

2011

The Effect of Shoe Forefoot Stiffness On The Windlass Mechanism In Running

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The effect of shoe forefoot stiffness on the windlass mechanism in running

by

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A thesis submitted to the graduate faculty
in partial fulfillment of the requirements for the degree of
MASTER OF SCIENCE

Major: Kinesiology (Biological Basis of Physical Activity)

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2011

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Chapter 1. Introduction

In the constant re-evaluation of the influence of footwear on sport performance, an increasing emphasis is being placed on footwear that improves performance. In running footwear research, an emphasis has been placed on investigation of the cushioning for impact protection, and construction to correct excessive foot motion. Features enhancing performance have largely been related to decreasing weight of footwear (Shorten, 2000).

One parameter that has positive effects on performance is an increased midsole bending stiffness about the medio-lateral (ML) axis of the shoe. Increased stiffness in this region is thought to affect the function of the metatarsophalangeal joint (MP joint). Wearing a stiff shoe has resulted in a 1% metabolic energy saving during running and a 1.7 cm increase in jump height (Roy & Stefanyshyn, 2006, Stefanyshyn & Nigg, 2000). Similar increases in jump height, and decreased time to complete a cutting drill were the result of wearing footwear with greater ML bending stiffness (Tinoco, Bourgit, Moran, 2010).

While the improved performance was quantified, a mechanism for these increases was not determined. There was not an increase in shank muscular activity, and the metatarsophalangeal joint actually absorbed more energy than was generated during the push off phase in running (Roy & Stefanyshyn, 2006, Stefanyshyn & Nigg, 1997). This could indicate altered function of passive tissues—ligament, tendon, and the plantar fascia of the feet are responsible for the performance increases. Morio et al. (2009) compared running barefoot to running in sandals with varying bending stiffness.

It was determined that the sandals restricted the torsion of the foot in comparison to barefoot conditions.

The windlass mechanism is a complex anatomical function of the plantar fascia working in response to metatarsal-phalangeal (MTP) dorsiflexion (Hicks, 1954). The plantar fascia spans the length of the foot, and becomes taut under dorsiflexion. It aids in maintaining the integrity of the medial longitudinal arch, and during running ensures that the foot is stiff enough to propel the body forward. Furthermore, the function of the windlass mechanism and the force generated by calf muscles are highly interdependent. Carlson, Fleming, & Hutton (2000) documented that as MTP joint dorsiflexion increases, the truss formed by the plantar fascia increases the stiffness of the foot. In turn, a larger magnitude of force transfers through the triceps surae to the foot, enhancing applied forces.

Assessment of foot motion is difficult during shod conditions. The most common method involves placing retroreflective markers on the surface of the shoe. The upper material of footwear prevents placement anywhere on the foot below the malleoli of the tibia. Material movement not representative of foot movement can often give exaggerated results of foot motion when contrasting motion of the foot assessed with intracortical markers, and markers placed on the shoe. Stacoff et al. (2000) found that markers placed on the shoe were tracking movement of the shoe, rather than movement of the foot within the shoe. Different shoe constructions yielded large effects with the use of markers placed on the shoe, but the same effects were not found utilizing intracortical markers. Intracortical pins pose a greater health risk to the

participants, require medical staff for insertion, and require rehabilitation for the participant and are not allowed for use in the United States. While they directly measure the movement of the bone they are attached to, these markers are not practical for most studies.

In order to utilize intracortical pins without excessive detriment to participants, researchers often use cadaver specimens outfitted in machinery to mimic gait. This may allow for use of an invasive method like intracortical pins, but the scope of the study is restricted by the ability to recreate normal gait. These movements do not fully replicate running gait, and still require specialized staff to prepare specimens.

Dynamic fluoroscopy provides a solution that allows for participants to complete shod dynamic movements while still assessing motion of multiple bones or segments of the foot. However, there are only a handful of labs with adequate machinery to conduct these studies. Furthermore, the procedure exposes the participants to the radiation of x-rays.

Placing markers directly on the skin of the foot allows for more representative motion analysis, but does not allow for analysis of footwear effects.

The purpose of the present study is to develop and validate a technique to assess arch motion during shod running. A secondary purpose is to determine the effect of increased forefoot bending stiffness on the arch during running. Due to the difficulty in assessment of the arch during shod conditions, a modified running shoe will be utilized to allow for motion capture of arch dynamics in running. It is hypothesized that the

kinematics of the arch, foot, and shank will be restricted with a stiff longitudinal bending stiffness, particularly during the push off phase in running.

