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The effect of arch height on tri-planar foot kinematics during gait

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THE EFFECT OF ARCH HEIGHT ON TRI-PLANAR FOOT KINEMATICS DURING GAIT

by

Jason Mitchell Wilken

An Abstract

Of a thesis submitted in partial fulfillment
of the requirements for the Doctor of
Philosophy degree in Physical Rehabilitation Science
in the Graduate College of
The University of Iowa

May 2006

Thesis Supervisor: Associate Professor H. John Yack

ABSTRACT

Deviations from normal foot structure are thought to result in hyper or hypo mobility preventing the foot from providing sufficient stability or mobility at appropriate times during the stance phase of gait. Such deviations in foot structure are often suggested as a risk factor for the development of common foot and ankle injuries. Though foot structure is believed to influence function its effect on foot motion and mechanisms thought to allow stability during the end of stance phase remain poorly understood. The purpose of this study was to determine the extent to which arch height affects foot kinematics during gait using a four segment foot model.

Subjects included 17 healthy individuals with a wide range of arch heights and no past history of foot pathology. A three-dimensional motion tracking system was used to determine kinematics of the hallux, first metatarsal, forefoot, calcaneus and tibia during gait while subjects ambulated at a predetermined walking velocity of 0.78 statures/s. AP and lateral radiographs were used to determine coordinate system orientation for the first metatarsal, forefoot and calcaneus. The sagittal plane angle between the first metatarsal and calcaneus was used to represent foot structure

No association was observed between foot structure, as represented by arch height, and foot kinematics as represented by excursion or coupling during the stance phase of gait. This surprising result provides no evidence to support the assertion that foot structure influences mobility during gait in individuals without foot pathology.

The kinematic patterns observed were highly consistent between individuals and provide new insight into the mechanisms underlying the formation of a rigid lever for pushoff at the end of stance phase. Supination, consisting of forefoot and first metatarsal adduction and plantarflexion along with hindfoot inversion, was found to occur at approximately 76 percent stance. Kinematic coupling associated with activation of the windlass mechanism is consistent with the concept of supination and occurs shortly after forward rotation of the first metatarsal and hallux dorsiflexion.

In conclusion, deviation from normal foot structure appears to have little or no influence on foot mobility or intersegmental coupling due to activation of the windlass mechanism.

Abstract Approved: ______________________________
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CERTIFICATE OF APPROVAL

PH.D. THESIS

This is to certify that the Ph.D. thesis of

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has been approved by the Examining Committee
for the thesis requirement for the Doctor of Philosophy
degree in Physical Rehabilitation Science at the May 2006 graduation.

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CHAPTER I

INTRODUCTION

It is commonly accepted that many foot injuries result from deviations from normal foot structure that cause hypo or hyper mobility preventing the foot from providing sufficient stability or mobility at the appropriate times during gait. As such, foot structure is often identified as a risk factor for the development of some of the most common foot and ankle injuries including stress fractures of the lower extremity (Sullivan, Warren et al. 1984; Simkin, Leichter et al. 1989; Kaufman, Brodine et al. 1999), posterior tibialis tendonitis (Williams, McClay et al. 2001), central heel pain, achilles tendonitis (Clement, Taunton et al. 1981) and plantar fasciitis among others (Cowan, Jones et al. 1993; Williams, McClay et al. 2001). The relationship between foot structure and pathology is not, however always clear as the foot has demonstrated ability to accommodate significant pathology and structural deviations (Giladi, Milgrom et al. 1985; Cowan, Jones et al. 1993; Hogan and Staheli 2002). There is overwhelming evidence that individuals can experience pain free function for extended periods of time while having abnormal foot structure or complete loss of motion at major joints (Mazur, Schwartz et al. 1979; Beischer, Brodsky et al. 1999). Though deviations from normal foot structure or function may result in pathology under high loading conditions or over longer periods of time (Ahlberg and Henricson 1981; Lidor, Ferris et al. 1997; Coester, Saltzman et al. 2001) the adaptability of the foot is remarkable. Despite decades of foot and ankle research, large gaps exist in our knowledge of dynamic foot function adding additional uncertainty to the treatment of foot and ankle pathology.

The dynamic nature of the foot and the ability to accommodate and adapt substantiates the need for in-vivo examination of foot function and leads us to re-examine conclusions based solely on in-vitro testing methods that can’t adequately capture dynamic function. The most commonly used in-vitro testing methods involve imposing single plane motion or non-physiological application of known forces to normal and surgically altered cadaveric feet (Ouzounian and Shereff 1989; Hintermann, Sommer et al. 1995; Kitaoka, Lundberg et al. 1995; Valderrabano, Hintermann et al. 2003; Valderrabano, Hintermann et al. 2003; Valderrabano, Hintermann et al. 2003; Imhauser, Siegler et al. 2004). Though providing a well controlled examination of foot motion common testing methods rarely attempt to emulate function of key muscles and often apply loads that are notably different than those observed during gait. The observed motion is then used to make conclusions regarding the stability and mobility of the foot (Hicks 1953; Gellman, Lenihan et al. 1987; Ouzounian and Shereff 1989; Hintermann and Nigg 1995), its ability to accommodate structural changes (Gellman, Lenihan et al. 1987; Savory, Wulker et al. 1998; Wulker, Stukenborg et al. 2000; Roling, Christensen et al. 2002; Imhauser, Siegler et al. 2004), supporting structures (Hicks 1954; Hicks 1955; Ker, Bennett et al. 1987; Hintermann, Sommer et al. 1995; Kogler, Solomonidis et al. 1996) as well as axes of rotation (Hicks 1953) and movement coupling parameters (Hicks 1953; Olerud and Rosendahl 1987; Valderrabano, Hintermann et al. 2003). A large portion of current theory regarding the function of particular joints or joint complexes is based on this method of testing.

The effect of ankle fusion or arthroplasty on motion within the foot has been characterized by several in-vitro studies (Valderrabano, Hintermann et al. 2003; Valderrabano, Hintermann et al. 2003). These studies suggest that fusion of the ankle joint alters foot function resulting in a general stiffening of the foot, decreasing its overall mobility (Gellman, Lenihan et al. 1987; Valderrabano, Hintermann et al. 2003), altering joint loading characteristics, as well as its movement coupling properties (Valderrabano, Hintermann et al. 2003) and that arthroplasty may minimize these effects (Valderrabano, Hintermann et al. 2003; Valderrabano, Hintermann et al. 2003). Early in-vivo studies, that attempted to validate the findings of the in-vitro studies were, however, only able to detect gross changes associated with movement of the whole foot relative to the leg rather than the more subtle changes in motion that occur within the foot (Mazur, Schwartz et al. 1979; Hefti, Baumann et al. 1980; Buck, Morrey et al. 1987). Therefore in-vivo testing was unable to substantiate the marked effects of ankle fusion on motion within the foot that were seen in cadaveric testing because of its limited sensitivity.

The recent development of multi-segment foot models allows the ability to assess in-vivo motion with greater sensitivity, thus providing the opportunity to challenge and assess current theory regarding foot function. Two major insights have come about as the result of examination of ankle fusion patients using multi-segment foot models. The first is quantitative confirmation of the ability of the foot to accommodate to the major changes associated with ankle fusion. In-vivo kinematic data suggests that ankle fusion patients are able to accomplish near normal motion of the forefoot (Wu, Su et al. 2000) and first metatarsal (Chapter II) relative to the tibia confirming the remarkable adaptability of the foot. Of great importance was the finding that the ability to accommodate to ankle fusion procedure varied between individuals (Su, Wu et al. 2001) suggesting a subject specific response to the changed demands on the foot.

The second major insight into dynamic foot function, resulting from the in-vivo testing of patients with fusion, relates to arch motion during the stance phase of gait. Recent investigations into arch motion have, with few exceptions (Leardini, Benedetti et al. 1999; MacWilliams, Cowley et al. 2003), relied on two dimensional approaches (King, Watkins et al. 1980; Kappel-Bargas, Woolf et al. 1998; Wearing, Urry et al. 1998; Cashmere, Smith et al. 1999; Hunt, Fahey et al. 2000; MacWilliams, Cowley et al. 2003; Wearing, Smeathers et al. 2004) with several studies examining arch motion in non-normal feet (King, Watkins et al. 1980; Wearing, Urry et al. 1998; Wearing, Smeathers et al. 2004). New findings regarding dynamic foot function following fusion and arthroplasty suggest that current theory regarding arch motion during gait may need to be rethought (Chapter II). By assessing the motion of the first metatarsal relative to the calcaneus the motion of the medial longitudinal arch was determined using surface markers. The first finding of interest is that arch range of motion from initial contact to peak elongation during the stance phase of gait is not changed by ankle fusion or arthroplasty and is independent of walking velocity (Chapter II). Contrary to clinical reasoning, suggesting hyper or hypomobility of the foot following loss of ankle motion, both groups demonstrated arch motion of approximately ten degrees, nearly identical to that of control subjects. This finding suggests that the arch is able to maintain its sagittal plane function independent of ankle motion and independent of loading changes associated with walking velocity (Chapter II).

The second, and potentially most important finding, relates to arch motion during the stance phase of gait and the contribution of the inclination of the first metatarsal to the total motion that occurs. It was observed that the first metatarsal maintains a nearly static inclination relative to the floor for nearly half of the stance phase of gait and that the previously reported intersegmental motion is primarily due to motion of the calcaneus as it is unloaded and plantarflexes relative to the floor. This finding was very robust as the patterns and magnitudes of motion were highly repeatable within an individual and nearly identical for patients with fusion, arthroplasty and controls and across walking velocities (Chapter II). This finding is important in that it indicates that the medial longitudinal arch is functioning in a manner considerably different than proposed by current theory and as modeled in cadaveric testing. Therefore current theory and efforts to model and understand arch function during gait may not represent in-vivo foot function.

The contradiction between our results and current theory may be rooted in the models that have been generated to represent structures and mechanisms thought to control complex motions and maintain a normal balance between foot mobility and stability during the stance phase of gait. The most common models attempt to communicate dynamic foot function using mechanical analogs such as the windlass (Hicks 1954; Sarrafian 1987; Fuller 2000), leaf spring (Van Boerum and Sangeorzan 2003) universal joint (Olerud and Rosendahl 1987), beam (Hicks 1955; Sarrafian 1987), truss and tie-rod (Hicks 1955; Sarrafian 1987) and arch (Hicks 1955; Sarrafian 1987) among others. It is no surprise that many of the mechanical representations focus on the support mechanisms of medial longitudinal arch and its ability to withstand the loads that are imposed during the stance phase of gait. The truss and tie-rod model is one of the most commonly used representations and focuses solely on sagittal plane motion of the foot during the stance phase of gait. In this model the axial load imposed by body weight is distributed through proximal and distal struts often thought to represent the calcaneus and the first metatarsal. The distance between the most inferior ends of the struts is maintained by a tie-rod that is often represented as having elastic properties. In this representation it is the plantar fascia that is believed to provide a large portion of the necessary support to prevent the apex of the arch from lowering to the floor. It is important to note, however, that this and all other currently used models assume that the proximal and distal ends of the foot are loaded and in contact with the floor. Recent kinematic (Chapter II) and kinetic data (Warren, Maher et al. 2004) suggest that this manner of loading does not represent what occurs during the stance phase of gait, and that the calcaneus rotates about a nearly stationary first metatarsal (Chapter II), calling into question the validity of these models to represent foot function and support mechanisms during gait.

As per the modeling work that has been done, many researchers consider the integrity and mobility of the medial longitudinal arch to be of primary importance. This is, in part, because arch height is believed to be a key determinant of foot mobility during gait. Segmental and articular orientations associated with normal arch structure are believed to be necessary to allow for normal muscular function, appropriate accommodation to varied terrain, attenuation of impact loads and mobility within normal articular ranges. There is, however, limited understanding of the affect of arch height on dynamic foot mobility under standardized walking conditions let alone the widely varying demands that are placed on the foot (Hunt, Fahey et al. 2000; Hunt and Smith 2004).

Given this orientation and justification, current theories and models of foot motion in conjunction with normative kinematic databases, are used to support the belief that foot pathology associated with high or low arches is due to associated changes in intersegmental foot motion. There are several inter-related ways that arch height is believed to influence foot motion during gait. It is commonly hypothesized that the mobility of the foot is inversely related to the height of the medial longitudinal arch (Kirby 1989; Arangio, Chen et al. 1998; Van Boerum and Sangeorzan 2003; Wang and Crompton 2004). It is commonly assumed that as arch height decreases articular geometry, supporting soft tissue structures and the windlass effect are less able to effectively withstand the imposed loads (Hicks 1955; Mann and Inman 1964; Van Boerum and Sangeorzan 2003; Wang and Crompton 2004) leading to decreased rigidity and increased motion. The foot is thought to succumb to the imposed load resulting in a "splaying" of the foot due to the combined effects of increased hindfoot eversion and forefoot abduction and dorsiflexion (Kitaoka, Lundberg et al. 1995; Van Boerum and Sangeorzan 2003). In contrast, a high arched foot is thought to be excessively rigid allowing considerably less motion than the normal foot (Arangio, Chen et al. 1998).

In addition to influencing the total motion that occurs within the foot, arch height is also thought to influence the coupling of motion across movement planes. According to current theory arch height should influence the key mechanisms that influence the ability of the foot to supinate for propulsion. It has been repeatedly demonstrated that the coupling of frontal and transverse plane motion between the hindfoot and tibia is affected by arch height during running (Nigg, Cole et al. 1993; Nawoczenski, Saltzman et al. 1998; DeLeo, Dierks et al. 2004). It is, however, unclear how arch height influences intersegmental motion at other joints during gait. In particular the lowering of medial longitudinal arch potentially alters the frontal and sagittal plane inclination of the transverse tarsal joint. If true, the kinematic patterns would likely demonstrate an increase in sagittal plane arch motion with decreased coupling between sagittal and transverse plane motion. It is also thought that due to the everted position of the calcaneus and associated decrease in inclination of the STJ axis of rotation that the hindfoot inverts less relative to the tibia (Kirby 1989; Arangio, Chen et al. 1998). The net effect is believed to be increased sagittal plane motion (Franco 1987; Van Boerum and Sangeorzan 2003) with less forefoot adduction (Van Boerum and Sangeorzan 2003; Hunt and Smith 2004) and inversion such that the forefoot is less able to form a rigid lever for pushoff (Fuller 2000; Kirby 2000; Van Boerum and Sangeorzan 2003). The opposite would be true in the cavus foot in which the inclination of the transverse tarsal joint would theoretically be higher resulting in relatively more forefoot adduction as compared to forefoot plantarflexion resulting in a different coupling ratio. Though current theory suggests substantial changes in the magnitude and coupling of foot motion associated with arch height such claims have not been validated using in-vivo kinematic data (Hunt, Fahey et al. 2000; Hunt and Smith 2004).

In order to gain a better understanding of how foot mobility affects foot function during gait it will therefore be necessary to assess foot dynamics using a multi-segment foot model. The purpose of this study is to determine the extent to which arch height affects foot kinematics during gait using a five segment foot model. This data will allow the ability to determine if arch height effects dynamic foot function to the extent and in the manner currently hypothesized based on current theory regarding foot function.

Purpose

The purpose of this study is to describe the kinematic interdependence between foot segments and planes of motion as they occur within the context of varied arch height.

Specific Aims

1. To describe the effect of foot structure (specifically arch height) on the three-dimensional kinematics of the first metatarsal, lateral forefoot, hindfoot and tibia during gait.
2. To determine the extent to which inter-segmental foot motion is coupled during terminal stance and determine if these coupling values vary in a consistent manner due to changes in foot architecture associated with arch height.

Hypothesis #1

Arch height will have a moderate negative association (correlation values between 0.5 and 0.75) with sagittal (first metatarsal relative to the calcaneus), transverse (forefoot relative to the calcaneus) and frontal plane (calcaneus relative to the tibia) motion of the foot.

Rationale for Hypothesis #1

Current theory regarding dynamic foot function during gait suggests that arch height plays an important role in determining the magnitude of motion that occurs between segments of the foot as well as between the foot and leg during gait. In-vitro study and current theory suggest that as arch height decreases the structural stability of the foot decreases resulting in decreased rigidity of the foot and increased excursion at the joints of interest during gait (Kirby 1989; Arangio, Chen et al. 1998; Van Boerum and Sangeorzan 2003; Wang and Crompton 2004). Though increased mobility associated with decreased arch height is commonly accepted in clinical education and accommodating for these supposed changes in motion is commonly the goal of clinical intervention little in-vivo evidence exists to support these claims (Hunt, Fahey et al. 2000; Hunt and Smith 2004). To date the few studies that have attempted to determine the effect of arch height on foot motion during gait have used randomly selected populations or poorly defined arch height groups providing limited insight into the effect of foot structure on function (Hunt, Fahey et al. 2000; Hunt and Smith 2004). This investigation is supported by pilot data (n=5) suggesting strong relationships do exist between arch height and foot mobility. This study will serve as the first systematic examination of the effect of radiographically determined arch height on intersegmental foot mobility during gait.

Hypothesis #2

Foot structure changes associated with arch height will lead to altered motion coupling between segments. These will be demonstrated through moderate positive associations (correlation values between 0.5 and 0.75) between arch height and the coupling ratios obtained during the terminal stance.

Coupled motions: a) Transverse plane motion of the CALC vs. TIB and frontal plane motion of the CALC vs. TIB. b) Transverse plane motion of the FORE vs. CALC and frontal plane motion of the CALC vs. TIB. c) Transverse plane motion of the first metatarsal vs. CALC and sagittal plane motion between the two segments. d) Transverse plane motion of the FORE vs. CALC and sagittal plane motion between the two segments.

Rationale for Hypothesis #2

Current theory regarding dynamic foot function during gait suggests that arch height plays an important role in determining the motion patterns that occur between segments of the foot as well as between the foot and leg during gait. It is commonly thought that with decreased arch height that the foot is less able to adduct and invert (supinate) at the end of stance phase to form a rigid lever for push off. This is thought to result from altered coupling of motions, i.e. the relative proportion of motion occurring in each plane, due to bony geometry changes which result in a lowering of the medial longitudinal arch. Though arch height has been demonstrated to influence subtalar joint motion and associated transverse and frontal plane motion between the hindfoot and tibia during running it is not known if the same is true during gait. Theoretically changes in foot structure associated with decreasing arch height result in decreased sagittal plane inclination, and increased medial deviation of both the subtalar and transverse tarsal joints axes of rotation due to the decreased apical nature of the arch. By altering the orientation of the axes of rotation the proportion of motion in each plane would be altered during terminal stance phase. The net effect of these changes would be demonstrated in altered coupling of motions across planes of movement and between segments dependent on the height of the medial longitudinal arch.

CHAPTER II

NEW INSIGHTS INTO ARCH FUNCTION DURING GAIT

Introduction

The medial longitudinal arch (MLA) is one of the most prominent morphologic features of the foot and, therefore, has garnered considerable attention. Changes in the structural integrity of the foot are commonly attributed to arch integrity, which is believed to play an important role in pathologic dynamics. Arch integrity is generally discussed using the truss and tie rod theory (Lapidus 1963) in which the arch is modeled with an anterior and posterior strut, representing the proximal and distal aspects of the arch. These two struts are connected at the apex and arch integrity is maintained via a tie-rod representing the plantar soft tissues, specifically, the plantar aponeurosis. In an effort to better understand arch structure and gain insight into dynamic function many attempts have involved qualitative and quantitative assessment of arch related variables such as height (Cowan, Robinson et al. 1994; Saltzman, Nawoczenski et al. 1995; Williams and McClay 2000), the effect on motion of proximal segments (Nawoczenski, Saltzman et al. 1998) and loading characteristics (Staheli, Chew et al. 1987; Vogler and Bojsen-Moller 2000; Tsung, Zhang et al. 2003; Wearing, Hills et al. 2004). The interpretation of these rather gross indicators is, however, biased and colored by basic assumptions that were primarily formed as a result of bench testing on cadaveric feet.

