3.0	-3.4	-4.7
11	-9.1	-7.6	-7.7	-7.4	-7.2
13	0.1	-4.6	-0.7	0.0	-0.5
14	-6.7	-3.0	-3.7	-4.6	-2.5
Mean	**-3.7**	**-3.4**	**-2.9**	**-3.9**	**-3.0**
s.d.	**4.0**	**3.6**	**3.2**	**2.4**	**3.0**

Table 26. Values of vLE_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	42.5	47.0	34.9	82.2	60.8
2	136.4	116.2	128.8	131.8	111.3
8	48.5	38.9	35.9	48.1	45.2
9	84.2	85.9	74.2	66.7	70.4
11	25.3	17.0	13.8	21.8	23.5
13	33.8	27.7	24.3	32.0	26.7
14	48.4	94.7	107.4	94.1	94.7
mean	**59.9**	**61.0**	**59.9**	**68.1**	**61.8**
s.d.	**38.5**	**37.7**	**44.4**	**38.3**	**33.1**

Table 27. Values of $tvLE_1$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.027	0.040	0.039	0.056	0.019
2	0.053	0.050	0.060	0.055	0.055
8	0.018	0.031	0.012	0.048	0.015
9	0.056	0.033	0.038	0.048	0.038
11	0.027	0.053	0.033	0.023	0.025
13	0.015	0.024	0.060	0.047	0.032
14	0.067	0.066	0.065	0.068	0.068
mean	**0.038**	**0.042**	**0.044**	**0.049**	**0.036**
s.d.	**0.021**	**0.015**	**0.019**	**0.014**	**0.019**

Table 28. Values of vLE_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-9.9	-64.2	-35.2	-179.2	-49.5
2	-124.7	-129.2	-111.1	-128.4	-109.4
8	-47.9	-37.1	-42.6	-50.5	-57.2
9	-65.6	-86.9	-62.9	-58.6	-82.7
11	-170.1	-158.0	-168.5	-158.4	-146.6
13	-29.2	-99.4	-37.2	-16.4	-22.8
14	-142.7	-88.7	-123.6	-126.5	-91.0
mean	**-84.3**	**-94.8**	**-83.0**	**-102.6**	**-79.9**
s.d.	**61.5**	**39.9**	**51.9**	**61.0**	**41.1**

Table 29. Values of $tvLE_2$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.215	0.189	0.187	0.228	0.241
2	0.214	0.213	0.218	0.209	0.215
8	0.211	0.214	0.217	0.197	0.210
9	0.201	0.209	0.197	0.213	0.210
11	0.210	0.211	0.201	0.203	0.204
13	0.231	0.241	0.226	0.225	0.222
14	0.232	0.231	0.233	0.230	0.237
mean	**0.216**	**0.215**	**0.211**	**0.215**	**0.220**
s.d.	**0.011**	**0.017**	**0.017**	**0.013**	**0.014**

Table 30. Values of LR_0 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	1.3	0.5	-0.3	2.0	1.1
2	-1.7	-4.5	0.8	-3.6	-5.1
8	1.3	1.4	1.7	2.1	1.0
9	7.4	5.6	3.8	4.7	4.6
11	10.2	12.3	11.1	13.9	13.2
13	10.5	8.6	8.9	8.1	8.5
14	9.4	9.0	7.7	8.9	8.4
mean	**5.5**	**4.7**	**4.8**	**5.2**	**4.5**
s.d.	**5.0**	**5.8**	**4.4**	**5.7**	**6.1**

Table 31. Values of LR_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-8.2	-10.3	-6.8	-8.8	-8.2
2	-5.7	-2.9	-5.7	-4.0	-3.4
8	-4.9	-3.5	-3.7	-5.2	-3.7
9	-7.9	-6.8	-4.6	-5.2	-4.5
11	-7.5	-9.6	-7.5	-11.0	-9.7
13	-13.6	-11.2	-12.1	-12.5	-12.6
14	-16.0	-17.5	-17.1	-17.4	-18.3
mean	**-9.1**	**-8.8**	**-8.2**	**-9.2**	**-8.6**
s.d.	**4.1**	**5.0**	**4.8**	**4.8**	**5.5**