Chapter 2. Review of Literature

Running Biomechanics and Footwear

During running locomotion, the foot serves unique, multiple purposes through the gait cycle. When the foot contacts the ground, it provides shock absorption in order to dissipate the ground reaction forces. During midstance or static standing, the foot supports body weight. During pushoff, the foot forms a rigid lever to overcome the force of gravity and friction to propel the body forward (Whiting & Zernicke, 1998). The phases of the gait cycle are most commonly described as ground contact, weight acceptance, midstance, pushoff, and toe off (Vito & Kalish, as cited in Donatelli, 1996).

At ground contact, the subtalar axis is supinated relative to the tibia. During stance, the subtalar joint rotates inward into a further supinated position. This added motion allows the foot to become more flexible to accommodate the high forces encountered during running. Maximum pronation occurs at the end of the stance phase, and the foot then re-supinates at the subtalar joint, ensuring that the foot is rigid to propel the body forward (Norkin & Lavangie, 2001 as cited in Bolga & Malone, 2004).

The use of radiographic analysis of the foot during running has led to the identification of strain placed on the plantar fascia during running. As pronation occurs during the first phase of stance, the distance between the calcaneus and metatarsals increases, which increases plantar fascia stress (Vito & Kalish, as cited in Donatelli, 1996). When the foot supinates to increase stiffness during propulsion, the windlass effect occurs through the dorsiflexion of the phalanges, and stress increases in the plantar fascia.

The focus of footwear biomechanics is most commonly focused on injury prevention and performance improvement. A brief survey of available running footwear reveals a wide variety of options to choose from. Outside of stylistic components, there are four basic features in shoe construction that affect the foot-ground interface (Shorten, 2000). The most basic function of a running shoe is to protect the plantar surface from abrasion or puncture. Through lightweight design or enhanced traction, shoes can increase running performance. Cushioning helps to reduce the impact forces runners experience at foot strike. Shoe design can also influence foot motion, particularly at the subtalar axis, with the aim of preventing injury associated with excessive or mistimed subtalar motion.

Through a complex chain, the foot's interaction with the ground affects tibial rotation. As the vertical force increases beneath the foot during running, the bones comprising the arch of the foot collapse, causing translation between the talus and tibia that results in tibial rotation. While this is necessary for adequate range of motion to complete a step, excessive rotation is linked to chronic injury such as patellofemoral pain syndrome and shin splints (Butler, Hamill, Davis, 2006). By strategically placing firmer material beneath the medial arch, the deformation of the arch is restricted, subsequently affecting tibial rotation.

During running, the entire body is active in a complex chain of movements as it interacts with the ground. Bellchamber & van den Bogert (2000) compared lower limb kinematics during walking and running. During walking, the flow of power travels from proximal to distal, which means the foot is following the movement of the body.

However, during running, there are periods during the stance phase where this power flow is reversed, indicating the foot influences tibial rotation. This indicates that the use of orthotics largely re-distributes forces during walking, but during running, they could directly influence lower limb kinematics.

Motion control footwear that features firm material beneath the medial-longitudinal arch of the foot, along with corrective orthotics, are common interventions to minimize injury associated with excessive lower extremity eversion and tibial rotation. Utilizing intracortical bone pins with markers attached, Stacoff et al. (2000) determined a minimal effect of footwear, and footwear plus orthotic use versus barefoot running that was nonsignificant and nonsystematic. The effect of the different interventions was less than a few degrees, while between-subject variation was often approaching 10°. The authors conclude that any effect of orthotics will vary from individual to individual.

Arch type (high, low) has been documented to have an interaction with footwear designed to correct excessive tibial rotation (Butler, Davis, & Hamill, 2006). High arched runners had decreased loading rates while wearing motion control footwear, while loading rates increased for low arched runners.

Assessment of Footwear Effects on Gait

Traditionally modeled as a single rigid segment, the foot is better represented through a multi-segment model that offers gait analysis with greater detail. The foot is comprised of 26 bones, and has 10 joint articulations in the midfoot and rearfoot alone. These joints have complex translational and rotational motion that is difficult to assess,

even with the use of intracortical bone pins. Wolf et al. (2008) identified four segments within the foot: calcaneus, navicular-cuboid, medial cuneiform-first metatarsal, and fifth metatarsal. These units can be utilized to better assess the motion of the foot during gait. In the present study, the fifth metatarsal was excluded due to a negligible windlass effect at this joint.