Quantification of arch kinematics during functional activities, such as gait, has until recently been limited. Recent advances in tracking technology and interest in dynamic foot function has resulted in several studies attempting to quantify and describe arch motion. (King, Watkins et al. 1980; Kappel-Bargas, Woolf et al. 1998; Wearing, Urry et al. 1998; Cashmere, Smith et al. 1999; Leardini, Benedetti et al. 1999; Hunt, Smith et al. 2001; Hunt, Smith et al. 2001; MacWilliams, Cowley et al. 2003). Studies that are available provide similar though limited data regarding dynamic arch motion during gait. When taken as a whole this body of work suggests an elongation of the medial longitudinal arch due to first metatarsal dorsiflexion until late in the stance phase at which point arch height increases due to the function of the windlass mechanism. Interpretation of the data beyond this initial understanding is complicated by the frequent use of 2D modeling approaches (King, Watkins et al. 1980; Kappel-Bargas, Woolf et al. 1998; Wearing, Urry et al. 1998; Cashmere, Smith et al. 1999; Wearing, Urry et al. 1999; Hunt, Smith et al. 2001; Hunt, Smith et al. 2001), the absence of direct comparison between proximal and distal bony segments (Leardini, Benedetti et al. 1999), the assumption of segment rigidity (MacWilliams, Cowley et al. 2003), and the near exclusive focus on relative segment motion. Similarly, no effort has been made to control for the potentially considerable effect of walking velocity on dynamic arch motion.

When combined with available arch kinematic data, fluoroscopic examination of sagittal plane calcaneal motion performed by Wearing et al. (1999) raises many questions regarding the motion of the segments that comprise the arch. It is difficult to interpret the striking similarity in timing and magnitude of calcaneal motion (Wearing, Urry et al. 1999) and the timing and magnitude of arch motion collected on the same set of subjects (Wearing, Urry et al. 1998). When taken in the context of the truss and tie-rod mechanism of arch function, such a relationship would seem improbable because of the supposed proportionate contribution of the proximal and distal segments. Though the results of Wearing et al. (1998; 1999) provide new insight, the respective roles of calcaneal and first metatarsal motion in producing arch motion remain unclear.

While examining the effect of ankle fusion and arthroplasty on dynamic foot function it was observed that arch motion occurred in a manner different than commonly accepted. By examining arch kinematics not only in terms of the first metatarsal relative to the calcaneus (arch range of motion), but also as motion of the first metatarsal relative to the floor a new perspective on dynamic foot motion was uncovered. The purpose of this work is to define the segmental kinematics of the first metatarsal and calcaneus that expose this new perspective on arch motion during gait.

Methods

Subjects

Data was analyzed for a total of 23 limbs from a study designed to assess dynamic foot function following ankle fusion (n=6) or arthroplasty (n=9) as compared to controls (n=8). For inclusion in the study all subjects were at minimum one year from their most recent surgical intervention and had fully returned to activity with reported pain of less than 2/10 at worst. All ankle fusion subjects had undergone tibio-talar fusion or arthroplasty for isolated ankle arthritis of traumatic origin. Ankle replacement was performed using the Scandinavian Total Ankle Replacement (S.T.A.R.) method for all arthroplasty subjects. Control limbs were comprised of the un-involved contralateral limbs that were pain free during daily activities. Data was not collected on the un-involved limb for 7 subjects resulting in a total of eight control limbs. Control limbs were evenly distributed between S.T.A.R. (n=4) and fusion patients (n=4). Subjects were on average 65 ± 13 years old with a mean weight and height of 89± 15 Kg and 170 ± 8 cm respectively.

Procedure

An Optotrak motion analysis system (Optotrak 3020, Northern Digital Inc.) was used to record the 3-D position of the first metatarsal and calcaneus during gait. Marker data was collected at a rate of 120Hz. Motion of the first metatarsal was tracked with a lightweight marker triad mounted on the first metatarsal, medial to the extensor hallucis longis tendon (Leardini, Benedetti et al. 1999; Umberger, Nawoczenski et al. 1999; Cuddeford 2000). Hindfoot motion was assessed using two markers on the lateral calcaneus proximal to the calcaneal fat pad and one on the posterior aspect of the calcaneus. A digitizing process in conjunction with lateral and AP x-rays of the foot in the patient population, and palpation of the foot in control subjects were used to identify the location of underlying bony geometry relative to each segment's marker triad. For the first metatarsal segment the local coordinate system was adjusted to align with the long axis of the first metatarsal in the sagittal and transverse planes to provide an anatomical reference frame for determining segment motion relative to the floor and calcaneus. Due to the inability to determine frontal plane orientation of the first metatarsal the medio-lateral axis was fixed parallel to the floor. For the calcaneal segment the AP axis was aligned with the long axis of the foot, ML axis was parallel to the floor with the vertical axis orthogonal to the other two.

An overhead tracking system was used to ensure that subjects ambulated at the controlled speeds of 0.9, 1.1, 1.3 and 1.6 m/s as well as self selected walking velocity. A Kistler force plate (Kistler, Inc.) sampling at a rate of 360 Hz was used to determine vertical ground reaction forces for normalization of kinematic data to 100% percent stance phase. Heel strike and toe off events were determined using a 5N threshold for vertical ground reaction force values. Visual 3D (C-motion Inc.) was used to calculate 3-D displacements. To determine arch kinematics during gait, motion of the first metatarsal relative to the calcaneus (arch motion) and sagittal plane angular motion of the first metatarsal in the global (lowering of the proximal end of the first metatarsal) were calculated.

Definitions

Arch range of motion is defined as the total range of movement of the first metatarsal relative to the calcaneus in the sagittal plane. Range values were determined from heel strike until the point of peak 1MT dorsiflexion representing what is also referred to as peak "arch elongation". This interval was chosen as it was felt to best represent motion of the arch under fully loaded conditions while avoiding undue variability associated with the minimally loaded condition at the end of pre-swing.

First metatarsal lowering is defined as the change in angular inclination of the first metatarsal relative to the floor (global) in the sagittal plane. This variable was chosen to represent the magnitude of inclination change of the first metatarsal that would result in a lowering of the proximal end of the bone. This range was calculated using the difference between the peak first metatarsal plantar flexion value following forefoot contact and the peak relative dorsiflexion of the first metatarsal relative to the floor for the remainder of the stance phase.

First metatarsal inclination was determined using the peak first metatarsal dorsiflexion value that was used to calculate first metatarsal lowering. This value represents the lowest angle of inclination of the first metatarsal relative to the floor during the mid-stance phase of gait. To provide insight into how this value relates to quiet standing the mean inclination value obtained during a quiet standing calibration trial was subtracted to provide a difference value.

The angular velocity of the first metatarsal relative to the floor in the sagittal plane was used to determine the timing of first metatarsal motion. A velocity threshold of 50 deg/sec was chosen for determining the percent of stance phase at which the first metatarsal slowed its velocity relative to the floor at forefoot contact as well as the point at which it began its forward (plantarflexion) rotation as the foot rises from the floor at the end of terminal stance.

Data Analysis

Mean and standard deviation values were calculated for the variables of interest. A two way ANOVA (α=.05) with one crossover and one nested factor was used to evaluate for effects of velocity and group, respectively. Paired t-tests were used to evaluate between group and between velocity differences when appropriate. To prevent inflation of type I error a Bonferroni correction was used.

Results

Arch Range of Motion

Arch range of motion from FF to PAE (Figure 2-1) was very consistent both between and within walking velocity conditions for all subjects (Table 2-1). Group or walking velocity were not found to have a significant effect on arch range of motion ($p=0.86$ and 0.28 respectively). The mean standard deviation value for the 5 trials at each walking velocity (within) was less than one degree on average for subjects in all three groups. Similarly, the mean (± 1 SD) standard deviation when comparing within a subject between walking velocity conditions was 0.58 (0.47), 0.68 (0.38) and 0.66 (0.15) degrees for the fusion, S.T.A.R. and control groups respectively.

First Metatarsal Lowering

The magnitude of lowering of the proximal end of the first metatarsal (Figure 2-2) is very similar between groups and across walking velocities (Table 2-2). A significant effect of walking velocity was observed in the ankle fusion group, however, the magnitudes of differences were less than one degree on average.

First Metatarsal in Global Relative to Standing

The smallest angle between the first metatarsal and the floor during midstance was compared to the angle that was observed during quiet standing. The angular difference between the two was typically less than one degree (Table 2-3) and not significantly affected by walking velocity or surgical intervention.

Timing of First Metatarsal Motion

The first metatarsal was observed to come to a static resting position at approximately 17 percent of the stance phase (Table 2-4, Figure 2-2, 2-3, 2-4, 2-5). A significant effect of walking velocity was observed ($p<.001$) with the first metatarsal reaching a stable orientation later in the stance phase as walking velocity increased. The first metatarsal began its forward rotation at about 67 percent of the stance phase with forward rotation beginning earlier ($p<.001$) as walking velocity increased (Table 2-5).

Discussion

Arch Motion

In spite of the considerable interest in function of the medial longitudinal arch no prior study has used three-dimensional analysis to specifically assess in-vivo motion of the first metatarsal relative to the calcaneus and the floor during the stance phase of gait. In the present study this analysis was performed at self selected, as well as controlled walking velocities providing insight into the effect of walking velocity on these motions. Such assessment of arch and first metatarsal motion has served to provide new insight into dynamic foot function. These results indicate that the current use of the truss and tie-rod theory does not adequately describe in-vivo arch motion and that arch motion is independent of walking velocity.

The kinematics of the first metatarsal relative to the calcaneus presented here (Table 2-1, Figure 2-1) are in strong agreement in pattern, timing and magnitude with the comparable data available in the literature (Wearing, Urry et al. 1998; Cashmere, Smith et al. 1999; Leardini, Benedetti et al. 1999; Hunt, Smith et al. 2001; Hunt, Smith et al. 2001; MacWilliams, Cowley et al. 2003; Wearing, Smeathers et al. 2003). Consistent with the current understanding of arch motion, following heelstrike the first metatarsal was found to dorsiflex relative to the calcaneus approximately 10 degrees (arch elongation) until late in the stance phase. Late in the stance phase the metatarsal-phalangeal joint then extends as the whole foot rotates forward activating the windlass mechanism and increasing arch height.

Simultaneous assessment of arch and first metatarsal kinematics provides a unique insight into dynamic foot function during the stance phase of gait. Because of the long standing use of the truss and tie-rod model to represent the medial longitudinal arch it is commonly assumed that both the distal calcaneus and the proximal first metatarsal lower to the floor during the stance phase of gait. Application of this simple mechanical analog would suggest that arch motion during mid-stance phase of gait would be achieved by nearly proportional contributions of the proximal and distal foot. This is not, however, supported by the minimal lowering of the first metatarsal during the stance phase of gait observed in this study (Table 2-2). Because only three individuals (4/23 limbs) had first metatarsal lowering of greater than 2 degrees it calls into question the use of the truss and tie rod model (or cadaveric analog) vertically loaded at its apex with corresponding loading at the hind and fore foot to represent midstance foot motion.

The finding of a nearly static first metatarsal inclination from just after forefoot contact until nearly 70% of stance phase (Table 2-4, Table 2-5, Figure 2-2) has considerable implications for how dynamic foot motion is understood. This relative lack of motion is particularly relevant in light of the continuous arch motion that occurs during this same interval (Figure 2-1, Figure 2-2). The combination of arch mobility and first metatarsal stability indicates that the motion is driven by the calcaneus as it rotates relative to the nearly stationary first metatarsal. The idea of arch motion occurring during the mid-stance phase of gait driven primarily by rotation of the calcaneus fits well with, and serves to explain, the interesting findings of Wearing et al (Wearing, Urry et al. 1998; Wearing, Urry et al. 1999). Based on these results it appears that as ground reaction forces move anteriorly on the foot and the ankle moment increases (Hunt, Smith et al. 2001; MacWilliams, Cowley et al. 2003) the considerable pull of the gastroc-soleus complex serves to gradually elevate, rotate, and unload the calcaneus (Warren, Maher et al. 2004). Though further study to understand this mechanism is necessary, such a mechanism of arch elongation is plausible when in-vivo plantar pressure, EMG and kinematic data as well as comparative anatomy are also considered.

Gradual elevation and associated unloading of the calcaneus is supported by the assessment of calcaneal loading as presented by Warren et al. (2004) among others. It is well recognized that forces peak under the heel just as the forefoot reaches the floor. It is also important to note that these peak forces quickly decline by mid stance phase. This is demonstrated in plantar pressure data (Warren, Maher et al. 2004) as well as foot switch data that suggests the heel is completely off of the floor between 56% and 62% of stance phase (Cornwall and McPoil 1999; Hunt, Smith et al. 2001; Hunt, Smith et al. 2001; Hunt and Smith 2004). By comparison our data suggests that the first metatarsal does not begin its forward rotation until as late as 70% of stance phase at normal walking speeds with peak arch elongation occurring even later. It is particularly interesting when the timing of these events are compared with available EMG data of the intrinsic muscles of the foot (Mann and Inman 1964). It is clear that that intrinsic musculature activation begins not long after forefoot contact with activation of abductor hallucis, flexor digitorum brevis, and interossei muscles beginning just as the heel would be elevated from the walking surface. All intrinsic muscles tested (Mann and Inman 1964) demonstrated activation from the time when the heel would be elevated from the walking surface until toe-off. The additional support of intrinsic musculature prior to the full activation of the windlass mechanism via forward rotation of the metatarsals would likely be necessary to assist ligamentous structures in maintaining foot stability as the calcaneus is raised from the floor.

Comparative anatomy combined with static in-vivo testing may provide additional support for the importance of midfoot mobility allowing the calcaneus to rise while the distal aspect of the foot is stationary. In the early work by Close and Inman (1967) they address the foot motion of the chimpanzee emphasizing the considerable differences in motion as compared to humans. They state: “The tarsals of the chimpanzee are arranged for the most part in the horizontal plane. This results in a series of joints very difficult to control. The chimp foot characteristically “breaks”, the heel rising from the walking surface while the structures distal to the transverse tarsal joint are still flat on the ground”(Close, Inman et al. 1967). Though human arch motion is small in comparison to that of primates, the concept of first metatarsal and forefoot stability with relative calcaneal motion is supported by the kinematic data presented here. Such discussion of foot motion is also consistent with the recent emphasis on the importance of transverse tarsal joint motion (specifically talonavicular joint) on dynamic foot function. In static in-vivo work by Lundberg et al (Lundberg, Svensson et al. 1989; Lundberg, Svensson et al. 1989) the talo-navicular joint was found to allow the greatest total rotation of all foot joints. If arch elongation occurs in the manner proposed it would likely necessitate considerable talo-navicular joint motion for calcaneal rotation to occur while the first metatarsal remains stationary.

The findings presented here suggest that the current use of the truss and tie-rod model does not adequately represent the dynamics of the foot during gait. Though the truss and tie-rod model can be a useful mechanical analog for a simplistic understanding of complex foot motion the location and effect of the applied loads used to represent gait should be altered from its current use. Rather than representing the foot as a passive structure with axial loading at the talocrural joint resulting in a lowering of the apex the importance of the plantarflexors as a driving force should be acknowledged. Shifting the hindfoot loading from the plantar surface of the calcaneus to the pull of the Achilles tendon as the tibia progresses over the foot during stance would produce motion consistent with the results obtained in this study.

Walking Velocity Independence

The excellent repeatability of all subjects both within and between walking velocity conditions provides added confidence to these results and additional insight into the consistency of arch motion. As anticipated we observed very high repeatability within each walking velocity condition due to the minimal soft tissue covering of the first metatarsal resulting in minimal error associated with motion of the skin relative to the underlying bone (Tranberg and Karlsson 1998; Umberger, Nawoczenski et al. 1999). We were, however, surprised to find remarkable consistency of arch kinematics over a wide range of walking velocities. It was our expectation that as walking velocity and corresponding loading increased(Warren, Maher et al. 2004) the magnitude of the arch motion would increase. This is clearly not the case with very minimal changes in arch kinematics observed over the wide range of velocities tested (Table 2-1). Though a significant effect of walking velocity was observed for first metatarsal lowering the magnitude of change is small and likely not of clinical relevance. Due to the very high consistency across velocities and repeated measures design even such small differences reached significance.

The mechanism and implications of this remarkable consistency across a wide range of walking velocities are not yet known. Factors such as bony, ligamentous and soft tissue constraint likely all play an important role in the consistent arch elongation observed. Though there are many contributory factors the plantar fascia is an obvious first choice in identifying a structure of importance. Given its stiffness and important role in maintaining the integrity of the arch it more than any other structure could act to provide such consistent motion. The rotation of the calcaneus relative to the first metatarsal would gradually increase tension in the plantar fascia transferring loading to the toes via the windlass effect (Warren, Maher et al. 2004). As the first metatarsal and forefoot begin their forward rotation the plantar fascia would halt arch elongation and lead to the rapid increase in arch height seen at the end of stance phase.

Though the kinematics of the arch are not influenced, the timing of first metatarsal motion is affected by changes in walking velocity. As walking velocity increases the first metatarsal becomes stationary later and begins its forward rotation earlier as a relative percent of the stance phase (Table 2-4, Table 2-5). Such a change in the timing of motion is likely the effect of alterations in stride length. It appears as though in an effort to achieve higher speeds the use of the heel and forefoot rockers increase so that it takes longer to reach foot flat and forward rotation of the foot occurs earlier as a percentage of the stance phase.

It was previously accepted that as the foot is loaded during the stance phase of gait both the proximal first metatarsal and distal calcaneus underwent considerable lowering to the floor resulting in arch elongation or lowering. While examining the effect of ankle fusion and arthroplasty on dynamic foot function it was observed that arch motion occurred in a manner different than commonly accepted. The examination of segmental (motion of the first metatarsal relative to the floor) as well as intersegmental (first metatarsal relative to the calcaneus) foot motion has served to provide new insights into the dynamics of arch motion.

Summary

1. Motion of the medial longitudinal arch is walking velocity independent.
2. Motion of the arch was not altered by ankle fusion or arthroplasty in the population tested.
3. The first metatarsal maintains a nearly static position for nearly 50% of the stance phase.
4. The inclination of the first metatarsal during mid-stance is nearly identical to that observed during quiet bilateral standing.
5. The heel rotates relative to the first metatarsal resulting in arch elongation.
6. The timing of first metatarsal motion is influenced by walking velocity with foot flat coming later and forward rotation beginning earlier as velocity increases.

Table 2-1. Mean (± 1 SD) arch range of motion from heel strike to peak first metatarsal dorsiflexion (degrees).

Group	Self Selected	0.9 m/s	1.1 m/s	1.3 m/s	1.6 m/s
Fusion	10.93 ± 3.58	11.16 ± 3.84	10.93 ± 3.35	10.54 ± 3.3	10.28 ± 4.02
S.T.A.R.	10.36 ± 2.98	10.89 ± 3.45	10.06 ± 2.31	10.08 ± 2.53	9.1 ± 2.37
Control	11.8 ± 4.73	10.38 ± 2.56	10.19 ± 2.67	10.24 ± 2.17	10.31 ± 2.51

Table 2-2. Mean (± 1 SD) maximal lowering of first metatarsal relative to floor during mid-stance phase (degrees).

Group	Self Selected	0.9 m/s	1.1 m/s	1.3 m/s	1.6 m/s
Fusion	1.98 ± 2.28	2.33 ± 1.89	1.91 ± 1.83	1.62 ± 1.76*	2.01 ± 2.25*
S.T.A.R.	1.46 ± 1.04	1.52 ± 1.07	1.11 ± 0.81	1.15 ± 1.09	1.33 ± 1.39
Control	1.61 ± 0.91	1.55 ± 0.66	1.44 ± 0.69	1.27 ± 0.52	1.31 ± 0.41

*Significantly different (p <.005) than 0.9 m/s.

Table 2-3. Mean (± 1 SD) angular difference between 1MT angle during quiet standing and mid-stance (degrees).