Table 32. Values of LR_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-6.8	-6.5	-4.9	3.2	-7.3
2	6.7	6.2	3.6	5.9	5.3
8	2.7	1.6	1.3	2.4	3.8
9	1.6	4.5	1.2	2.5	4.0
11	9.9	8.6	8.3	7.9	7.9
13	-3.5	0.7	-2.4	-3.4	-3.6
14	8.1	5.1	6.4	7.2	4.5
mean	**2.7**	**2.9**	**1.9**	**3.7**	**2.1**
s.d.	**6.1**	**4.9**	**4.6**	**3.9**	**5.4**

Table 33. Values of vLR_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-81.2	-111.2	-55.1	-156.7	-107.8
2	-121.7	-68.6	-117.6	-100.1	-77.0
8	-82.1	-64.0	-62.3	-88.7	-62.2
9	-157.8	-138.8	-73.6	-110.2	-83.0
11	-159.5	-183.7	-170.8	-246.6	-183.0
13	-212.9	-155.5	-202.3	-215.9	-189.2
14	-220.9	-289.8	-289.6	-254.9	-301.9
mean	**-148.0**	**-144.5**	**-138.7**	**-167.6**	**-143.4**
s.d.	**56.6**	**77.6**	**86.9**	**71.1**	**86.3**

Table 34. Values of $tvLR_1$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.079	0.083	0.083	0.086	0.114
2	0.074	0.071	0.069	0.071	0.085
8	0.059	0.048	0.061	0.035	0.061
9	0.072	0.046	0.060	0.045	0.056
11	0.053	0.046	0.050	0.052	0.054
13	0.092	0.077	0.081	0.081	0.082
14	0.033	0.038	0.034	0.032	0.041
mean	**0.066**	**0.058**	**0.063**	**0.057**	**0.070[a,b]**
s.d.	**0.019**	**0.018**	**0.017**	**0.022**	**0.025**

a- Significantly different from c2 ($F = 11.720$, $P = 0.014$)
b- Significantly different from c4 ($F = 10.440$, $P = 0.018$)

Table 35. Values of vLR_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-107.0	56.8	-22.8	220.7	-11.2
2	156.2	156.9	118.4	174.4	136.9
8	52.0	48.8	28.0	55.5	79.9
9	-0.9	140.5	60.6	82.9	111.9
11	199.5	191.8	204.1	182.7	178.1
13	-112.7	39.0	-72.7	-67.7	-71.1
14	190.3	191.9	231.8	220.8	144.8
mean	**53.9**	**118.0**	**78.2**	**124.2**	**81.3**
s.d.	**133.5**	**67.9**	**113.2**	**106.3**	**90.6**

Table 36. Values of $tvLR_2$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.202	0.190	0.192	0.226	0.226
2	0.209	0.209	0.214	0.205	0.211
8	0.206	0.213	0.215	0.208	0.206
9	0.155	0.206	0.199	0.211	0.205
11	0.208	0.209	0.200	0.205	0.203
13	0.117	0.221	0.122	0.145	0.134
14	0.226	0.227	0.226	0.223	0.233
mean	**0.189**	**0.211**	**0.195**	**0.203**	**0.203**
s.d.	**0.039**	**0.012**	**0.034**	**0.027**	**0.032**

Table 37. Values of CX_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	5.5	5.4	5.3	5.5	5.3
2	5.9	5.9	6.1	6.0	5.7
8	5.3	5.2	5.1	5.1	5.0
9	6.5	6.4	5.8	5.8	6.0
11	5.8	5.5	5.7	5.6	5.8
13	5.6	5.4	5.4	5.4	5.7
14	5.3	5.1	5.1	5.2	4.8
mean	**5.7**	**5.5**	**5.5**	**5.5**	**5.5**
s.d.	**0.4**	**0.5**	**0.4**	**0.3**	**0.4**