Using motion analysis to analyze the effect of footwear is problematic. Foot motion is assessed through the use of retroreflective markers placed on the article of footwear. This limits the ability to distinguish between segments of the midfoot. Skin surface markers are difficult to utilize to model the foot, particularly during shod conditions. The upper of the footwear prevents placement anywhere on the foot below the malleoli of the tibia, and the midsole and outsole construction confound results about actual foot motion during gait. When contrasting motion of the foot assessed with intracortical markers, and markers placed on the shoe, a large degree of variation is observed (Stacoff et al. 2000). Furthermore, different shoe constructions yielded large effects with the use of shoe markers, but the same effects were not found utilizing intracortical markers. For the present study, the upper material was cut out in strategic locations to allow for the placement of skin surface markers without compromising the structure of the footwear.

Intracortical pins pose a greater health risk to the participants, require medical staff for insertion, and require rehabilitation for the participant and are barred from use in the United States. While they directly measure the movement of the bone they are attached to, these markers are not practical for most studies.

In order to utilize intracortical pins without excessive detriment to participants, researchers often use cadaver specimens outfitted in machinery to mimic gait. While this allows for use of such an invasive method, these movements do not fully replicate running gait, and still require specialized staff to prepare specimens.

Dynamic fluoroscopy provides a solution that allows for participants to complete shod dynamic movements while still assessing motion of multiple bones or segments of the foot. However, there are only a handful of labs with adequate machinery to conduct these studies. Furthermore, the procedure exposes the participants to the radiation of x-rays.

Placing markers directly on the skin of the foot allows for more representative motion analysis, but does not allow for analysis of footwear effects (Butler, Hamill, Davis, 2006).

Effect of Footwear on the Windlass Mechanism in Running

Within running shoes, there is almost always a trade-off between function and performance. A shoe with features designed to prevent excessive foot and medial longitudinal arch motion is heavier because of additional plastic or denser midsole materials. This, however, could pose an issue for running economy since each 100g increase in weight increases oxygen consumption approximately 1% (as cited in Shorten, 2000). Furthermore, the addition of firmer plastic and foam, or the addition of more foam to increase cushioning parameters also influences the longitudinal flexibility of footwear, which can also affect performance (Park, Choi, & Lee, 2007).

Oleson et al. (2005) calculated the stiffness of the MTPJ during running and compared it to the bending stiffness of running shoes. In calculating the stiffness, the bending moment is divided by the angle of flexion (Stefanyshyn & Nigg, 1998). As the maximum bending moment and maximum angle of flexion occur at the same point in stance, the authors suggest there is a strong active component to MTPJ dorsiflexion and is not solely caused by passive tissues, such as the plantar fascia. This agrees with findings reported earlier linking increased strain on the plantar fascia to increased Achilles tendon force.

A series of research has investigated the effect of footwear on enhancing athletic performance, focusing on the effect of increasing the longitudinal bending stiffness of a shoe. Footwear with midsoles stiffened by a plate of carbon fiber is shown to have positive effects on athletic performance. In separate experiments, stiffened shoes resulted in a 1% metabolic energy savings during running and a 1.7 cm increase in jump height (Roy & Stefanyshyn, 2006, Stefanyshyn & Nigg, 2000). Similar increases in enhanced jump height, as well as decreased time to complete a cutting drill were found (Tinoco, Bourgit, Moran, 2010). However, in each of the previous studies, the authors could not explain the performance increases, as there were no changes in joint energy absorbed or generated when compared to a control shoe. During running, there was not a significant difference in lower-limb muscle activity. The findings together demonstrated no difference in generation of energy at toe-off and the lack of active muscle increases while increasing numerous performance factors of both slow twitch and fast twitch muscle fibers. Therefore, it is likely that passive structural components,

such as tendon, ligament, and the plantar fascia, are accounting for the additional work of bending a stiffened shoe.

Morio et al. (2009) utilized articles of footwear with differing midsole hardness to determine the effect on walking and running kinematics. They concluded that the shoe constrained torsion and adduction ranges of motion during running when compared to barefoot running. Alteration in sole hardness between shod conditions was achieved through using more dense midsole foam in one condition. This alters both the flexibility and the cushioning properties of the footwear, creating two variables that could elicit a kinematic response.