Group	Self Selected	0.9 m/s	1.1 m/s	1.3 m/s	1.6 m/s
Fusion	-0.59 ± 1.73	-0.29 ± 1.64	-0.45 ± 1.67	-0.25 ± 1.72	-0.87 ± 1.73
S.T.A.R.	0.36 ± 1.60	0.32 ± 1.37	0.18 ± 1.47	0.45 ± 1.34	0.49 ± 1.44
Control	-0.77 ± 1.14	-1.03 ± 1.12	-1.04 ± 1.05	-1.12 ± 1.28	-0.73 ± 1.19

Table 2-4. Mean (± 1SD) percent stance phase for initiation of nearly stationary first metatarsal.

Group	Self Selected	0.9 m/s	1.1 m/s	1.3 m/s	1.6 m/s
Fusion	16.56 ± 3.86	16.97 ± 2.97	18.39 ± 2.96	19.24 ± 2.99*#	19.66 ± 3.38*#
S.T.A.R.	14.61 ± 3.81	15.29 ± 3.54	16.80 ± 3.24#	17.48 ± 3.15*#	17.09 ± 3.98*#
Control	15.88 ± 3.78	16.09 ± 2.15	17.50 ± 2.17	17.98 ± 2.31	19.22 ± 2.40*#

* Significantly different (p <.005) than 0.9 m/s

\# Significantly different (p <.005) than self selected.

Table 2-5. Mean (± 1SD) percent stance phase of initiation of forward (plantarflexion) first metatarsal rotation.

Group	Self Selected	0.9 m/s	1.1 m/s	1.3 m/s	1.6 m/s
Fusion	69.42 ± 7.43	70.37 ± 6.49	68.78 ± 7.05	68.78 ± 7.81	65.07 ± 5.44
S.T.A.R.	67.57 ± 8.17*#	69.26 ± 5.80*	65.63 ± 5.86*	65.48 ± 6.52*	59.37 ± 9.23
Control	68.04 ± 4.43*	68.73 ± 4.10*	67.07 ± 3.74	66.11 ± 5.15	63.35 ± 5.68

* Significantly different ($p < .005$) than 1.6 m/s

Significantly different ($p < .005$) than 1.3 m/s

Figure 2-1. Ensemble average sagittal plane kinematics of the first metatarsal relative to the calcaneus during the stance phase of gait at all five walking speeds. Note that SSWV, 1.1 m/s and 1.3 m/s kinematic curves are overlapped and nearly indistinguishable.

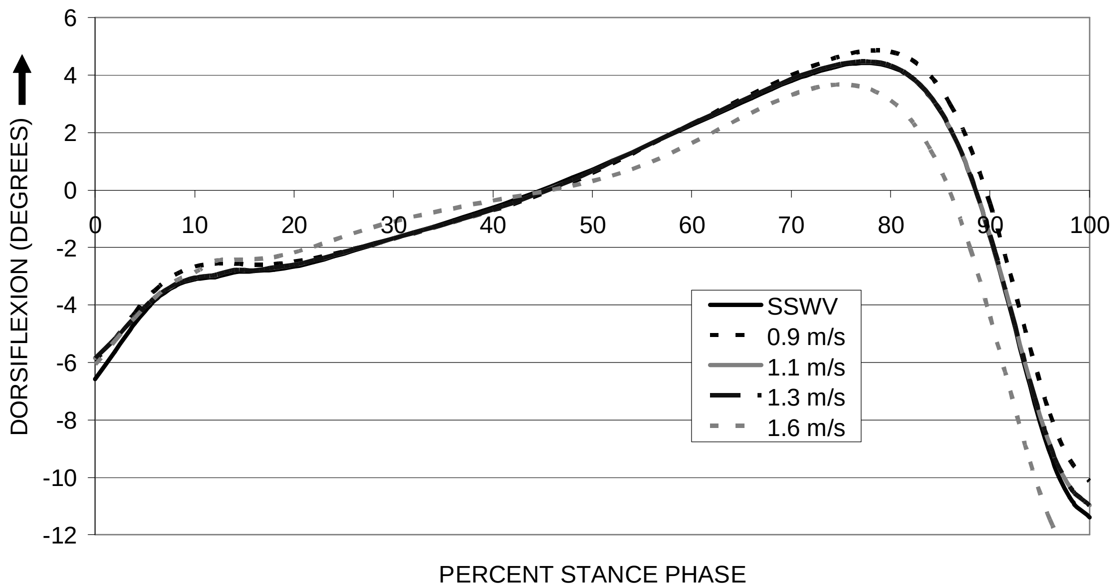


Figure 2-2. Ensemble average (± 1 SD) sagittal plane kinematics of the first metatarsal relative to the floor during the stance phase of gait.

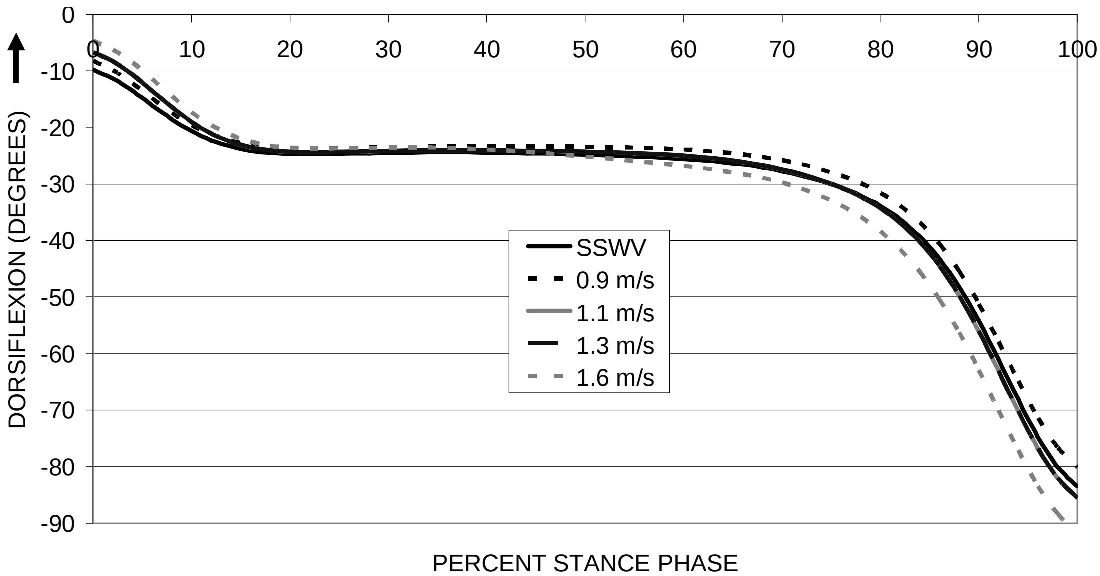


Figure 2-3. Mean sagittal plane kinematics of the first metatarsal relative to the floor during the stance phase of gait at all five walking speeds for all ankle fusion subjects.

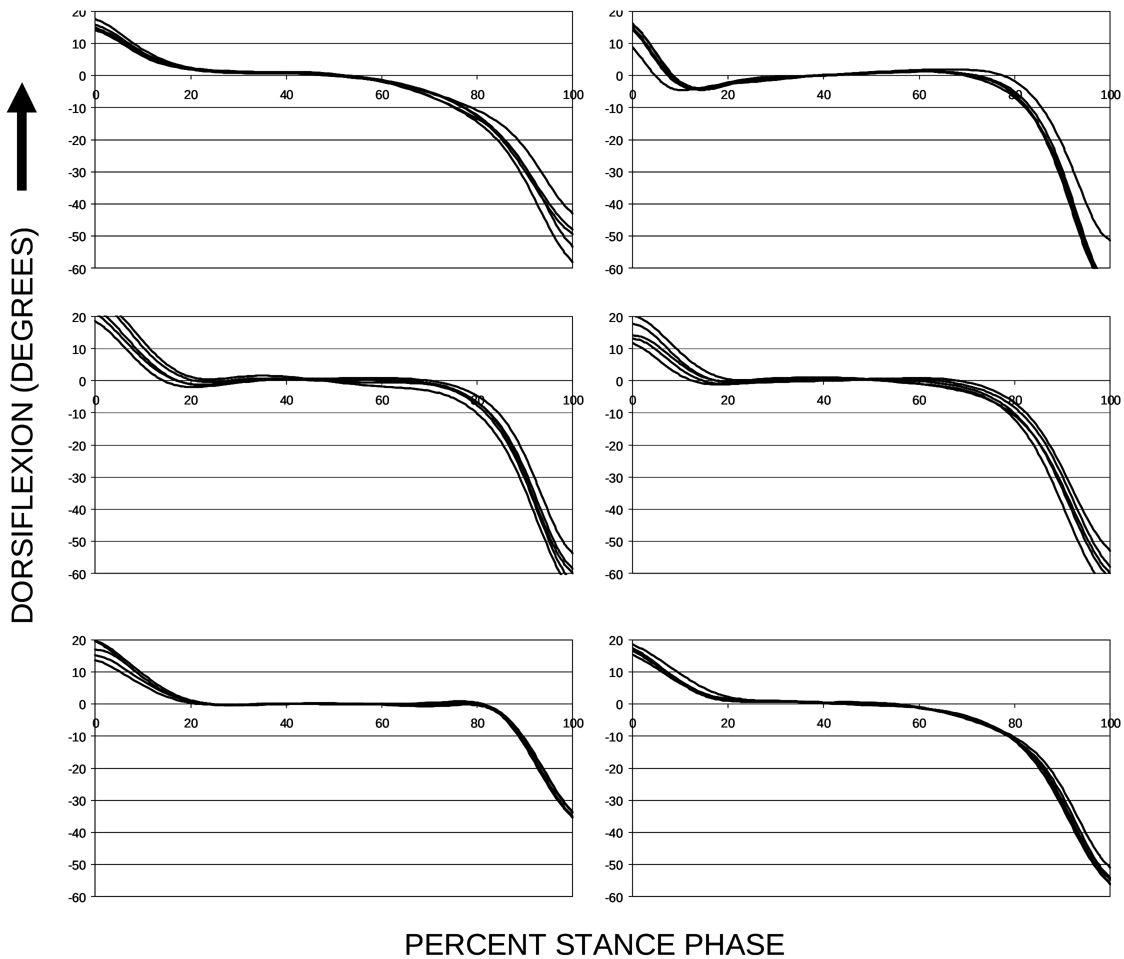


Figure 2-4 Mean sagittal plane kinematics of the first metatarsal relative to the floor during the stance phase of gait at all five walking speeds for all ankle replacement (S.T.A.R.) subjects.

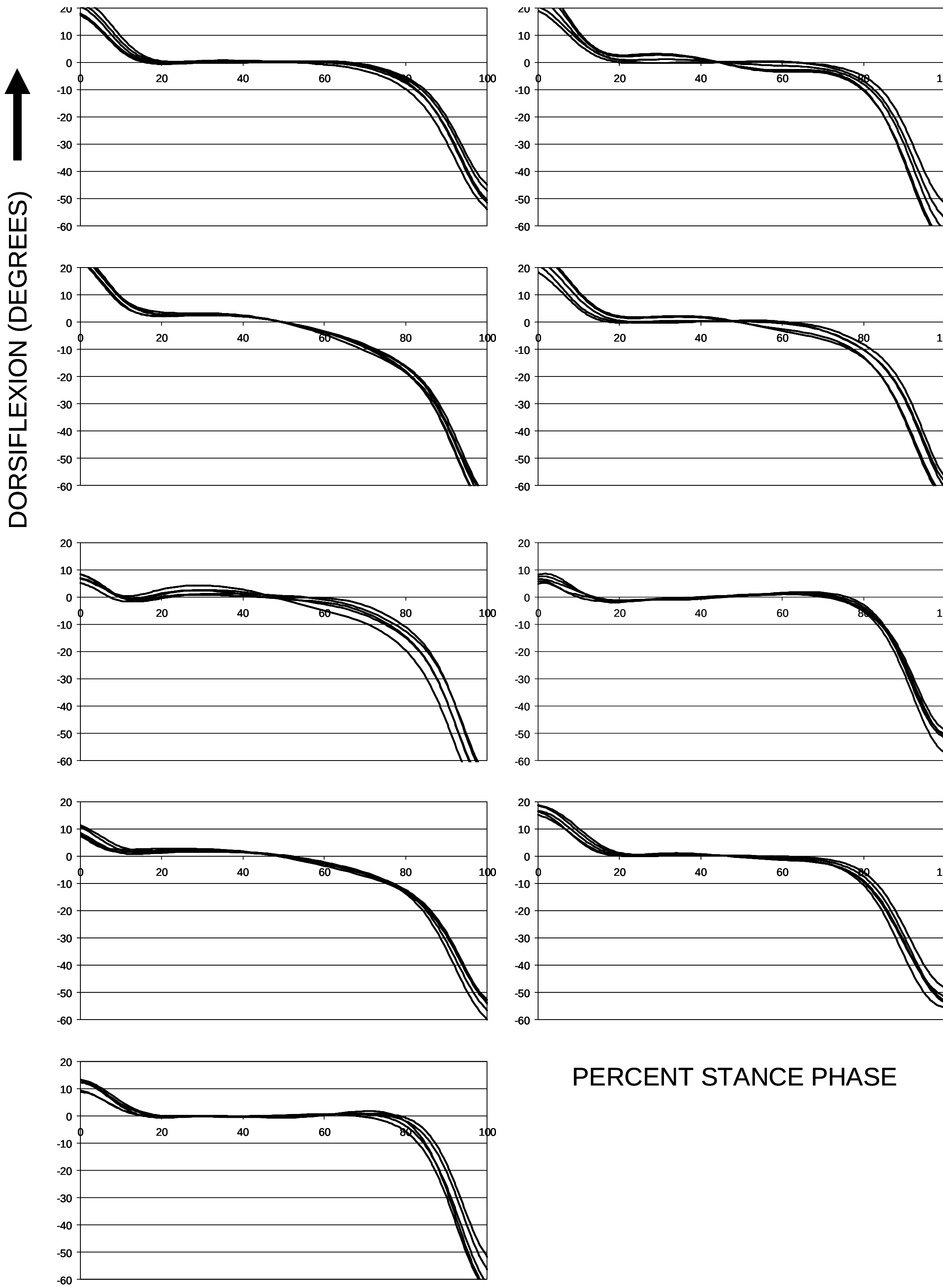
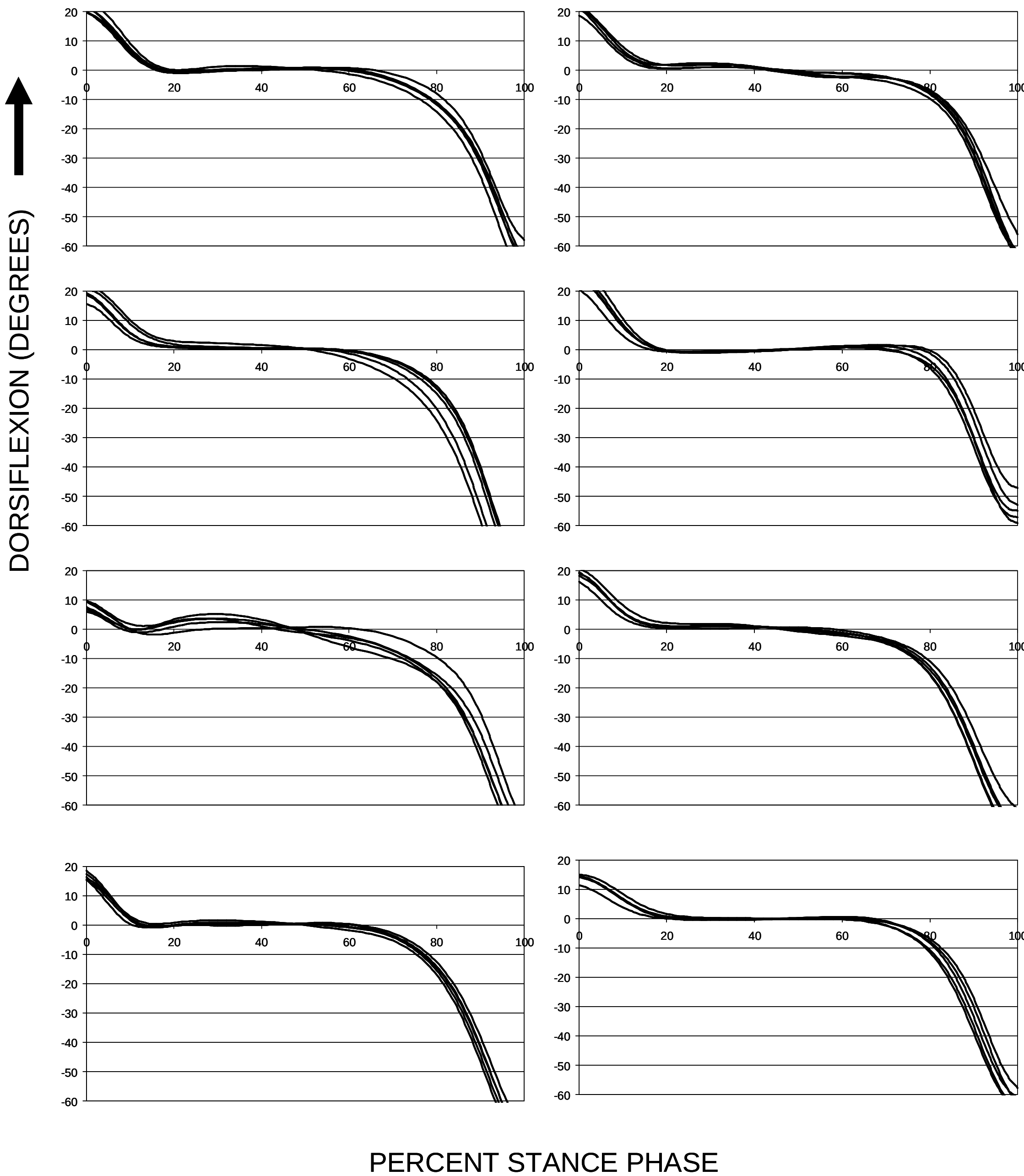


Figure 2-5. Mean sagittal plane kinematics of the first metatarsal relative to the floor during the stance phase of gait at all five walking speeds for all control limbs.



CHAPTER III

METHODS

Subjects

Seventeen physically active subjects between the ages of 18 and 45 with no current history of lower extremity pain were recruited to participate in this study. Informed consent was obtained from all subjects prior to participation. To ensure that subjects with a sufficient range of arch heights were represented only subjects with self reported high or low arches were recruited. Subject participation was regulated to ensure that a minimum of four participants in the study demonstrated arch heights greater, and four had arch heights less than one standard deviation from the mean value observed in a clinical patient population (Saltzman, Nawoczenski et al. 1995). Due to the likelihood of increased variability in foot structure in a patient population this benchmark was selected to ensure subjects with a wide range of arch heights were included.

For subjects meeting this initial screening criteria information including height, weight, age, gender, occupation and past history of lower extremity injuries was recorded. Subjects all had weight bearing AP, lateral and hindfoot alignment view radiographs taken of the foot.

Radiographs

A total of two weight bearing radiographs were taken of the foot tested in a manner consistent with common clinical practice (Saltzman, Nawoczenski et al. 1995). Antero-posterior and lateral radiographs of the foot (Saltzman, Brandser et al. 1994; Saltzman, Nawoczenski et al. 1995) as well as a hindfoot alignment view radiograph (Saltzman and el-Khoury 1995) were used to determine the static foot alignment for each subject as well as allow anatomically based alignment of local coordinate systems for the foot. The angle between the dorsal surface of the first metatarsal (1MT) and the inferior border of the calcaneus (CIA) (Saltzman, Nawoczenski et al. 1995), chosen to represent arch height, served as the primary measure representing foot architecture in the subjects to be tested.

Procedures

The effect of walking velocity on plantar loading and muscle activity can be considerable and therefore controlling for the effect of walking velocity is desirable (Burnfield, Few et al. 2004; Warren, Maher et al. 2004). Each subject ambulated barefoot at a controlled walking velocity of .78 satures/second as suggested by Morag et al. (1999). An overhead timing system was used to guide subjects to ambulate at the appropriate velocity. Only trials within ± 5% of chosen walking velocity were used for analysis. A total of 5 trials meeting the velocity criteria during which footstrike was within the bounds of the force plate and the subject did not appear to target the force plate was used to determine kinematic peak and range values.