Table 38. Values of tCX_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.094	0.092	0.078	0.087	0.117
2	0.063	0.060	0.064	0.056	0.048
8	0.017	0.017	0.027	0.036	0.022
9	0.037	0.017	0.017	0.017	0.017
11	0.017	0.026	0.017	0.019	0.017
13	0.019	0.017	0.022	0.033	0.017
14	0.073	0.069	0.069	0.058	0.045
mean	**0.046**	**0.043**	**0.042**	**0.044**	**0.040**
s.d.	**0.031**	**0.031**	**0.027**	**0.025**	**0.036**

Table 39. Values of CX_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	4.7	4.5	4.1	4.4	4.3
2	4.5	4.7	4.8	4.6	4.5
8	4.6	4.3	4.4	4.5	4.6
9	4.8	4.4	4.3	4.4	4.6
11	4.9	4.9	5.0	4.8	4.7
13	3.4	3.6	3.6	3.5	3.4
14	4.5	4.3	4.2	4.3	4.0
mean	**4.5**	**4.4**	**4.3**	**4.4**	**4.3**
s.d.	**0.5**	**0.4**	**0.5**	**0.4**	**0.5**

Table 40. Values of tCX_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.176	0.162	0.169	0.203	0.186
2	0.182	0.175	0.185	0.170	0.180
8	0.172	0.185	0.182	0.183	0.166
9	0.168	0.173	0.175	0.175	0.178
11	0.162	0.161	0.151	0.154	0.156
13	0.219	0.197	0.181	0.186	0.190
14	0.201	0.182	0.191	0.194	0.191
mean	**0.183**	**0.176**	**0.176**	**0.181**	**0.178**
s.d.	**0.020**	**0.013**	**0.013**	**0.016**	**0.013**

Table 41. Values of vCX_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	27.6	26.5	36.1	40.6	24.1
2	18.2	12.7	24.1	27.3	20.1
8	0.8	2.5	3.1	9.4	3.0
9	14.9	57.5	23.5	15.4	32.5
11	1.3	4.8	9.5	21.3	-0.5
13	5.6	-4.0	23.2	43.2	4.3
14	20.7	20.5	27.4	30.5	25.1
mean	**12.7**	**17.2**	**21.0**	**26.8**	**15.5**
s.d.	**10.4**	**20.6**	**11.1**	**12.5**	**13.0**

Table 42. Values of $tvCX_1$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.020	0.011	0.012	0.026	0.023
2	0.037	0.027	0.014	0.019	0.026
8	0.051	0.046	0.030	0.030	0.015
9	0.019	0.006	0.002	0.010	0.007
11	0.095	0.005	0.002	0.005	0.002
13	0.012	0.049	0.009	0.011	0.002
14	0.032	0.015	0.015	0.015	0.006
mean	**0.038**	**0.023**	**0.012**	**0.017**	**0.012**
s.d.	**0.028**	**0.018**	**0.010**	**0.009**	**0.010**

Table 43. Values of vCX_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	-21.6	-24.8	-29.4	-34.7	-24.5
2	-21.4	-20.3	-25.1	-21.9	-19.2
8	-7.5	-11.9	-8.0	-8.9	-4.8
9	-35.0	-11.5	-13.9	-17.2	-9.5
11	-5.1	-9.0	-5.2	-6.7	-4.4
13	-17.8	-15.9	-42.1	-17.9	-16.9
14	-15.2	-14.1	-16.1	-15.1	-14.3
mean	**-17.7**	**-15.3**	**-20.0**	**-17.5**	**-13.4**
s.d.	**10.0**	**5.5**	**13.0**	**9.2**	**7.5**

Table 44. Values of $tvCX_2$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.144	0.137	0.138	0.151	0.158
2	0.149	0.141	0.140	0.140	0.142
8	0.147	0.165	0.153	0.177	0.178
9	0.151	0.145	0.150	0.153	0.146
11	0.128	0.126	0.101	0.126	0.098
13	0.139	0.152	0.146	0.129	0.094
14	0.160	0.151	0.158	0.177	0.157
mean	**0.145**	**0.145**	**0.141**	**0.150**	**0.139**
s.d.	**0.010**	**0.012**	**0.019**	**0.021**	**0.032**