Shoes that alter the resting dorsiflexion angle of the metatarsals have been shown to alter the muscle activity in the leg during various fitness exercises, walking, and running (Bourgit, Millet, & Fuchslocher, 2008). During lunge and squat exercises, the tibialis anterior and leg extensors had increased activation when the metatarsals were moderately dorsiflexed (2° , 4°). During running, a more complex relationship exists. During the eccentric phase, any increase in dorsiflexion angle resulted in higher triceps surae activation. The authors speculated that higher activation could affect leg stiffness during this phase. In the concentric phase, there was no significant increase in muscle activation due to the increased dorsiflexion angle of the metatarsals.

Windlass Mechanism In Running

First described by Hicks (1954), the windlass mechanism describes a passive structural change of the medial longitudinal arch of the foot. Spanning the length of the foot from the calcaneal tuberosity to the base of the phalanges, the plantar aponeurosis,

or plantar fascia, is the key tissue responsible for the windlass mechanism. The windlass is created by the plantar fascia wrapping around the metatarsal heads. When a phalange is dorsiflexed, the plantar fascia is pulled in the anterior direction. This forces the arch of the foot to increase in height and shorten in length. Likewise, when the weight of the body is applied to a foot flat on a surface, the arch is flattened as the toes plantarflex and release the tension of the plantar fascia.

The windlass mechanism is present in each ray of the foot, but its effect is greatest at the first metatarsal and least at the fifth metatarsal. This is due to the increased radius and range of motion of the first metatarsal, which leads to the generation of greater tension on the medial aspect of the plantar fascia (Hicks, 1954).

Bojsen-Moller (1979) conducted a series of experiments to determine the importance of the medial longitudinal arch during walking and described two axes of push-off as high-gear and low-gear. High-gear occurs in a transverse axis through the first and second metatarsal heads, and low-gear through an oblique axis from the second through fifth metatarsal heads. Due to the increased action of the windlass effect during high-gear running, it was proposed that this is a more effective motion during locomotion.

Factors Affecting Plantar Fascia

Lee, Hertel, and Lee (2010) measured kinetics and kinematics of walking and running, placing emphasis on measuring the change in arch height of the foot. Dynamic arch height and plantar fascia stress were calculated using markers placed at the calcaneus, navicular tuberosity, and first metatarsal head. Approximately 82% of the

variance in plantar fascia stress during the stance phase of running could be accounted for by arch height and increased rearfoot eversion. Orthotic devices designed to reduce plantar fascia stress are documented to be most effective when they support the bony structures of the medial longitudinal arch (Kogler, Solomonidis, & Paul, 1996). By physically limiting the distance the arch can travel, the strain placed on the plantar fascia is reduced, relieving pain caused by plantar fasciitis.

The role of the plantar fascia becomes more complex when considering that it is connected at the base of the calcaneus. Because the Achilles tendon also inserts on the calcaneus, the plantar fascia is coupled with plantarflexor musculature in the calf. When tensile force increases in the Achilles tendon, strain in the plantar fascia increases (Carlson, Fleming, & Hutton, 2000). Furthermore, as the dorsiflexion angle of the metatarsophalangeal joint (MTPJ) increases, the plantar fascia stiffens. This, in turn, increases the tensile force applied by the Achilles tendon.

Injury

Running injuries are most often classified in terms of a “stress” or “fatigue” type injury. The mechanism for these injuries is the cumulative effect of the repetitive loading the body experiences while running (Whiting & Zernicke, 1998).

Plantar fasciitis, or the inflammation of the plantar fascia, is a very common injury. Approximately 10% of runners are affected by this condition (Chandler & Kibler, 1993). The mechanism of the injury is not entirely understood, but it is directly related to excessive stress placed on the plantar fascia. While altered or improper kinematics certainly can increase loading and stress placed on the plantar fascia, Cornwall (2000)

pointed out that the joints and tissues of the foot regularly function beyond a normal range. Therefore, an effect of fatigue from repetitive loading must play a role in the development of this injury.