A five segment rigid-body model including the foot and lower leg was used to assess dynamic foot function during gait for subjects meeting the studies inclusion and exclusion criteria. Three infrared light emitting diodes (IREDs) placed to minimize skin-bone movement (Tranberg and Karlsson 1998; Umberger, Nawoczenski et al. 1999; Houck, Yack et al. 2004; Longworth, Chockalingam et al. 2005) were tracked (Optotrak 3020, Northern Digital Inc.) to determine the three dimensional kinematics of the hallux, first metatarsal, lateral forefoot, calcaneus and tibia during the stance phase of gait. Move 3D software (C-Motion Inc.) was used to determine segment and joint angles and velocities using an Euler angle approach. Kinematic data was collected at 120 Hz and low pass filtered using a 6Hz cutoff frequency. System tracking capabilities were previously examined indicating precision of <0.1mm at velocities similar to those observed during gait (unpublished data). Vertical ground reaction forces collected via a Kistler force plate (Kister Instrument Corp., Model 9865B) at a sampling rate of 360 Hz and synchronized with kinematic data were used to determine heelstrike and toe off events. To prevent targeting of the force plate while walking they were not informed of its location and were required to look straight ahead while walking.

Lower Extremity Modeling

The foot model used is a modified version of the methods used by Leardini et al. (1999), Cuddeford et al. (2000) and Wilken (Chapter II). The location of IREDs for each segment of the model are identified in Table 3-1.

Local Coordinate System (LCS) definitions for each segment are presented in Table 3-2. A digitizing process in conjunction with palpation and lateral, AP and hindfoot alignment x-rays of the foot were used to identify the location of underlying bony geometry relative to each segment's marker triad. All segment orientations determined using radiographic methods are denoted using "X-ray" in Table 3-2. The local coordinate systems for the first metatarsal and forefoot segments were aligned with the long axis of the first and second metatarsals respectively using AP and lateral x-rays. Prior to radiographic examination two lead beads were placed over the proximal and distal first and second metatarsals. These lead beads, aligned with the long axis of the bone via palpation, corresponded to points digitized relative to the three markers for their respective segments. Rotations of the local coordinate systems in the global (room) coordinate system were used to account for orientation differences between a line connecting the two lead beads (digitized points) and the long axis of the bone as determined radiographically (Figure 3-1). These angular adjustments were made in the sagittal and transverse planes based on measurements taken from lateral and AP x-rays, respectively.

Data Analysis

Kinematic and kinetic events used to determine the intervals of interest are listed in Table 3-3. All kinematic data was normalized to 100% stance phase (heel strike-toe off) for qualitative inspection and examination of trial-to-trial variability while mean data representing each subject was calculated to examine intersubject variability. Range values for use with hypothesis one were determined by calculating the difference between the maximum and minimum kinematic values for the three variables of interest during the interval from FF to PAE. This was performed for each trial with the mean of the trials used to represent the motion present for each subject. A correlational analysis was performed to determine the association between sagittal, frontal and transverse plane kinematics and arch height as outlined for specific aim number one. Normality of the data was verified using Kolmogrov-Smirnov tests. To ensure that the association was properly modeled both linear and quadratic analysis was performed. After linear regression was performed a quadratic term was then added to determine if its addition considerably increased the predictive value of the regression model. The r-squared values from the linear and quadratic regression analysis was compared to determine the most appropriate model for explaining the association between the two variables of interest.

To determine coupling ratios as discussed in hypothesis two, kinematics for the two variables of interest in each comparison were plotted relative to each other from FR to PAE using an angle-angle plot (Nigg, Cole et al. 1993; Stacoff, Nigg et al. 2000). This was performed for each trial and the slope of the linear regression line fit to the data for each individual trial was used to represent a coupling ratio for that trial. The median of the five trials for each subject was used to represent the coupling ratio demonstrated by that subject. A correlational analysis was then used to determine the relationship between coupling ratio and arch height in a manner similar to that used in hypothesis one.

A power analysis was performed using custom software (http://www.stat.uiowa.edu/~rlenth/Power) to determine the appropriate sample size for the linear regression analysis. Based on an alpha level of .05 and power value of .80 a sample size of 16 subjects was deemed appropriate (1 predictor, error SD=1, arch height SD=10, effect size of 3 degrees over a range of 40 degrees arch height). Similarly when a total of

16 subjects are used an r value of .5 will reach significance (Pearson r product moment correlation coefficient; df =14, alpha =0.05, rcrit =0.497)(Portney and Watkins 1999). Correlation values of less than .5 were deemed not of clinical significance and therefore, even if statistically significant were not addressed as relevant.

Repeat testing was performed on three subjects of differing arch heights to demonstrate repeatability of the measures obtained. Each subject was tested twice with a minimum of two days in between testing sessions. Intraclass correlation coefficient (ICC) and standard error of the estimate were used to determine repeatability of the measures of interest.

Limitations and Assumptions

1. Because only individuals without current lower extremity problems were included in the study, the tested population may be biased to represent individuals who are able to successfully accommodate to their foot structure.
2. Because of the young active population tested the results may not be generalizable to older or more sedentary individuals.
3. The markers used to track motion of the segments may have moved relative to the underlying bone due to the effect of skin-bone movement adding noise to the kinematic data obtained.
4. Contrary to pilot data and current theory, foot structure (arch height) may not influence the variables chosen to evaluate dynamic foot function during gait.

Table 3-1. Marker Placement.

Segment	Marker	Location
Hallux (HX)	4 5 6	Mounted on triad on dorsal surface of proximal phalanx Mounted on triad on dorsal surface of proximal phalanx Mounted on triad on dorsal surface of proximal phalanx
First Metatarsal (1MT)	1 2 3	Mounted on triad on dorsal medial surface of first metatarsal Mounted on triad on dorsal medial surface of first metatarsal Mounted on triad on dorsal medial surface of first metatarsal
Forefoot (FORE)	7 8 9	Proximal end of 2nd metatarsal Distal end of 2nd metatarsal Distal end of 5th metatarsal
Calcaneus (CALC)	10 11 12	Posterior surface of calcaneus Lateral aspect of calcaneus superior to calcaneal fat pad Lateral aspect of calcaneus superior to calcaneal fat pad
Leg (TIB)	13 14 15	Medial surface of tibia Medial surface of tibia Medial surface of tibia

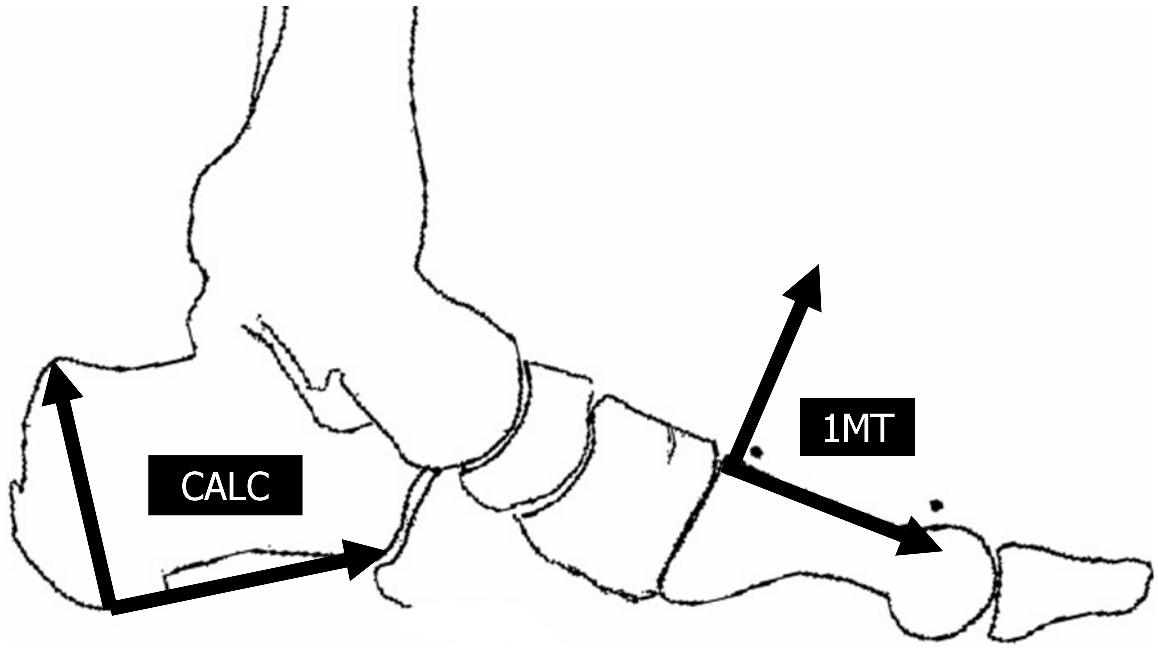
Table 3-2. Local Coordinate System definitions.

Segment	Axis	Definition
Leg (TIB)	X	Orthogonal to Y and Z axes
	Y	Passing through midpoint of femoral condyles and malleoli
	Z	Aligned with malleoli and passing through mid malleolar point (orthogonal to Y)
Calcaneus (CALC)	X	Posterior heel to midpoint of foot at the level of the 5th MT flair. Inclined to match sagittal inclination of plantar surface of calcaneus (CIA)(X-ray)
	Y	Determined using palpation and verified using X-ray
	Z	Orthogonal to X and Y axes
Forefoot (FORE)	X	Aligned with long axis of second metatarsal in frontal and sagittal planes (X-ray)
	Y	Orthogonal to X and Z axes
	Z	Orthogonal to long axis of 2nd MT and parallel to the floor
First Metatarsal (1MT)	X	Aligned with long axis of first metatarsal in frontal and sagittal planes (X-ray)
	Y	Orthogonal to X and Z axes
	Z	Parallel to floor and orthogonal to X axis
Hallux (HX)	X	Middle of first metatarsophalageal joint to midpoint of distal phalanx.
	Y	Orthogonal to X and Z axes
	Z	Aligned parallel to the floor

Table 3-3. Definitions of gait events.

Event		Definition	Percent Stance
Heelstrike	HS	Vertical Ground reaction forces rise above 5N	0%
Forefoot Contact	FF	1MT rotation in sagittal plane with respect to global coordinate system of less than 50 deg/s	10-20%
First Rotation	FR	1MT rotation in sagittal plane with respect to global coordinate system of greater than 50 deg/s	60-70%
Peak Arch Elongation	PAE	Point of peak dorsiflexion of 1MT relative to CALC	75-80%
Toe Off	TO	Vertical Ground reaction forces fall below 5N	100%

Figure 3-1. Representation of medial longitudinal arch structure as observed with a lateral x-ray of the foot. Lateral x-rays were used to determine sagittal plane local coordinate system orientation for the first metatarsal (1MT), forefoot (not shown) and calcaneus (CALC).



CHAPTER IV
THE EFFECT OF ARCH HEIGHT ON FOOT MOTION DURING GAIT

Introduction

As the most prominent structural feature of the foot the medial longitudinal arch has long been identified as playing a key role in the structural integrity of the foot. The conventional understanding of the foot suggests that deviations from normal arch height results in hypo or hyper mobility affecting stability or mobility during gait, where the demands on the foot change throughout the cycle. Results of studies examining the effect of foot structure on mobility are, however, inconclusive or contradict this presumption, drawing into question this commonly held belief (Cashmere, Smith et al. 1999; Hunt, Fahey et al. 2000; Hunt and Smith 2004). Despite decades of foot and ankle research, large gaps exist in our knowledge of dynamic foot function and many conflicting opinions regarding the consequences of abnormal arch height add uncertainty to the treatment of foot and ankle pathology.

The current understanding of foot function has primary evolved from mechanical models (Hicks 1954; Hicks 1955; Olerud and Rosendahl 1987; Sarrafian 1987; Fuller 2000; Van Boerum and Sangeorzan 2003) that have been generated to represent the structures and mechanisms that account for the normal balance between foot mobility and stability during the stance phase of gait. Based on these mechanical representations of foot structure and function the medial longitudinal arch, typically represented by arch height, is believed to be a key determinant of foot mobility during gait (Kirby 1989; Arangio, Chen et al. 1998; Van Boerum and Sangeorzan 2003; Wang and Crompton 2004). Segmental and articular orientations, associated with normal arch structure, are believed to be necessary to allow normal muscular function, appropriately accommodate varied terrain, attenuate impact loads and maintain mobility within normal articular ranges. It is commonly assumed that as arch height decreases articular geometry, supporting soft tissue structures and the windlass effect are less able to effectively respond to the imposed loads (Hicks 1955; Mann and Inman 1964; Van Boerum and Sangeorzan 2003; Wang and Crompton 2004) leading to decreased rigidity and increased motion. Hypermobility of the foot is thought to result in a “splaying” of the loaded foot due to the combined effects of increased hindfoot eversion and forefoot abduction and dorsiflexion (Kitaoka, Lundberg et al. 1995; Van Boerum and Sangeorzan 2003). In contrast, a high arched foot is thought to be excessively rigid allowing considerably less motion than the normal foot (Arangio, Chen et al. 1998) and, therefore, is less able to absorb imposed loads.

In spite of the repeated emphasis in previous research on foot structure and its relationship to foot function, there remains a limited understanding of the effect of arch height, or other related measures of foot structure, on dynamic foot mobility during walking (Hunt, Fahey et al. 2000; Hunt and Smith 2004). While the association between arch height and foot mobility is of primary interest, there are several limitations to available literature examining this association. One of the greatest limitations to these studies is related to the populations studied. The failure of studies to recruit subjects with an adequate dispersion in arch height measures makes it difficult to establish associations between structure and function. One recent study was identified where an effort was made to include non-normal foot structures. Unfortunately, foot structure was implied based on clinical issues associated with a planus or pronated foot and no attempt was made to quantify foot structure (Hunt, 2004). The inclusion of individuals with pathology raises the question of whether the results are associated with pathology or foot structure.

The purpose of this study is to determine the extent to which arch height affects foot kinematics during gait using a multi-segment foot model. This question will be addressed though recruitment of subjects with a wide range of arch heights and no history of foot pathology. The use of a multi-segment foot model will enable foot structures affecting foot function during gait dynamics to be assessed. This data will help determine if arch height effects dynamic foot function to the extent and in the manner currently hypothesized based on current theory.

Methods

Subjects

Seventeen physically active subjects (8 male, 9 female) between the ages of 18 and 36 with no current history of lower extremity pain were recruited to participate in this study. Subjects were on average 25 ± 4.5 years old with a mean weight and height of 74± 14 Kg and 1.7 ± .1 m respectively. Subjects reported no history of knee injuries, stress fractures, plantar fasciitis or other foot pathology. Informed consent was obtained from all subjects prior to participation. To ensure that subjects with a wide range of arch heights participated subjects with self reported high or low arches were recruited.

Radiographs

Two weight bearing radiographs were taken of the tested foot in a manner consistent with common clinical practice (Saltzman, Nawoczenski et al. 1995). Antero-posterior and lateral radiographs of the foot (Saltzman, Brandser et al. 1994; Saltzman, Nawoczenski et al. 1995) were used to determine static foot alignment for each subject as well as allow anatomically based alignment of local coordinate systems for the foot. The angle between the dorsal surface of the first metatarsal (1MTA) and the inferior border of the calcaneus (CIA)(Saltzman, Nawoczenski et al. 1995) was chosen to represent arch height.

Procedure

A four segment rigid-body model including the foot and lower leg was used to assess dynamic foot function during gait for subjects meeting the studies inclusion and exclusion criteria. Three infrared light emitting diodes (IREDs) placed to minimize skin-bone movement (Tranberg and Karlsson 1998; Umberger, Nawoczenski et al. 1999; Houck, Yack et al. 2004) were tracked (Optotrak 3020, Northern Digital Inc.) to determine the three dimensional kinematics of the first metatarsal (1MT), lateral forefoot (FORE), calcaneus (CALC) and tibia (TIB) during the stance phase of gait. Tibial motion was determined in a manner consistent with Houck et al. (2004) while calcaneal motion was assessed using two markers on the lateral calcaneus proximal to the calcaneal fat pad and one on the posterior aspect of the calcaneus. Forefoot motion was determined by markers on the proximal and distal second metatarsal and the proximal fifth metatarsal. Motion of the first metatarsal was tracked with a lightweight marker triad mounted on the first metatarsal, medial to the extensor hallucis longis tendon in a manner similar to Leardini et al. (Leardini, Benedetti et al. 1999; Umberger, Nawoczenski et al. 1999; Cuddeford 2000). A digitizing process in conjunction with lateral and AP x-rays of the foot was used to identify the location of underlying bony geometry relative to each segment's marker triad. For the first metatarsal and forefoot segments the local coordinate system was adjusted to align with the long axis of the first and second metatarsals respectively. For the calcaneal segment the AP axis was aligned from the posterior heel to the midpoint of the midfoot and inclined in the sagittal plane to match the calcaneal inclination angle (Saltzman, Nawoczenski et al. 1995). The vertical axis was aligned parallel to the calcaneal bisector line in the frontal plane with the medo-lateral axis orthogonal to the other two.

Move 3D software (C-Motion Inc.) was used to determine segment and joint angles and velocities using an Euler angle approach. Kinematic data was collected at 120 Hz and low pass filtered using a 6Hz cutoff frequency. Ground reaction forces collected via a Kistler force plate (Kister Instrument Corp., Model 9865B) at a sampling rate of 360 Hz and synchronized with kinematic data were used for normalization of kinematic data to 100% percent stance phase. An overhead timing system was used to guide subjects to ambulate at a controlled speed of 0.78 statures/second.

Kinematics

Range of motion values were determined by calculating the difference between the maximum and minimum kinematic values for the three variables of interest during the interval from forefoot contact (FF) to peak arch elongation (PAE). In accordance with previous work (Chapter II) FF was determined using rotational velocity of the first metatarsal while PAE was determined by the point of peak 1MT dorsiflexion relative to the calcaneus. Mean range of motion values during the FF-PAE interval were used to represent the motion for each individual. Additional events during the stance phase were also determined for use in graphical representation. The timing of forward rotation of the first metatarsal (FR) as well as a minima in forward rotation velocity of the first metatarsal (MINVEL) are also presented. To allow side by side comparison of kinematic curves, the mean over the interval from forefoot contact to forward rotation of the first metatarsal was subtracted from the mean kinematic pattern for each subject (Hunt, Smith et al. 2001). As a result of this shift in the data the standard deviation value is frequently smallest at the midpoint of the interval. Peak dorsiflexion (1MT - CALC), eversion (CALC - TIB) and abduction (FORE - CALC) over the stance phase were determined prior to subtracting mean values. A correlation analysis was performed to determine the association between sagittal (1MT - CALC), frontal (CALC - TIB) and transverse plane kinematics (FORE-CALC) and arch height (1MTA-CIA). A similar examination was performed for peak kinematic values.

Results

Subjects demonstrated a wide range of radiographically determined foot structures with mean (± 1 SD) 1MTA-CIA angles of 129.8 (12.1) degrees on average and a range from 114-153 degrees. Five participants in the study demonstrated arch heights greater, and four had arch heights less than one standard deviation from the mean value observed in a clinical patient population (Saltzman, Nawoczenski et al. 1995). This diversity was also evident in the individual measures of 1MTA and CIA with mean (± 1 SD) angles of 26.5 (4.8) and 24.3 (8.1) degrees respectively. In spite of this diversity of foot structures (Figure 4-1A) kinematic pattern variability was low (Figure 4-1B). Mean standard deviation values (when averaged over the stance phase) for the ensemble average curves are less than or equal to 1.5 degrees for each of the primary kinematic measures (Figure 4-1B, Figure 4-2, Figure 4-3).