Table 45. Values of CY_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	5.6	6.0	5.4	5.4	5.2
2	5.5	6.1	6.7	5.6	5.2
8	10.1	8.9	8.7	8.7	8.8
9	11.1	10.6	8.5	7.8	9.0
11	7.8	7.0	8.2	7.4	8.0
13	6.4	6.4	5.8	5.5	5.8
14	4.8	5.5	5.2	4.9	5.1
mean	**7.3**	**7.2**	**6.9**	**6.5**	**6.7**
s.d.	**2.4**	**1.8**	**1.5**	**1.4**	**1.8**

Table 46. Values of tCY_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.014	0.010	0.011	0.022	0.027
2	0.017	0.013	0.029	0.013	0.013
8	0.008	0.026	0.022	0.026	0.012
9	0.053	0.019	0.008	0.008	0.014
11	0.008	0.008	0.008	0.008	0.008
13	0.012	0.025	0.011	0.011	0.010
14	0.018	0.014	0.019	0.019	0.018
mean	**0.019**	**0.016**	**0.015**	**0.015**	**0.015**
s.d.	**0.016**	**0.007**	**0.008**	**0.007**	**0.006**

Table 47. Values of CY_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	22.8	22.3	22.6	22.3	22.7
2	21.5	21.2	21.3	21.1	21.4
8	19.4	18.8	18.4	18.5	18.5
9	22.7	21.8	21.4	21.4	21.6
11	21.4	21.2	21.3	21.2	21.3
13	23.0	22.1	22.3	22.1	22.2
14	22.2	22.0	22.6	22.3	22.8
mean	**21.8**	**21.4**[a]	**21.4**	**21.3**[a]	**21.5**[d]
s.d.	**1.3**	**1.2**	**1.4**	**1.3**	**1.5**

b- Significantly different from c1 (P < 0.05)
d- Significantly different from c4 (P < 0.05)

Table 48. Values of tCY_2 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.180	0.165	0.172	0.183	0.192
2	0.184	0.176	0.165	0.174	0.181
8	0.158	0.172	0.163	0.172	0.160
9	0.143	0.174	0.171	0.176	0.174
11	0.176	0.173	0.164	0.163	0.169
13	0.200	0.204	0.197	0.197	0.197
14	0.218	0.198	0.216	0.215	0.214
mean	**0.180**	**0.180**	**0.178**	**0.183**	**0.184**
s.d.	**0.025**	**0.015**	**0.020**	**0.018**	**0.018**

Table 49. Values of vCY_1 across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	251.2	203.5	217.4	210.7	209.7
2	204.9	212.2	210.5	220.5	240.5
8	98.9	161.3	149.0	155.5	133.6
9	246.0	172.7	207.4	199.6	210.9
11	230.7	162.1	190.1	175.3	193.9
13	322.9	222.0	275.0	242.7	252.4
14	271.1	249.0	291.1	277.4	268.4
mean	**232.2**	**197.6**	**220.1**	**211.7**	**215.6**
s.d.	**69.3**	**33.4**	**48.7**	**40.8**	**44.8**

Table 50. Values of $tvCY_1$ across all subjects and conditions.

subject	c1	c2	c3	c4	c5
1	0.038	0.034	0.030	0.047	0.053
2	0.046	0.043	0.037	0.038	0.043
8	0.047	0.040	0.030	0.037	0.023
9	0.029	0.036	0.020	0.034	0.033
11	0.041	0.034	0.026	0.027	0.027
13	0.039	0.066	0.036	0.041	0.038
14	0.050	0.042	0.043	0.043	0.049
mean	**0.041**	**0.042**	**0.032**	**0.038**	**0.038**
s.d.	**0.007**	**0.011**	**0.008**	**0.006**	**0.011**

Vita

Michael Wortley was born in Latrobe, Pennsylvania in 1978. After moving several times, his state of residence has been Virginia since 1987. He graduated from Jefferson Forest High School in Forest, VA in 1997, and then from Johns Hopkins University in 2001 with a Bachelor of Science degree in biomedical engineering and a minor in computer science. He received a Master of Science degree in Human Performance and Sport Studies, with a focus on biomechanics, in 2003.

Wortley is also an avid runner, having been the captain of his varsity track teams in high school and college. He has also been coaching youth athletes through the Knoxville Track Club since 2001.