Chapter 3. The effect of shoe forefoot stiffness on the windlass mechanism in running

A paper to be submitted to
Footwear Science

Eric G. Sterner and Timothy R. Derrick

Abstract

Research evaluating the effects of running footwear on gait has deduced foot motion from upper and sole movement of footwear. The aim of the present study was to evaluate a technique that allows for direct assessment of the medial longitudinal arch (MLA) in running. A secondary purpose was to evaluate the effect of increased bending stiffness of footwear on the MLA during running. Using a unique marker set, a multi-segment foot model was created to analyze dorsiflexion of the first metatarsal, navicular displacement, rearfoot motion, and tibial rotation. Virtual markers were created based on the movement of these foot segments. Two different pairs of running shoes (flexible, stiff) were evaluated. 13 participants ran barefoot, and in both shoe conditions. The mean difference between actual and virtual markers created was 0.69 mm. Independent t-tests determined first metatarsal dorsiflexion was restricted in the stiff condition compared to the flexible ($p < 0.05$) with an effect size of 0.36. The study provides a useful method of assessing foot motion while wearing footwear, and indicates that stiffer shoes restrict foot motion.

Introduction

Assessment of foot motion is difficult during shod conditions. The upper material of footwear prevents placement anywhere on the foot below the malleoli of the tibia, and the midsole and outsole construction confound results about actual foot motion during gait. When contrasting motion of the foot assessed with intracortical markers, and markers placed on the shoe, a large degree of variation is observed (Stacoff et al. 2001). Furthermore, different shoe constructions yielded large effects with the use of markers placed on the shoe, but the same effects were not found utilizing intracortical markers. Other methods of analysis such as using intracortical pins or dynamic fluoroscopy are invasive, pose extra risk to the participants, and require specialized staff such as doctors and x-ray technicians to execute.

One parameter that has positive effects on performance is an increased midsole bending stiffness about the transverse axis of the shoe. Increased stiffness in this region is thought to affect the function of the metatarsophalangeal joint (MP joint). Wearing a stiff shoe has resulted in a 1% metabolic energy saving during running and a 1.7 cm increase in jump height (Roy & Stefanyshyn, 2006, Stefanyshyn & Nigg, 2000). Similar increases in enhanced jump height, and decreased time to complete a cutting drill were also the result of wearing footwear with greater longitudinal bending stiffness (Tinoco, Bourgit, Moran, 2010).

While the improved performance was quantified, a mechanism for these increases was not detected. There was not an increase in shank muscular activity, and the metatarsophalangeal joint actually absorbed more energy than was generated

during the push off phase in running (Roy & Stefanyshyn, 2006, Stefanyshyn & Nigg, 1997). This could indicate altered function of passive tissues—ligament, tendon, and the plantar fascia of the feet are responsible for the performance increases. Morio et al. (2009) determined that stiff shoes restrict the natural motion of the foot during running, specifically during the push off phase in running.

The windlass mechanism is a complex anatomical function of the plantar fascia working in response to phalangeal dorsiflexion (Hicks, 1954). The plantar fascia spans the length of the foot, and becomes taut under dorsiflexion. It aids in maintaining the integrity of the medial longitudinal arch, and during running ensures that the foot is stiff enough to propel the body forward. Furthermore, the function of the windlass mechanism and the force generated by calf muscles are highly interdependent (Carlson, Fleming, & Hutton, 2000). Optimal function of the windlass mechanism is essential to an effective running gait.

The purpose of the present study is to develop and validate a technique to assess arch motion during shod running. A secondary purpose is to determine the effect of increased forefoot bending stiffness on the arch during running. Due to the difficulty in assessment of the arch during shod conditions, a modified running shoe will be utilized to allow for motion capture of arch dynamics in running. It is hypothesized that the kinematics of the arch, foot, and shank will be restricted with a stiff longitudinal bending stiffness, particularly during the push off phase in running.

Methods

Participants

Fifteen males were recruited for participation in the present study. Potential participants were excluded if they had any lower extremity surgery, deformity, or current pain. Inclusion criteria included a weekly mileage of at least 10 miles per week, and fitting a size 9 shoe. Experiment procedures were approved by the university Institutional Review Board and all participants gave their written informed consent to participate. The participants received compensation for their time.

Due to tracking error, data from 2 participants were excluded from the study. The means and standard deviations of subject characteristics of age (years), height (m), weight (kg), weekly mileage (miles) are shown in Figure 1.