Consistent with previous reports (Chapter II) the first metatarsal was found to maintain a nearly stationary position for nearly 50 percent of the stance phase beginning forward rotation approximately 10 percent stance before peak arch elongation (Figure 4-4). Ensemble average curves (Figure 4-1B, Figure 4-2, Figure 4-3) demonstrate that the hindfoot everts, forefoot abducts and first metatarsal dorsiflexes after heel contact. At approximately the same time as the FF event most subjects transition to gradual inversion and the rate of forefoot abduction decreases. During the interval from FF to PAE (18.0 ± 3.1 to 76.2 ± 3.3 percent stance) subjects undergo gradual first metatarsal dorsiflexion, forefoot abduction and hindfoot inversion. Following peak arch elongation subjects undergo first metatarsal plantarflexion, forefoot adduction and continued inversion.

Poor associations were observed between arch height and sagittal, frontal and transverse plane kinematic range of motion during the FF-PAE interval (Table 4-1). Associations were, however, observed between peak hindfoot eversion, forefoot abduction and first metatarsal dorsiflexion excursion during the FF-PAE interval and 1MTA-CIA angle (Table 4-1).

Discussion

The height of the medial longitudinal arch has long been thought to have a strong association with foot mobility and function during gait. There is however little quantitative data to support current clinical reasoning and the numerous claims to this effect (Cashmere, Smith et al. 1999; Hunt, Fahey et al. 2000; Hunt and Smith 2004). The results of this study suggest that range of motion values during the stance phase of gait are not associated with radiographically determined arch height. The results obtained do, however, provide new insight into dynamic foot function during the stance phase of gait.

Subjects

The population tested demonstrates a large and well distributed range of radiographically determined arch heights. The mean and standard deviation values for radiographic measures confirm that the population tested represents a considerable diversity in foot structure. Mean values for first metatarsal inclination, calcaneal inclination angle and arch angle are all comparable with previous reports of foot structure (Steel, Johnson et al. 1980; Saltzman, Nawoczenski et al. 1995; Cavanagh, Morag et al. 1997; Wearing, Urry et al. 1998; Morag and Cavanagh 1999; Wearing, Urry et al. 1999). The diversity of the population tested is confirmed by comparison to these studies as the standard deviation values for each of the three variables is greater than previously reported. The population tested in this study includes extremes of arch height and though many of the subjects were highly active none reported experiencing pathology such as stress fractures, plantarfasciitis or knee problems.

Kinematics

The methodology used in the present study is novel with respect to coordinate system orientation as well as definition of the interval over which motion was determined, therefore, limiting the ability to directly compare the obtained results with available literature. The kinematic results obtained are, however, consistent with available literature with respect to both timing and total magnitude of motion. Though a wide range of foot structures were tested kinematic patterns were found to be highly consistent within individuals with a high degree of similarity between individuals when offsets due to static alignment differences are accounted for as previously suggested (Hunt, Smith et al. 2001).

Frontal plane motion of the calcaneus relative to the tibia is likely the most commonly reported measure of foot motion during gait. The results of this study are consistent with previous examination of frontal plane hindfoot motion (Figure 4-2). The mean motion values observed over the entire gait cycle were near the upper end of values found in the literature but fall well below what is physiologically available (Cornwall and McPoil 1999; Leardini, Benedetti et al. 1999; Rattanaprasert, Smith et al. 1999; Hunt, Fahey et al. 2000; Wu, Su et al. 2000; Wulker, Stukenborg et al. 2000; Hunt, Smith et al. 2001; Su, Wu et al. 2001; Westblad, Hashimoto et al. 2002; Nester, Findlow et al. 2003; Valderrabano, Hintermann et al. 2003; Hunt and Smith 2004). In spite of the large diversity of arch heights tested the pattern of motion was consistent with the current understanding of frontal plane hindfoot motion. While prolonged eversion into the middle of stance has been reported (Moseley, Smith et al. 1996; Cornwall and McPoil 1999) the results of this study are consistent with the many reports of peak eversion occurring between 20-30% of stance phase followed by gradual re-inversion (Kappel-Bargas, Woolf et al. 1998; Rattanaprasert, Smith et al. 1999; Reischl, Powers et al. 1999; Carson, Harrington et al. 2001; Hunt, Smith et al. 2001; Smith, Rattanaprasert et al. 2001; Powers, Chen et al. 2002; Nester, Findlow et al. 2003).

Transverse plane motion of the forefoot relative to the calcaneus observed in this study is similar in pattern and magnitude to that observed in previous studies (Figure 4-3). Consistent with other reports early abduction was followed by nearly static transverse plane orientation for much of stance followed by rapid adduction late in stance (Rattanaprasert, Smith et al. 1999; Wu, Su et al. 2000; Carson, Harrington et al. 2001; Hunt, Smith et al. 2001; Su, Wu et al. 2001; Hunt and Smith 2004).

Though arch motion and structure are both thought to play a critical role in dynamic foot function few studies have directly examined sagittal plane motion of the first metatarsal relative to the calcaneus (Wearing, Urry et al. 1998; Wearing, Smeathers et al. 2004). The results presented here (Figure 4-1B) are similar in magnitude and pattern to previously reported results using this approach to track sagittal plane arch motion (Chapter II). In spite of large differences in population demographics the difference in excursion is only approximately one standard deviation.

Previous efforts to determine the effect of arch height on foot function during gait have observed an association between static alignment and peak excursion (Hunt, Fahey et al. 2000). It should be noted, in light of the lack of association between arch height and range of motion, the magnitude of peak excursions are greatly influenced by offsets in coordinate system orientation that are made based on static position. As expected peak arch elongation was strongly associated with arch height or more simply put static position was strongly associated with dynamic position (Table 4-1). Likewise peak eversion is associated with arch height because eversion in the static condition is associated with arch height. The most interesting aspect of these findings is that the underlying offsets associated with arch height have little relation to the range through which individuals move during gait. It was observed that when these offsets were removed by normalizing to the position maintained from FF to FR that the patterns were consistent between individuals with markedly different static bony alignment (Figure 4-1, Figure 4-2, Figure 4-3).

Role of Arch Height

The results of this study when taken with available literature strongly indicate that in the non-pathologic foot arch height is not associated with the magnitude of intersegmental foot motion during the stance phase of gait (Hunt, Fahey et al. 2000; Hunt and Smith 2004). Clinical dogma has long suggested that individuals with very high or low arches are destined to experience foot and ankle pathology due to altered foot kinematics secondary to abnormal foot mobility. In spite of the pervasive opinion that mobility of the foot is inversely related to the height of the medial longitudinal arch (Kirby 1989; Arangio, Chen et al. 1998; Van Boerum and Sangeorzan 2003; Wang and Crompton 2004) few studies have examined the association between arch height and foot function under dynamic conditions. Efforts to quantify the effect of arch height on foot motion have failed to present evidence that decreased arch height causes articular geometry, supporting soft tissue structures and the windlass effect to be less able to effectively withstand the imposed loads leading to decreased rigidity and increased motion (Hunt, Fahey et al. 2000; Hunt and Smith 2004). The results of this study indicate that from forefoot contact to peak arch elongation, the interval when the foot is experiencing the greatest load, there was not a significant effect of arch height with respect to any of the three kinematic measures assessed (Table 4-1, Figure 4-1, Figure 4-2, Figure 4-3). It was also observed that when the entire stance phase was examined there was again no association between arch height and foot mobility (Table 4-1). These results would also seem to be contrary to arguments for a dynamic splaying of the foot, as demonstrated by increased hindfoot eversion, forefoot abduction and arch elongation (Kitaoka, Lundberg et al. 1995; Van Boerum and Sangeorzan 2003).

Conclusion

The purpose of this study was to determine the extent to which arch height is associated with foot kinematics during gait using a multi-segment foot model. The results of this study along with those of others (Hunt, Fahey et al. 2000; Hunt and Smith 2004) serve to refute the belief that decreasing arch height is associated with increased motion of the foot during gait. To the contrary it was observed that in a population of asymptomatic individuals kinematic patterns were highly consistent for the study population in spite of the extreme range of arch heights tested. Though statements to the effect that pes planus is a variant of normal are likely overstated (Van Boerum and Sangeorzan 2003) it is clear that arch height does not determine foot mobility.

Table 4-1. Mean (± 1 SD) range of motion (degrees) for kinematic patterns of interest as well as correlation coefficient values representing the association between radiographicially determine arch angle (degrees) and range and peak kinematic values. (CALC= Calcaneus, TIB= Leg, FORE= Forefoot, 1MT= First metatarsal, X= frontal plane motion, Y= transverse plane motion, Z= sagittal plane motion.)

	Range of motion from FF-PAE	Correlation ROM: 1MTA-CIA	Correlation Peak: 1MTA-CIA
CALC-TIB X	5.8 ± 2.2	-.01	-.78
FORE-CALC Y	3.4 ± 1.0	.25	-.58
1MT-CALC Z	7.4 ± 2.0	-.11	.95

Figure 4-1. Sagittal plane kinematics of the first metatarsal relative to the calcaneus during the stance phase of gait. A) Mean data for each subject and B) ensemble average (± 1 SD) sagittal plane kinematics for all subjects after subtracting the mean value from forefoot contact to first metatatarsal forward rotation from the mean data for each subject. (FF= Forefoot contact, FR= First metatarsal forward rotation, PAE= Peak arch elongation, MINVEL= Minima in first metatarsal forward rotation velocity).

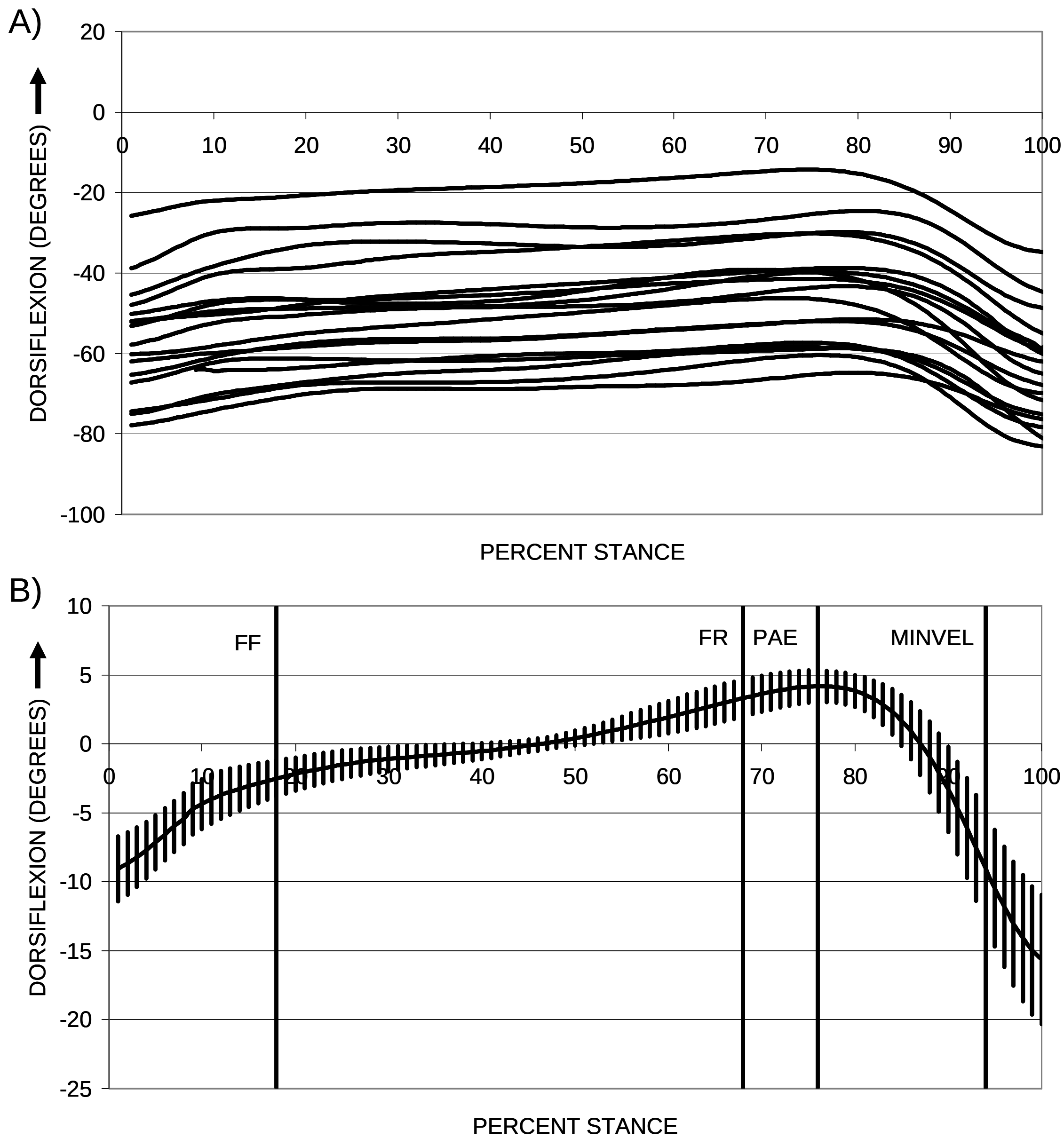


Figure 4-2. Ensemble average (± 1 SD) frontal plane kinematics of the calcaneus relative to the tibia during the stance phase of gait.

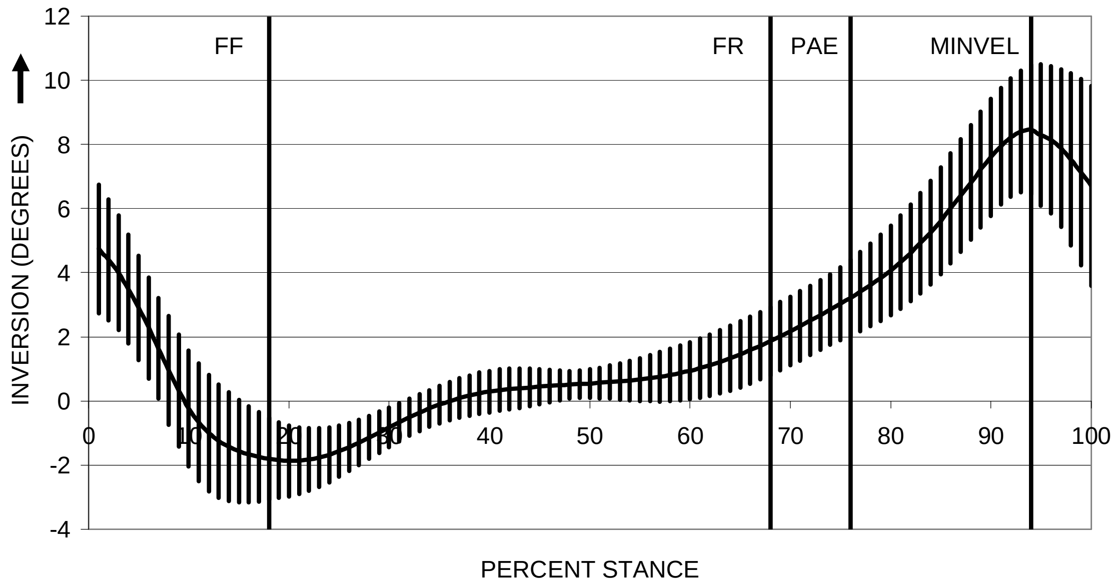


Figure 4-3. Ensemble average (± 1 SD) transverse plane kinematics of the forefoot relative to the calcaneus during the stance phase of gait.

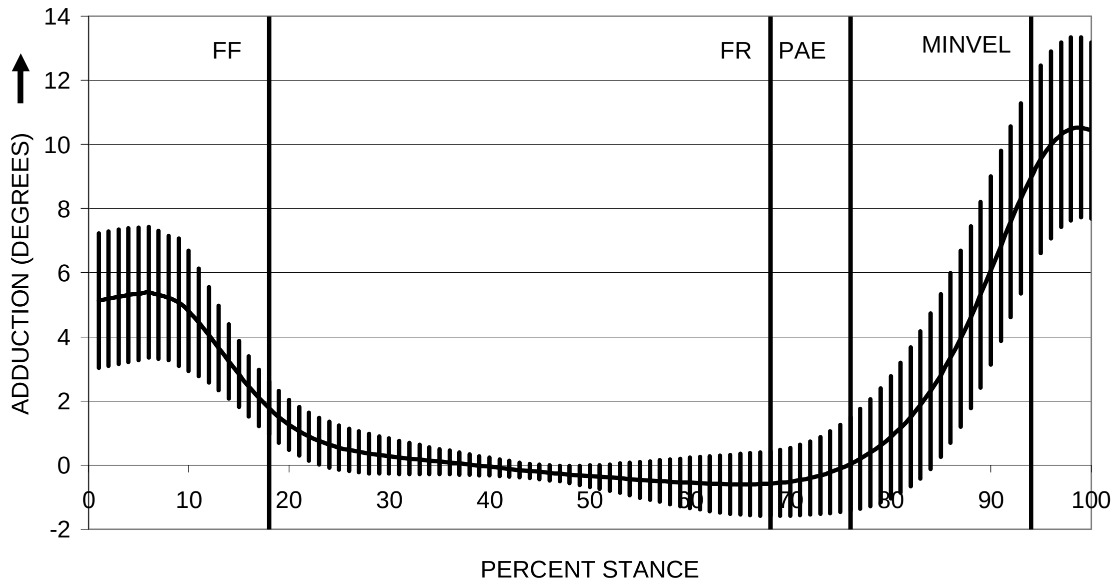


Figure 4-4. Ensemble average (± 1 SD) sagittal plane kinematics of the first metatarsal relative to the floor (global) during the stance phase of gait.

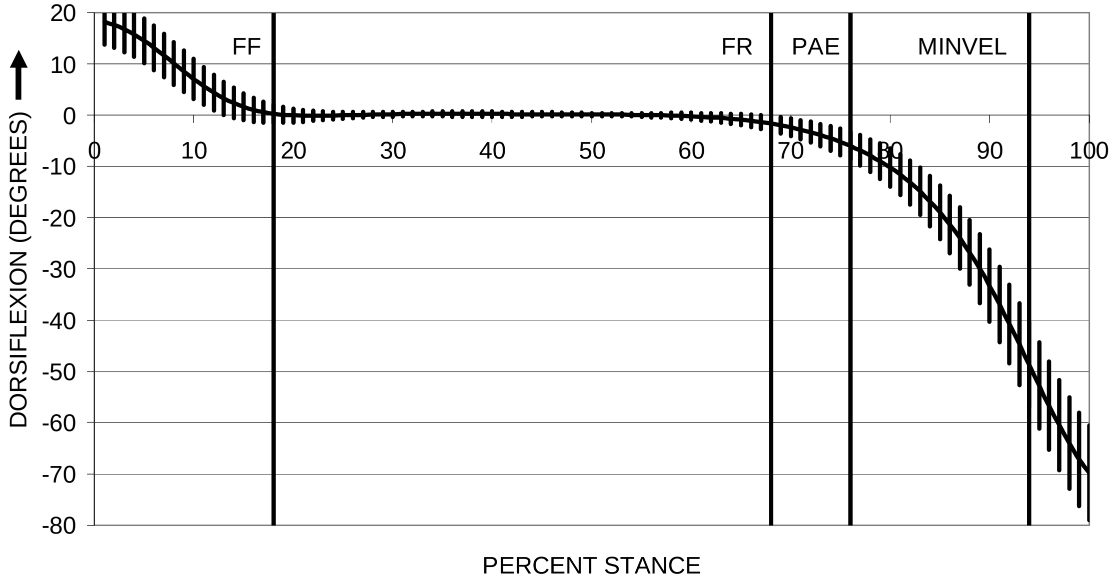
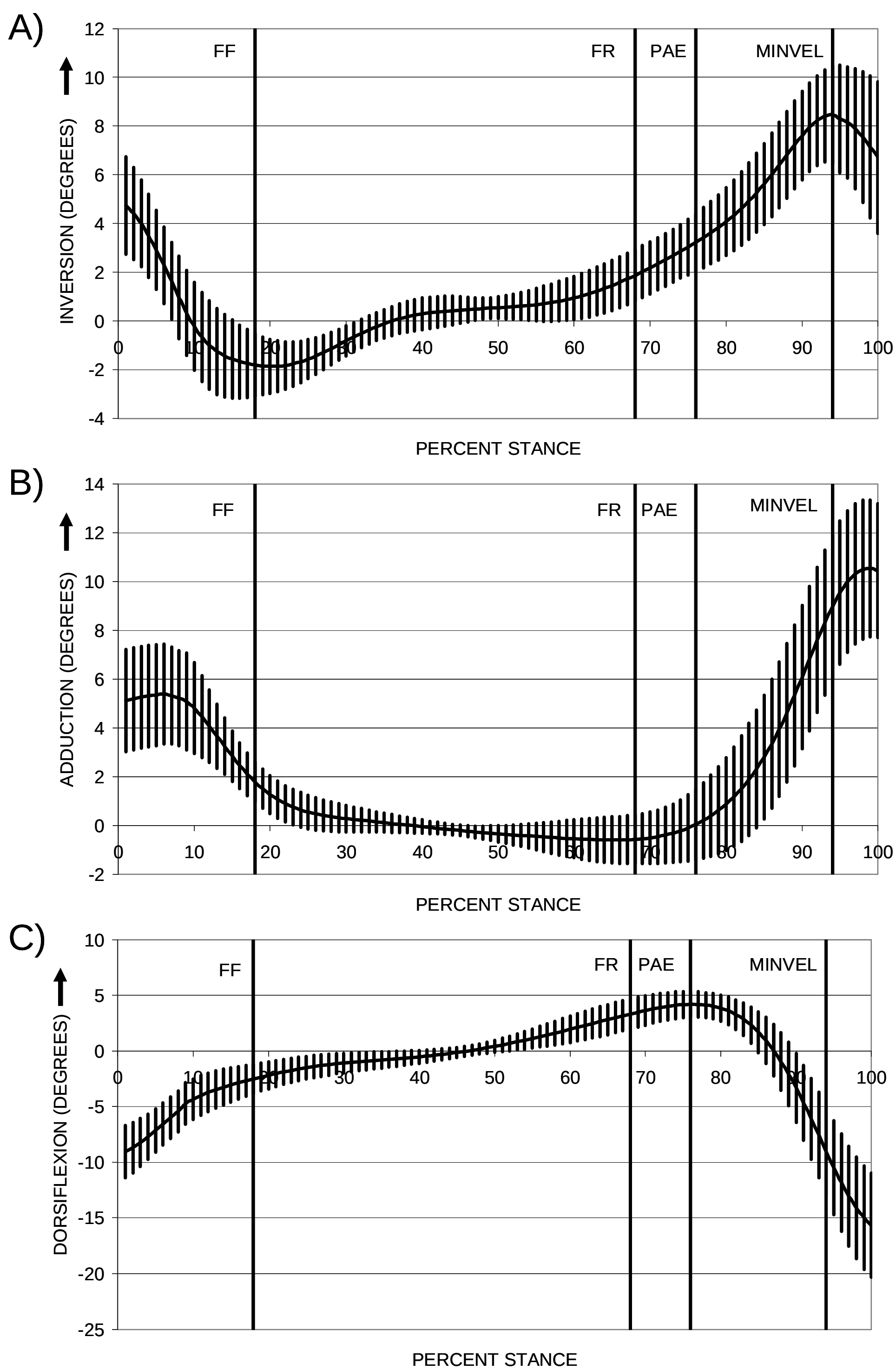


Figure 4-5. Ensemble average (± 1 SD) A) frontal plane kinematics of the calcaneus relative to the tibia, B) transverse plane kinematics of the forefoot relative to the calcaneus and C) sagittal plane kinematics of the first metatarsal relative to the calcaneus during the stance phase of gait.



CHAPTER V

THE WINDLASS MECHANISM, SUPINATION AND THE EFFECT OF ARCH HEIGHT

Introduction

Many of the current theories regarding foot function have their foundation in insights gained through clinical experience, early theoretical discussions of the foot function and in-vitro testing. In an effort to understand and communicate dynamic foot function many mechanical models have been generated to represent structures and mechanisms thought to control the complex motions and maintain a normal balance between mobility and stability in the foot during the stance phase of gait. The most common models attempt to communicate dynamic foot function using mechanical analogs such as the windlass (Hicks 1954; Sarrafian 1987; Fuller 2000), leaf spring (Van Boerum and Sangeorzan 2003), twisted plate, universal joint (Olerud and Rosendahl 1987), beam (Hicks 1955; Sarrafian 1987), truss and tie-rod (Hicks 1955; Sarrafian 1987) and arch (Hicks 1955; Sarrafian 1987) among others.

Because of its simplicity and hypothesized relevance to dynamic foot function the windlass mechanism, one of the oldest mechanical representations used in the foot, is a staple of foot mechanics instruction. In an elegant series of studies performed in the 1950's Hicks (1954; 1955) demonstrated the importance of the windlass mechanism to support of the static foot. The simple concept of the winding of the plantar fascia around the metatarsal heads during late stance to produce arch elevation and support is both easy to understand and communicate.

Recent kinematic data have, however, called into question some of the long held beliefs regarding dynamic foot function in the mid to late stance phase when the foot must function as a rigid lever for pushoff. Both the concept of foot supination as well as the common representation of sagittal plane arch motion using the truss and tie rod have been found inconsistent with observed kinematics of the foot (Hunt, Smith et al. 2001). These findings, while serving to challenge many of the common assumptions regarding foot function may help to clarify the role of the windlass mechanism and the changes in foot orientation it produces as the foot forms a rigid lever for push-off.

Insight into the action of the windlass mechanism can be inferred by comparing kinematics of the arch and first metatarsal with that of previous evaluations of foot function (Hunt, Smith et al. 2001). As suggested by Hunt et al. (2001) the foot does not appear to form a supinated rigid lever during mid stance because arch elongation occurs until late into the stance phase (Hunt, Smith et al. 2001; MacWilliams, Cowley et al. 2003). It should, however, be noted that forward rotation of the first metatarsal (Chapter II) appears to coincide with the motions of forefoot plantarflexion and adduction along with hindfoot inversion. Unfortunately there has not yet been a direct comparison of the timing and direction of motions associated with action of the windlass mechanism and the formation of a lever for pushoff. Such a comparison would provide valuable new insight into the timing of motions that are believed to contribute to and result from activation of the windlass mechanism.

The height of the medial longitudinal arch is thought to influence arch function in part due to a decreased ability of the windlass mechanism to produce a stable foot configuration for pushoff. By altering the function of the windlass mechanism through altered articular geometry and supporting soft tissue structures the foot is thought to be less able to effectively withstand the imposed loads (Hicks 1955; Mann and Inman 1964; Van Boerum and Sangeorzan 2003; Wang and Crompton 2004) leading to decreased rigidity and increased motion. The forefoot abduction, dorsiflexion and hindfoot eversion that are commonly seen with decreased arch height are thought to alter the effect of plantarfascia tension. It has been repeatedly demonstrated that the coupling of frontal and transverse plane motion between the hindfoot and tibia is affected by arch height during running (Nigg, Cole et al. 1993; Nawoczenski, Saltzman et al. 1998; DeLeo, Dierks et al. 2004). It is, however, unclear how arch height influences intersegmental foot motion while it is acting as a lever for pushoff during gait. In particular the lowering of medial longitudinal arch potentially alters the frontal and sagittal plane inclination of the transverse tarsal joint. If true, in low arch subjects tension in the plantar fascia would result in relatively greater proportion of sagittal plane arch motion with a relative decrease in transverse plane motion. It is also believed that due to the everted position of the calcaneus and associated decrease in inclination of the STJ axis of rotation that the hindfoot inverts less relative to the tibia (Kirby 1989; Arangio, Chen et al. 1998). The net effect would be increased sagittal plane motion (Franco 1987; Van Boerum and Sangeorzan 2003) with less forefoot adduction (Van Boerum and Sangeorzan 2003; Hunt and Smith 2004) and inversion such that the windlass mechanism is less able to effectively produce a rigid lever for pushoff (Fuller 2000; Kirby 2000; Van Boerum and Sangeorzan 2003). The opposite would be true in the cavus foot in which the inclination of the transverse tarsal joint would theoretically be higher resulting in relatively more forefoot adduction as compared to forefoot plantarflexion resulting in a different coupling ratio. Though current theory suggest substantial changes in the magnitude and coupling of foot motion associated with arch height such claims have not been validated using in-vivo kinematic data (Hunt, Fahey et al. 2000; Hunt and Smith 2004).

In order to gain a better understanding of the timing of activation of the windlass mechanism as well as the effect of arch height on the ability to form a rigid lever for propulsion it will therefore be necessary to assess foot dynamics using a multi-segment foot model. The purpose of this study is three fold. The first is to examine the extent to which timing of forefoot adduction and plantarflexion coincide with activation of windlass mechanism via hallux dorsiflexion. The second is to quantify the coupling of intersegmental foot motion after activation of the windlass mechanism as the foot becomes a rigid lever for pushoff. The third is to determine the extent to which foot structure, as represented by arch height, influences the ability of the windlass mechanism to produce a rigid lever for pushoff.

Methods

Subjects

Seventeen physically active subjects (8 male, 9 female) between the ages of 18 and 36 with no current history of lower extremity pain were recruited to participate in this study. Subjects were on average 25 ± 4.5 years old with a mean weight and height of 74 ± 14 Kg and 1.7 ± .1 m respectively. Subjects reported no history of lower extremity injury including stress fractures, plantar fasciitis or knee injuries. Informed consent was obtained from all subjects prior to participation. To ensure that subjects with a wide range of arch heights participated only subjects with self reported high or low arches were recruited.

Radiographs

Two weight bearing radiographs were taken of the foot tested in a manner consistent with common clinical practice (Saltzman, Nawoczenski et al. 1995). Antero-posterior and lateral radiographs of the foot (Saltzman, Brandser et al. 1994; Saltzman, Nawoczenski et al. 1995) were used to determine the static foot alignment for each subject as well as allow anatomically based alignment of local coordinate systems for the foot. The angle between the dorsal surface of the first metatarsal (1MTA) and the inferior border of the calcaneus (CIA) (Saltzman, Nawoczenski et al. 1995) was chosen to represent arch height.

Procedure

A five segment rigid-body model including the foot and lower leg was used to assess dynamic foot function during gait for subjects meeting the studies inclusion and exclusion criteria. Three infrared light emitting diodes (IREDs) placed to minimize skin-bone movement (Tranberg and Karlsson 1998; Umberger, Nawoczenski et al. 1999; Houck, Yack et al. 2004; Longworth, Chockalingam et al. 2005) were tracked (Optotrak 3020, Northern Digital Inc.) to determine the three dimensional kinematics of the hallux (HX), first metatarsal (1MT), lateral forefoot (FORE), calcaneus (CALC) and tibia (TIB) during the stance phase of gait. Tibial motion was determined in a manner consistent with Houck et al. (2004) while calcaneal motion was assessed using two markers on the lateral calcaneus proximal to the calcaneal fat pad and one on the posterior aspect of the calcaneus. Forefoot motion was determined by markers on the proximal and distal second metatarsal and the proximal fifth metatarsal. Motion of the first metatarsal was tracked with a lightweight marker triad mounted on the first metatarsal, medial to the extensor hallucis longis tendon in a manner similar to Leardini et al. (Leardini, Benedetti et al. 1999; Umberger, Nawoczenski et al. 1999; Cuddeford 2000). Hallux motion was determined using a light weight marker triad with a rigid dorsal cuff and elastic band to secure it to the proximal phalanx (Longworth, Chockalingam et al. 2005). A digitizing process in conjunction with lateral and AP x-rays of the foot was used to identify the location of underlying bony geometry relative to each segment's marker triad. For the first metatarsal and forefoot segments the local coordinate system was adjusted to align with the long axis of the first and second metatarsals respectively. Adjustments if necessary were made in the sagittal and transverse planes to provide an anatomical reference frame for determining segment motion relative to the floor and calcaneus. For the calcaneal segment the AP axis was aligned from the posterior heel to the midpoint of the midfoot and inclined in the sagittal plane to match the calcaneal inclination angle (Saltzman, Nawoczenski et al. 1995). The vertical axis was aligned parallel to the calcaneal bisector line in the frontal plane with the medio-lateral axis orthogonal to the other two.

Move 3D software (C-Motion Inc.) was used to determine segment and joint angles and velocities using an Euler angle approach. Kinematic data was collected at 120

Hz and low pass filtered using a 6Hz cutoff frequency. Vertical ground reaction forces collected via a Kistler force plate (Kister Instrument Corp., Model 9865B) at a sampling rate of 360 Hz and synchronized with kinematic data were used to determine vertical ground reaction forces for normalization of kinematic data to 100% percent stance phase. An overhead timing system was used to guide subjects to ambulate at a controlled speed of 0.78 statures/second.

Timing

The angular velocity of first metatarsal sagittal plane rotation relative to the floor and sagittal plane motion of the first metatarsal relative to the calcaneus were used to determine the interval over which the coupling of motions was determined. Sagittal plane rotational velocity of the first metatarsal relative to the floor was used to determine foot flat (FF) and first metatarsal forward rotation (FR) events (Chapter II). Peak arch elongation (PAE) was defined as the maximally dorsiflexed position of the first metatarsal relative to the calcaneus during the stance phase of gait. The MINVEL point represents a minimum in forward rotation velocity of the first metatarsal relative to the floor. This point was chosen as it is highly repeatable within an individual and corresponds to the time at which the metatarsal head is lifted from the walking surface. By doing so the interval during which joint coupling was determined excludes the time when the toe is solely in contact with the floor.

Data Analysis

Mean and standard deviation values were calculated for the variables of interest. All kinematic data was normalized to 100% stance phase (Heel strike to toe off) for qualitative inspection and examination of trial to trial variability while mean data representing each subject was calculated to examine intersubject variability.

To determine coupling ratios between the pairs of variables to be examined kinematics were plotted relative to each other from PAE to MINVEL using an angle-

angle plot (Nigg, Cole et al. 1993; Stacoff, Nigg et al. 2000). This was performed for each trial and the slope of the linear regression line fit to the data for each individual trial was used to represent a coupling ratio for that trial. The median value was then used to represent the coupling ratio for each subject. A correlational analysis was then performed to determine the relationship between coupling ratio and arch height.

A total of four coupling ratios were calculated to represent foot function in terminal stance. Angle angle plots were generated comparing transverse and sagittal plane motion between the forefoot and calc (FORE-CALCY – FORE-CALCZ) as well as between the first metatarsal and calcaneus (1MT-CALCY – 1MT-CALCZ). The two other measures were transverse plane motion of the forefoot relative to frontal plane motion of the calc (FORE CALCY - CALC-TIBX) and transverse and frontal plane motion of the hindfoot relative to the calcaneus (CALC-TIB Y – CALC-TIB X). To ensure that the coupling ratio determined using linear regression was representative of the data a threshold for acceptance of the coupling ratio was set at an r-squared value of 0.7. This step was taken to guard against inappropriate representation of curvilinear data using a linear coupling ratio. In addition to using linear regression to determine coupling ratios absolute change over the interval was used to determine joint coupling. The total difference between max and min values from PAE to MINVEL was used, consistent with the work of others (Nawoczenski, Saltzman et al. 1998) to compare the similarity of the two methods.

Results

The population tested in this study represents a wide diversity of foot structures as demonstrated by radiographic measures characterizing foot structure. Mean (± 1 SD) first metatarsal inclination was 26.5 ± 4.8 degrees with a range from 16.0 to 34.6 degrees while calcaneal inclination angle had a range from 10.8 to 35.6 about a mean (± 1 SD) of 24.3 ± 8.1 degrees. This resulted in mean (± 1 SD) 1MTA-CIA angles of 129.8 ±12.1

degrees with a range from 114-153. Five participants in the study demonstrated arch heights greater, and four had arch heights less than one standard deviation from the mean value observed in a clinical patient population (Saltzman, Nawoczenski et al. 1995). In spite of the large deviations from normal foot structure and high levels of activity no subjects reported experiencing commonly reported pathology such as stress fractures, plantarfasciitis or knee problems.

Time normalized kinematic patterns relevant to the activation of the windlass mechanism and the timing of supination are presented (Figure 5-1, Figure 5-2, Figure 5-3, Figure 5-4, Figure 5-5, Figure 5-6, Figure 5-7, Figure 5-8) with previously defined first metatarsal and arch kinematic events (FF, FR, PAE, MINVEL) overlaid for comparison between figures. The mean and standard deviation for the timing of these events are presented in Table 5-1. Intersubject variability at each percent stance interval can be observed for ensemble average data in Figures 5-1 through 5-7. Due to diversity in static foot structure standard deviation values were considerably greater for ensemble average curves prior to correction for offsets (Figure 5-2).

Kinematic coupling over the PAE-MINVEL interval and its relationship to arch height can be found in Table 5-2. No association greater than .5 was observed between arch height and coupling ratio with the exception of CTX-CTY coupling. Due to non-linear coupling relationships for several subjects this ratio was only calculated for ten of the 17 subjects (Figure 5-9). The association between methods of calculating coupling ratios was high as seen in Table 5-2. Coupling ratios were calculated such that coupling ratios greater than one indicate that, on average, transverse plane motion was greater than motion in other planes.

Discussion

Recent studies examining foot kinematics during the stance phase of gait (Hunt, Fahey et al. 2000; Hunt and Smith 2004) and theoretical discussions of foot function

(Sarrafian 1987) have called into question the effect of foot structure on mobility during gait as well as the concept of mid-stance supination as a mechanism for forming a rigid lever for pushoff. The results of this study demonstrate that supination due to activation of the windlass mechanism does in fact occur but takes place later in the stance phase than would be suggested using measures such as heel rise. It was also observed that though the coupling of foot motion after initiation of the windlass mechanism is consistent with supination arch height was not found to influence these coupling ratios.

Supination of the foot, as defined by plantarflexion and adduction of the forefoot combined with inversion of the hindfoot, has long been considered necessary for the foot to function as a rigid lever for pushoff. This combination of motions, which were found to occur shortly after hallux dorsiflexion (Figure 5-8), are thought to result in a segmental and articular configuration that allows sufficient foot rigidity to withstand the large loads experienced in the latter half of stance. The windlass mechanism contributes to the integrity of the supinated foot by producing tension in the plantar fascia. The resulting approximation of the first metatarsal head and inferior calcaneus, results in motion about the anterior and medially inclined oblique mid-tarsal, and talo-calcaneo-navicular axes of rotation (Hicks 1953). In addition to providing support to the arch the supinated position produced by tension in the plantar fascia produces what is thought to be a more stable foot position during pushoff.

The segmentation of stance phase using arch and first metatarsal kinematics provides a clear framework to examine the timing of events associated with supination of the foot via the windlass mechanism. Consistent with previous results (Chapter II) the first metatarsal was observed to maintain a nearly static sagittal plane angle relative to the floor for nearly 50% of the stance phase (Table 5-1, Figure 5-1). During this interval all subjects were found to demonstrate progressive forefoot dorsiflexion and gradual hindfoot inversion while the forefoot maintained an abducted position (Figure 5-2, Figure 5-5, Figure 5-6, Figure 5-8). These patterns of motion are consistent with a similar examination of foot kinematics (Hunt, Smith et al. 2001). While the forefoot continues to dorsiflex first metatarsal rotation (FR) leads to gradual dorsiflexion of the hallux and adduction of the forefoot relative to the hindfoot. There is a brief interval between FR and PAE during which loading of the plantar fascia causes arch kinematics to gradually change prior to forefoot plantarflexion. Though the windlass mechanism is activated during this time, due to hallux dorsiflexion (Figure 5-4), tension in the plantar fascia appears to be insufficient to overcome arch loading. It is, however, evident that tension in the plantar fascia eventually limits forefoot and first metatarsal dorsiflexion producing supination of the foot (Figure 5-2, Figure 5-4, Figure 5-5, Figure 5-8). Finally, at the time when the first metatarsal begins to raise from the walking surface (MINVEL) a transition in foot motion can be observed during which hallux rotation relative to the floor increases (Figure 5-3), the rate of first MTP joint dorsiflexion and forefoot abduction decrease and hindfoot motion transitions from inversion to eversion ending the supinatory motion (Figure 5-4, Figure 5-5, Figure 5-6, Figure 5-8).

Consistent with arch kinematics found in this study (Figure 5-2), progressive loading of the plantar fascia until late in the stance phase has been both quantified in-vitro and can be inferred from in-vivo testing. In recent cadaveric testing, loading of the plantar fascia during gait was simulated (Erdemir, Hamel et al. 2004). Though many assumptions were made to make dynamic testing possible, the results of the simulation produced a plantar aponeurosis force trace that is remarkably similar to arch kinematic patterns observed in this study. Recent work by Warren et al. (2004) examining plantar pressures in shod walking at multiple speeds demonstrates a gradual increase in loading under the hallux until late in stance phase. Even when accounting for the ability of muscular contraction to add to torque about the MTP joint due to the windlass mechanism, the results of both studies are consistent. The gradual increasing plantar fascia tension during mid-stance is what would be expected given the kinematics in the current study. Given the limited extensibility of the plantar fascia (Kitaoka, Luo et al.

1994; Erdemir, Hamel et al. 2004) and progressive forefoot dorsiflexion, changes in bony geometry and tension in the plantar fascia appear to play an important role in allowing gradual forefoot dorsiflexion prior to activation of the windlass mechanism at the end of stance phase.

To our knowledge this is the first study to determine the extent to which inter-segmental foot motion is coupled at the end of stance phase as the foot supinates to assume a more stable configuration for pushoff. Though stability in the end of stance is believed to be a key component of normal foot function the relative contribution of intersegmental motion to the production of supination has not been quantified. The coupling ratios provide insight into the motions that result from action of the windlass mechanism at the end of the stance phase (PAE until MINVEL).

All coupling relationships demonstrated a very high level of linear coupling as is evidenced by mean r-squared values greater than .90. Given these high r-squared values it is of no surprise that that calculation of coupling ratios using linear regression or absolute change over the interval of interest (PAE-MINVEL) produced nearly identical coupling ratios (Table 5-2). While the variability in coupling ratios, as demonstrated by the coupling ratio standard deviation values, for CALC-TIBX – FORE-CALCY coupling was equal to the lowest observed in the work by Nawoczenski et al. (1998) the values for 1MT-CALCZ – 1MT-CALCY and FORE-CALCZ – FORE-CALCY coupling were nearly a third of that value. This level of variability is lower than expected given the diversity of the population tested.

Consistent with the concept of supination the foot moved toward adduction and plantarflexion of the forefoot along with inversion of the hindfoot during the interval from PAE to MINVEL. Of the coupling relationships examined FORE-CALCY – CALC-TIBX was the only one to take into consideration two separate pairs of motion segments. This comparison allows the opportunity to examine the contributions of motion of the forefoot as well as hindfoot. During this interval forefoot adduction was greater than hindfoot inversion resulting in a coupling ratio of 1.65. All but two of the subjects demonstrating a coupling ratio greater than 1 indicating that transverse plane motion predominated. Adduction which is the direct result of approximation of the inferior calcaneus and first metatarsal head via the windlass mechanism was considerably greater than hindfoot inversion suggesting a functional axis of rotation biased toward adduction. Unlike FORE-CALCY – CALC-TIBX coupling the coupling between hindfoot inversion and adduction (CALC-TIBY – CALC-TIBX) produced inconsistent and somewhat surprising results (Figure 5-9). While the hindfoot was inverting (Figure 5-6), the pattern of adduction was less predictable (Figure 5-7) resulting in the inability to determine coupling ratios for seven individuals. In the subjects where a coupling ratio was calculated a moderate association between arch height and coupling ratio was observed (Table 5-2). These results should, however, be interpreted with caution due to the inability to calculate coupling ratios for all subjects and coupling relationships that were considerably less linear than observed with other variables. It was anticipated that a decreased coupling ratio would be observed in lower arched individuals due to a greater contribution of frontal plane motion. Such an effect is thought to result from lowering of the sagittal plane inclination of the subtalar joint axis as arch height decreases (Nawoczenski, Saltzman et al. 1998). The results of this study are surprising in that negative coupling ratios were observed in two subjects with very low arches. Consistent with kinematics reported on a subject with posterior tibialis tendon dysfunction (Rattanaprasert, Smith et al. 1999) abduction was found to occur after PAE rather than the commonly reported adduction that was seen in other subjects (Figure 5-5). It is unclear why for many subjects highly consistent and linear relationships were observed while for others non-linear patterns were observed preventing the calculation of coupling ratios during the interval of interest.

Motion of the first metatarsal relative to the calcaneus, while demonstrating characteristic adduction and plantarflexion produced by the windlass mechanism, was found to exhibit a coupling ratio considerably lower than that found using the forefoot (Table 5-2). A coupling ratio near 0.5 indicates that nearly twice as much plantarflexion of the first metatarsal is occurring when compared to adduction during this interval. Though it may at first be surprising that the coupling ratios differ considerably between segments this discrepancy can be accounted for by the modeling approach and its relation to the axes of rotation of the foot. The forefoot was represented such that two markers on the second metatarsal played a primary role in determining transverse and sagittal plane motion of the forefoot segment. Motion of this segment relative to the calcaneus occurs primarily about the oblique transverse tarsal joint axis. In contrast first metatarsal motion relative to the calcaneus is occurring about both the oblique transverse tarsal and TCN joint axes as well as about the independent axis of the first ray (Hicks 1953). As the foot and first metatarsal rotate forward (FR) the forefoot is plantarflexing and the foot rotates about metatarsal heads 2-5 (metatarsal break). It has been suggested that to maintain contact with the floor the first metatarsal head must rotate about its own independent axis producing relative pronation which results in abduction and plantarflexion. By decreasing relative adduction and increasing plantarflexion this motion would act to decrease the coupling ratio to a value less than that of the forefoot. It should be noted that the shift to a lower coupling ratio when using the first metatarsal was consistent and observed in all subjects.

In addition to questions raised about the role of supination and the windlass mechanism in producing a rigid lever for propulsion recent kinematic studies have called into question the long held importance of arch height on dynamic foot function (Hunt, Fahey et al. 2000; Hunt and Smith 2004). Because intersegmental foot motion had not been examined in a non-pathologic population with a diversity of arch heights questions regarding the role of arch height persisted. To determine the role of arch height on coupling of foot motion after the activation of the windlass mechanism an effort was made to include subjects with a wide range of arch heights. This was done to ensure that if a relationship did exist it would not be masked by a limited range in the independent variable of arch height.

In spite of efforts to maximize the potential for identifying an effect of arch height no relationship between arch height and joint coupling was observed. In fact, remarkable consistency in kinematic patterns was observed across participants (Figures 5-1 to 5-7). After accounting for intersubject differences (Hunt, Smith et al. 2001) during the FF to FR interval (Figure 5-2) subjects were found to demonstrate consistent patterns and timing of foot motion in spite of the large range of foot structures observed (Table 5-1, Figure 5-2).

The results of this study provide no evidence that radiographically determined arch height in non-pathologic feet is associated with coupling that occurs following initiation of the windlass mechanism (Table 5-2). It was anticipated that changes in foot structure associated with arch height would result in altered coupling ratios as tension in the plantar fascia produced motion about the subtalar, transverse tarsal and TCN joint axes of rotation. This was however clearly not the case with little or no association between arch height and coupling values (Table 5-2).

The results of the current study when combined with previous investigations indicate that foot structure in individuals without pathology, as represented by arch angle, has little association with dynamic foot function. Therefore, clinical and theoretical discussion of the influence of arch height on the ability to form a rigid lever for pushoff is not substantiated. As has been previously suggested factors such as bias resulting in selective interpretation of clinical findings might play an important role in the current theory regarding causative factors in the occurrence of foot and ankle pathology. While the influence of arch height in healthy active individuals does not manifest itself during gait it is not to say that under higher demand conditions that arch height is not relevant.

Conclusion

The results of this study provide new and unique insight into the timing of events that lead to supination of the foot via activation of the windlass mechanism. Segmentation of the stance phase using first metatarsal and arch kinematics provides a useful reference frame that directly relates to dynamic foot function. It is not until late in stance that hallux dorsiflexion and windlass mechanism activation occur allowing supination of the foot during pushoff. The results of this study demonstrate that kinematic patterns and coupling ratios are similar within a non-pathologic population in spite of a diversity of foot structures. The overall similarity in kinematic patterns and lack of association between arch height and coupling runs contrary to arguments for a systematic relationship between foot structure and function. Though subjects had very different foot structures activation of the windlass mechanism via first metatarsal forward rotation resulted in a consistent pattern of intersegmental foot motion.

Summary

1. Changes in the kinematic pattern consistent with function of the windlass mechanism begin to occur late in the stance phase when hallux dorsiflexion, first metatarsal forward rotation and forefoot adduction occur along with hindfoot inversion.
2. The foot does not demonstrate “supination” until approximately 76 percent of the stance phase at which time first metatarsal and forefoot plantarflexion is combined with hallux dorsiflexion, first metatarsal forward rotation, forefoot adduction and hindfoot inversion.
3. While motions consistent with the foot becoming a rigid lever for pushoff were observed they occurred much later in the stance phase than has been previously acknowledged.

4. Joint coupling after the beginning of forefoot plantarflexion was found to be highly linear with low variability observed for coupling of the forefoot and first metatarsal relative to the calcaneus.

5. No association was observed between arch height and the coupling of motion after initiation of plantarflexion of the first metatarsal and forefoot relative to the calcaneus.

Table 5-1. Mean and standard deviation values (percent stance phase) for the timing of first metatarsal and arch kinematic events. (FF= Forefoot contact, FR= Forward rotation of first metatarsal, PAE= Peak Arch Elongation, MINVEL= minima in first metatarsal forward rotation velocity.)

Event	Mean	SD between subjects	SD within subject
FF	18.0	3.1	1.0
FR	67.7	3.2	1.7
PAE	76.2	3.3	1.3
MINVEL	94.1	1.4	0.7

Table 5-2. Mean (± 1SD) values for both methods of calculating coupling ratios during the interval from PAE to MINVEL, the correlation between the two methods as well as the correlation between joint coupling as determined using linear regression and arch angle. Coupling ratio values greater than one indicate that transverse plane motion predominates. (CALC= Calcaneus, TIB= Leg, FORE= Forefoot, 1MT= First metatarsal; X= frontal plane rotation, Y= transverse plane motion, Z= sagittal plane motion.)

Comparison	Linear Regression	Change over interval	Correlation between methods	Correlation with arch angle
FORE-CALCZ-FORE-CALCY	0.96 ± 0.26	0.96 ± 0.27	.99	.34
1MT-CALCZ-1MT-CALCY	0.44 ± 0.24	0.43 ± 0.21	.99	-.12
CALC-TIBX-FORE-CALCY	1.62 ± 0.64	1.84 ± 0.80	.98	.17
CALC-TIBX-CALC-TIBY*	0.28 ± 0.44	0.56 ± 0.35	.71	-.62

*n=10

Figure 5-1. Ensemble average (± 1 SD) sagittal plane kinematics of the first metatarsal relative to the floor (global) during the stance phase of gait.

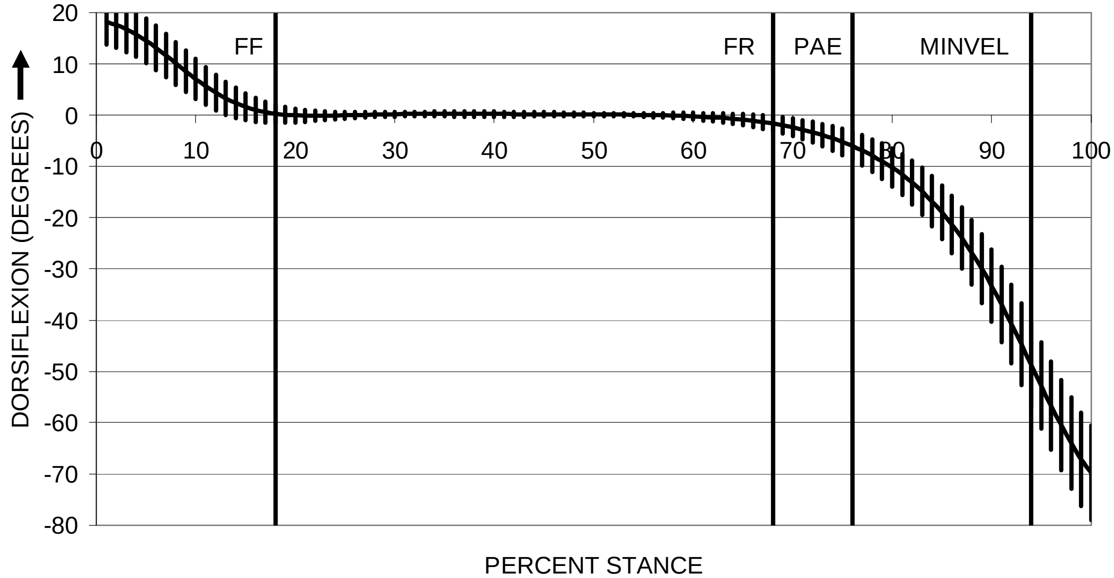


Figure 5-2. Sagittal plane kinematics of the first metatarsal relative to the calcaneus during the stance phase of gait. Mean data for each subject (A) and ensemble average (± 1 SD) sagittal plane kinematics for all subjects (B) after subtracting the mean value from forefoot contact to first metatatarsal forward rotation from the mean data for each subject. (FF= Forefoot contact, FR= First metatarsal forward rotation, PAE= Peak arch elongation, MINVEL= Minima in first metatarsal forward rotation velocity).

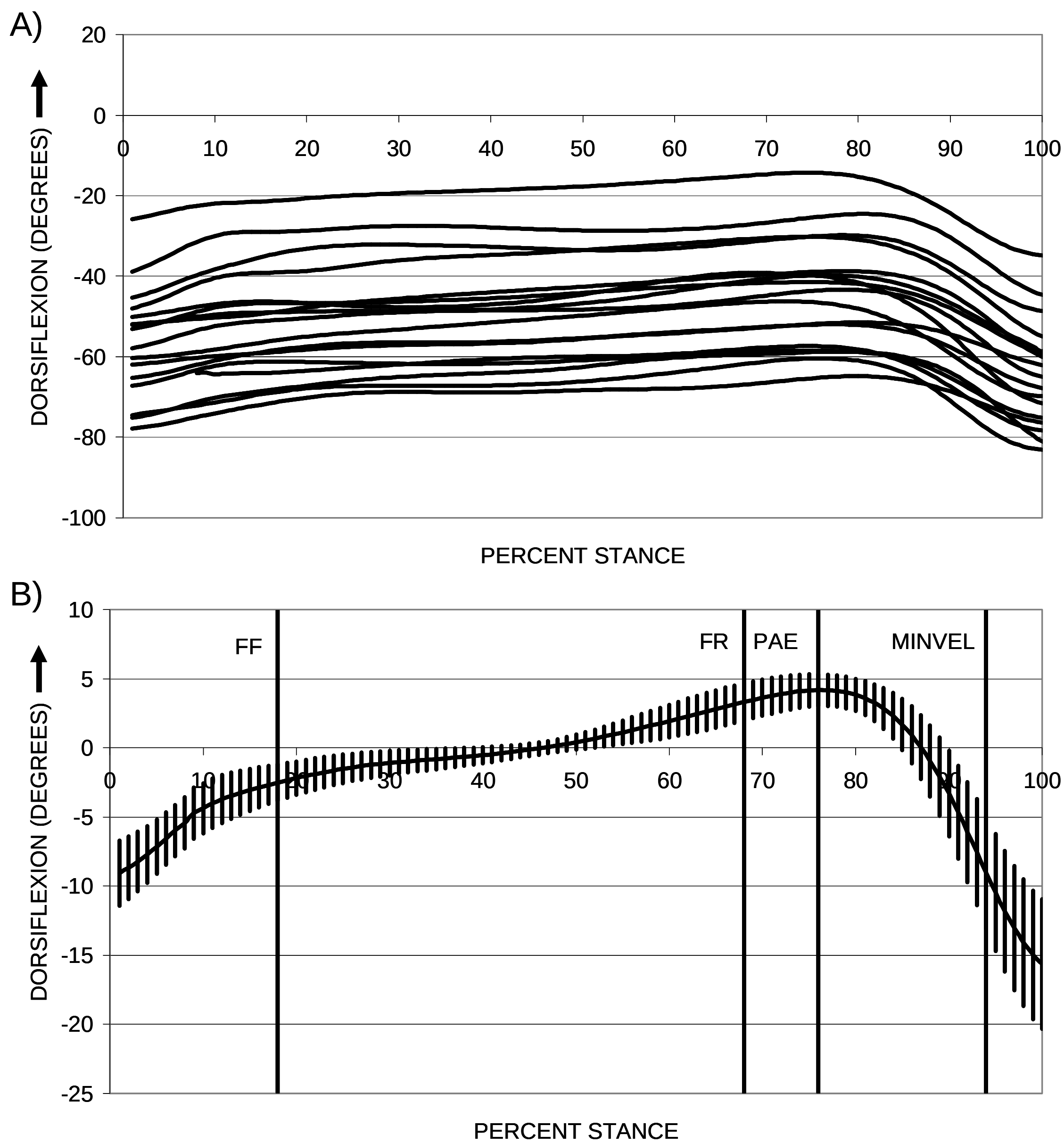


Figure 5-3. Ensemble average (± 1 SD) sagittal plane kinematics of the hallux relative to the floor during the stance phase of gait.

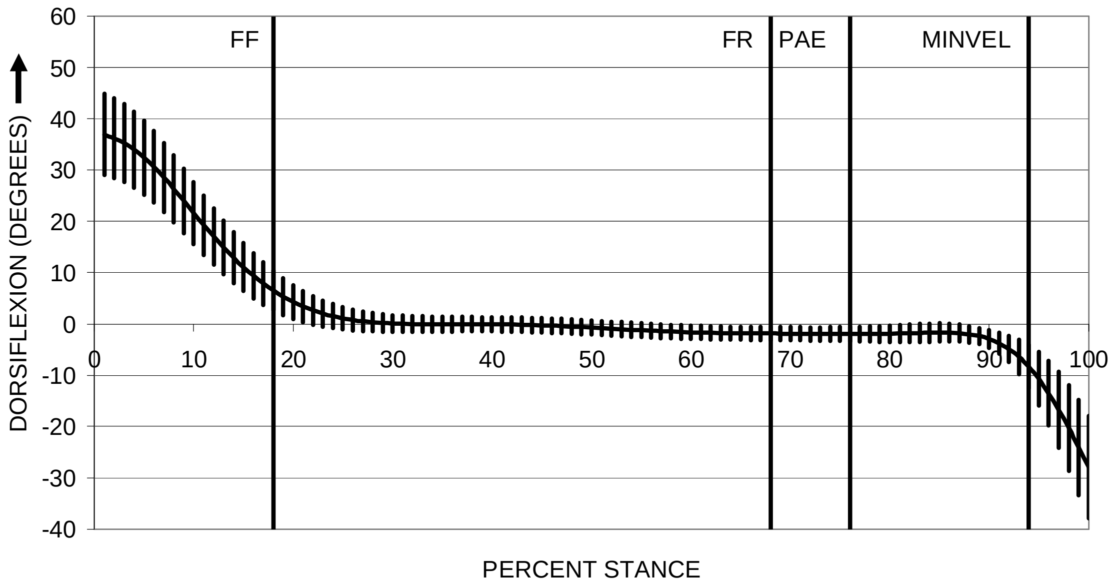


Figure 5-4. Ensemble average (± 1 SD) sagittal plane kinematics of the hallux relative to the first metatarsal during the stance phase of gait.

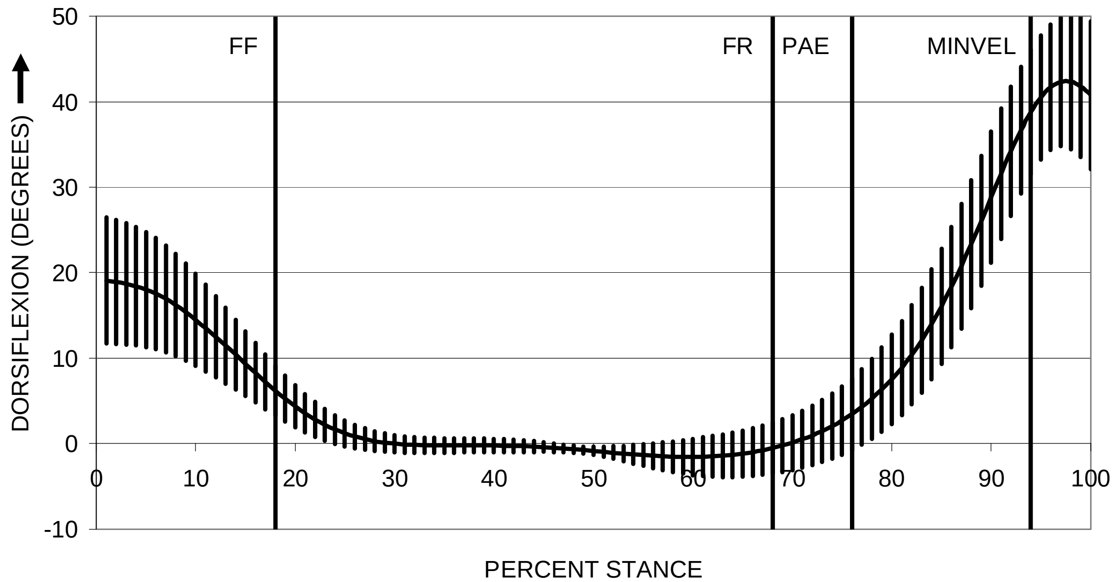


Figure 5-5. Ensemble average (± 1 SD) transverse plane kinematics of the forefoot relative to the calcaneus during the stance phase of gait.

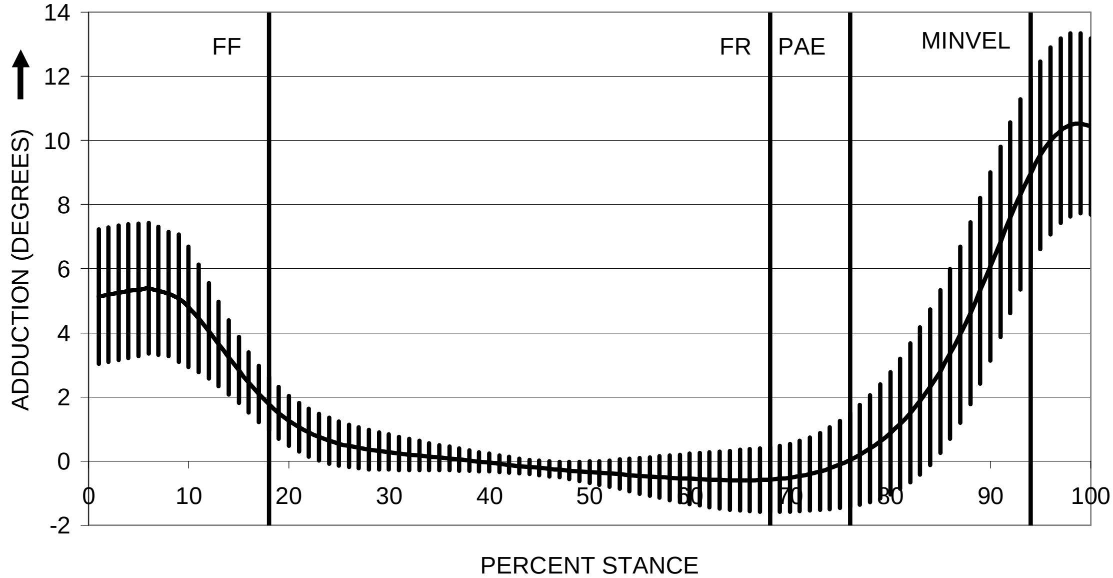


Figure 5-6. Ensemble average (± 1 SD) frontal plane kinematics of the calcaneus relative to the tibia during the stance phase of gait.

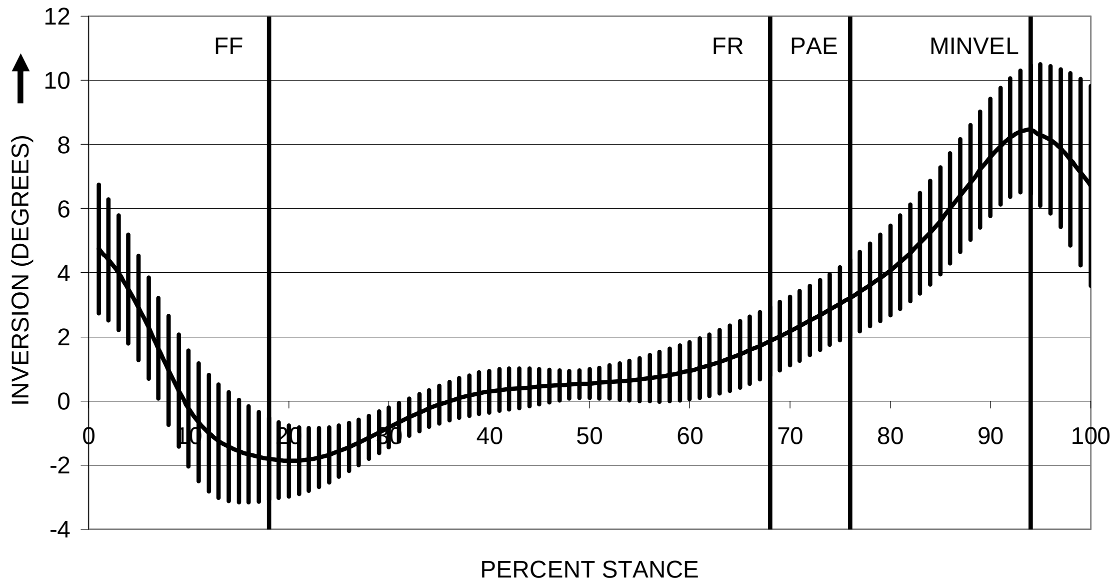


Figure 5-7. Ensemble average (± 1 SD) transverse plane kinematics of the calcaneus relative to the tibia during the stance phase of gait.

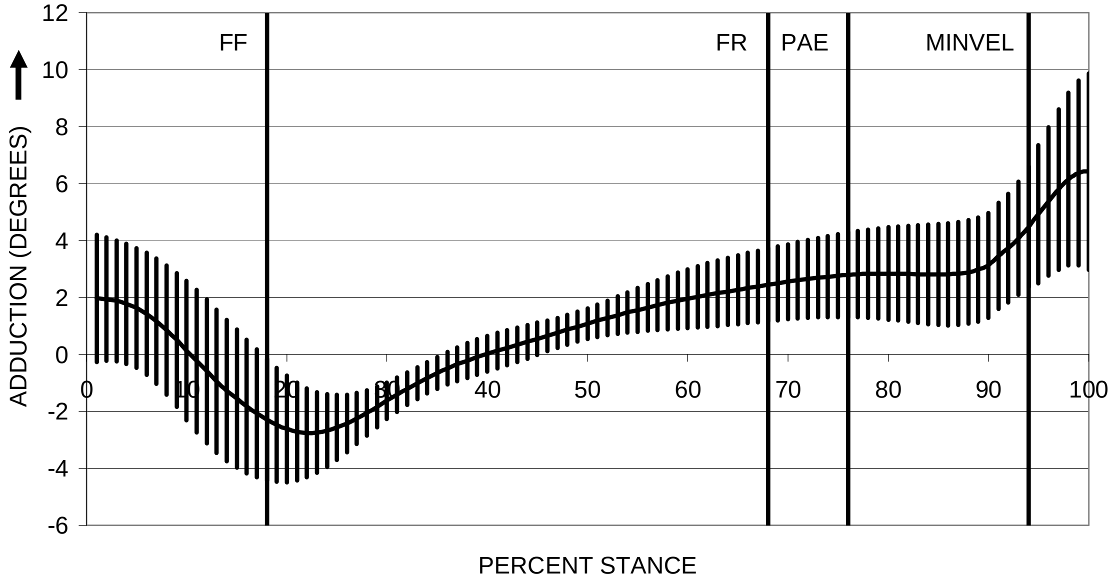


Figure 5-8. Ensemble average (± 1 SD) kinematics of A) frontal plane motion of the calcaneus relative to the tibia B) transverse plane motion of the forefoot relative to the hindfoot C) sagittal plane motion of the first metatarsal relative to the calcaneus and D) sagittal plane motion of the hallux relative to the first metatarsal during the stance phase of gait.

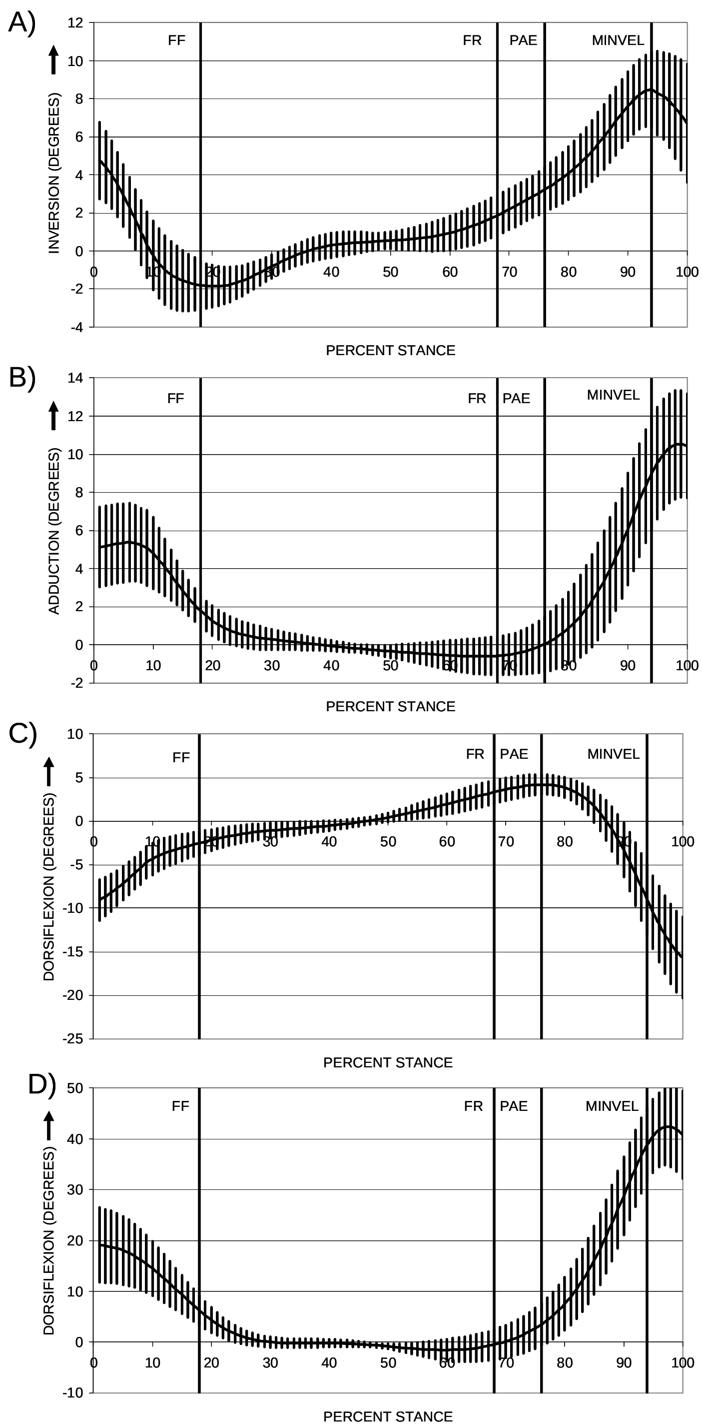
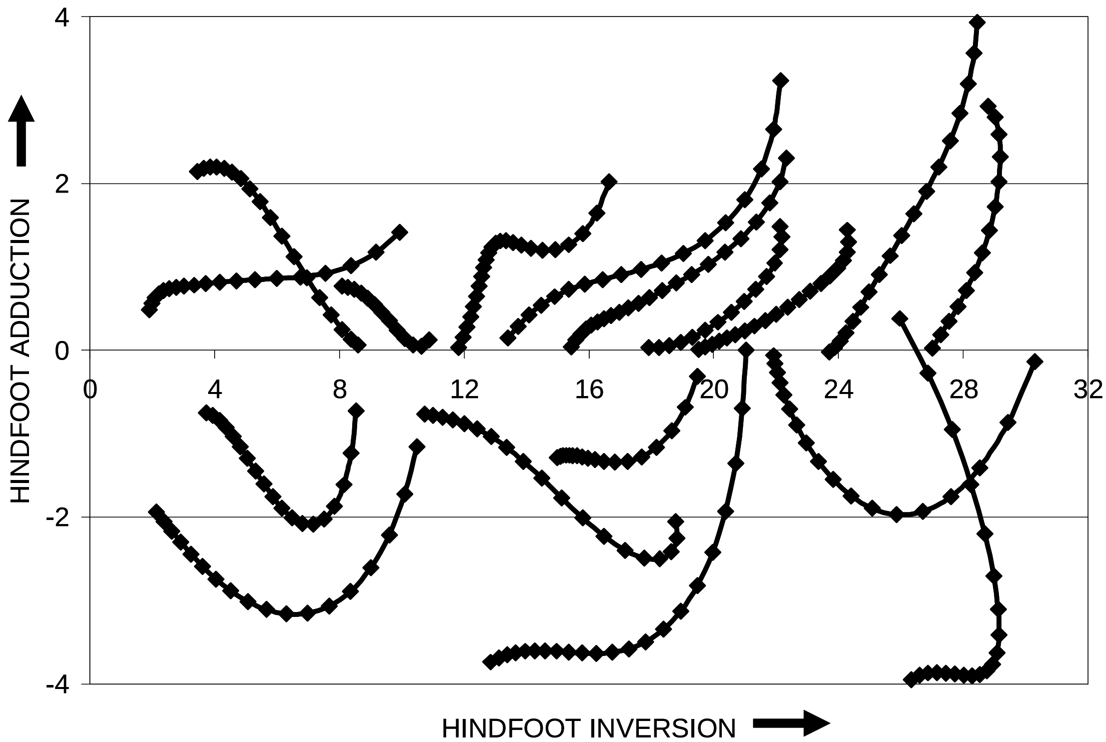


Figure 5-9. Kinematic curves demonstrating coupling of frontal and transverse plane motion between the calcaneus and tibia (n=17) from peak arch elongation (PAE) until the minima in the first metatarsal forward rotation velocity (MINVEL). The X and Y axes provide a reference point for magnitude of motion only. Linear regression was used to determine coupling ratios with the requirement that the minimum correlation coefficient for the data set was r= 0.75. Curves were shifted to differentiate between the 10 subjects that met the r > 0.75 criteria (above X axis) and those that did not (below X axis).



CHAPTER VI

CONCLUSIONS

The work summarized in the preceding chapters provide new insight into dynamic function of the foot, and the medial longitudinal arch in particular, during the stance phase of gait. Current theory regarding foot function during gait suggests that the structure and mobility of the medial longitudinal arch plays an important role in determining the magnitude of motion that occurs between segments of the foot during gait. Current theory is, however, primarily driven by clinical experience and in-vitro testing with little in-vivo data to support hypotheses regarding how the foot functions during gait. Questions regarding the validity of current theory demonstrate a need for in-vivo examination of foot function and re-examination of conclusions based solely on in-vitro testing methods that can't fully replicate dynamic function.

The results of the preceding chapters provide insight into 1) dynamic function of the arch and the validity of the truss and tie rod to model arch function, 2) the coordinated motions associated with the windlass mechanism resulting in supinatory motion at the end of stance phase and 3) the role of arch height on foot mobility and joint coupling.

The results presented in chapter two provide new insight into sagittal plane arch motion during the stance phase of gait. While examining the effect of ankle fusion and arthroplasty on dynamic foot function it was observed that arch motion occurred in a manner different than commonly accepted. Though arch function has long been modeled using the truss and tie rod the current use of this model does not adequately represent arch function. It was previously accepted that as the foot is loaded during the stance phase of gait both the proximal first metatarsal and distal calcaneus undergo considerable lowering to the floor resulting in first metatarsal dorsiflexion and arch lowering. This mechanism is, however, inconsistent with observed kinematics indicating that the calcaneus rotates about a nearly stationary first metatarsal for nearly 50% of stance.

These results, when combined with plantar pressure data (Warren, Maher et al. 2004), suggest that the current truss and tie-rod model should be modified to account for a nearly stationary first metatarsal as well as the role of tension produced by the plantarflexors which rotate and unload the calcaneus as the tibia progresses over the foot. Additional insights into arch function include the remarkable walking velocity independence of sagittal plane arch motion, the importance of walking velocity on the timing of first metatarsal rotation as well as the negligible effect of complete loss of ankle motion on sagittal plane arch kinematics.

The results presented in chapter five provide new insight into the series of events that lead to activation of the windlass mechanism and supinatory motion during the end of stance phase. Recent studies (Hunt and Smith 2004) have called into question the concept of mid-stance supination as a mechanism for forming a rigid lever for pushoff. The results of this study clearly demonstrate that supination does occur but due to activation of the windlass mechanism late in stance it occurs later than previously suggested. Examination of intersegmental coupling during this interval clearly demonstrates a rapid and linear motion into supination after hallux dorsiflexion generates sufficient tension in the plantar fascia to elicit forefoot plantarflexion.

In-vitro study and current theory suggest that arch height is a key determinant of structural stability of the foot. Though increased mobility associated with decreased arch height is commonly accepted in clinical education and accommodating for supposed changes in motion is commonly the goal of clinical intervention little in-vivo evidence exists to support these claims (Hunt, Fahey et al. 2000; Hunt and Smith 2004). Consistent with previous studies arch height was not found to influence dynamic function (Hunt, Fahey et al. 2000; Hunt and Smith 2004). A subject population without foot pathology that represented a wide range of arch heights was recruited in an effort to maximize the possibility of detecting an association between arch height and foot function during gait. In spite of these efforts there was clearly no association between arch height and intersegmental foot kinematics during the FF-PAE and PAE-MINVEL intervals.

The results of this study provide new insight into dynamic foot function and demonstrate the necessity for subsequent studies to examine the veracity of long held beliefs which guide management of foot and ankle pathology. The results of this study suggest arch height does not influence foot mobility. It is, however, possible that deviations from normal foot structure contribute to the development of pathology as a result of abnormal loading. Estimates of plantar fascia loading in individuals can be obtained by using plantar pressure data collected under the hallux, joint kinematics as collected in this study and radiographic measures. Assessment of the effect of arch height on in-vivo plantar fascia loading may confirm or refute assumptions regarding the role of foot structure in the development of foot pathology.

Currently there is a limited understanding of how articular geometry and plantar fascia tension interact to produce supination of the foot after activation of the windlass mechanism. It is possible that the saddle shaped articular surface of the calcaneocuboid joint serves as a pivot point allowing the hindfoot to rotate relative to a nearly stationary and stable forefoot. Further study is necessary to determine how the shapes of the hindfoot and midfoot articular surfaces allow rotation of the hindfoot relative to a nearly stationary first metatarsal during midstance.

Insights gained through this work may prove valuable for understanding the development and progression of clinical conditions such as hallux valgus. Though motion of the fist metatarsal relative to the floor is unaffected by arch height, ankle fusion or ankle replacement the potential exists for alterations in first metatarsal kinematics to contribute to foot pain and progressive foot deformity. Altered timing or magnitude of first metatarsal or first metatarsal phalangeal joint motion would likely alter the distribution of load within the foot and influence the ability of the foot to supinate in terminal stance. Further investigation is necessary to determine if foot pathology results in deviations from the highly consistent timing and magnitude of motion observed in this study.

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