Protocol

One visit to the lab was required for each participant. Following review of the IRB document and consent, anthropometric measurements were collected. Age (years), height (cm), weight (kg), weekly mileage (miles) were recorded. A total of 3 three-marker wands were attached using cyanoacrylate on the anterior-lateral surface of the hallux, between the navicular tuberosity and the center of the 1st cuneiform, and on the calcaneus (Nielsen et al. 2010). Athletic tape was adhered around the hallux wand, and partially around the navicular-cuneiform wand. Additional markers were placed at the distal and proximal ends of the 1st metatarsal, dorsifoot, navicular tuberosity, and 2 centimeters distal from medial aspect of the heel, and 2 centimeters above the ground.

Shank markers were placed on the lateral and medial malleolus, distal and proximal tibia, calf, medial and lateral knee, and sacrum.

A static standing trial was recorded while the participant was barefoot. Following the standing trial, participants were instructed to run barefoot, at a self – selected speed, over a force platform collecting at 1600 Hz (AMTI, Watertown, MA, USA). Motion analysis cameras (Vicon, Oxford, UK) simultaneously collected marker position at 200 Hz. Following 5 barefoot trials, another standing trial was collected. Markers placed on the distal and proximal 1st metatarsal, navicular tuberosity, dorsifoot, medial heel, medial ankle, and medial heel were removed.

The participants were then instructed to put on shoes that had portions of the footwear upper cut out in the areas over the wands. Two pairs of the same model of shoe were utilized (New Balance, Boston, USA) with the only change being a stiff thermoplastic polyurethane layered between the outsole and the midsole of one pair. The shoes were evaluated for their midsole height at the heel and forefoot, cushioning (peak g's), and forefoot bending stiffness (Nm/deg), as shown in Figure 2. The pairs were ordered, to ensure balanced presentation of conditions. The participants then completed 10 running trials in each pair of shoes, at a speed controlled to $\pm 5\%$ of their preferred running speed.

Following the running trials, the markers adhered with cyanoacrylate were removed with acetone.

Data Analysis

Kinematic data were low-pass filtered with a fourth-order Butterworth filter with a 16 Hz cut-off frequency in Matlab (7.8.0 R2009a Natick, MA). Ground reaction force (GRF) data remained unfiltered. Matlab was used to calculate all variables. The standing trials were used to create virtual markers at the locations of the actual markers that were removed during the shod conditions (Soderkvist and Wedin, 1993). Variables calculated were the calcaneal-phalangeal angle, dynamic arch height, calcaneal inversion/eversion, and tibial rotation. The order of rotation for calculation of kinematic data followed zxy (vertical, medial lateral, anterior posterior). The reference position for all kinematic data was the static standing trial collected at the start of each condition. These data were filtered with a cubic spline to create curve values at each 1% of the stance phase of running. The filtered data was then averaged to create ensemble curves as shown in Figures 1-4. Parameters of interest were the peak calcaneal-phalangeal angle, minimum and maximum navicular displacement, maximum calcaneal eversion, maximal calcaneal inversion, maximum tibial rotation, and stance time. The shod conditions of flexible and stiff were compared using IBM SPSS Statistics 19 (IBM Corp., Somers, NY). A two-tailed independent T-test with an alpha of 0.05 was used to determine if the differences between means was statistically significant. Effect sizes were also calculated with an effect size of 0.20, 0.50, and 0.80 indicating a small, medium, or large effect size, respectively (Cohen, 1992).

Results

Average running speed was not significantly different for the barefoot (BF), Flexible (FL), or stiff (ST) conditions (3.18 ± 0.64 3.58 ± 0.55 3.63 ± 0.64 m/s, respectively). The maximum calcaneal-phalangeal angle was significantly different for flexible ($26.18^\circ \pm 4.48$) and stiff ($24.52^\circ \pm 4.74$), $t(13) = 4.682$, $p < 0.01$. As shown in Figure 1, the variables of navicular displacement, maximum rearfoot eversion, maximum rearfoot inversion, tibial rotation, and stance time did not show differences that were statistically significant (Table 3). The effect size for the calcaneal-phalangeal angle was 0.359, which is between small and moderate. Ensemble curves of the rearfoot angle and navicular displacement are shown in Figures 2 and 3, respectively. Neither of these parameters demonstrated statistical significance, and both had negligible effect sizes. Aside from the small-moderate effect size of the calcaneal-phalangeal angle, all other effect sizes were less than 0.15, which does not even classify as small.

Figure 4 displays the ensemble curve for the actual navicular marker, and a reconstructed virtual marker. The mean difference between the actual navicular marker and the virtual navicular marker was 0.69 mm during the barefoot running condition. The greatest difference between actual and virtual position occurs during the last 15% of stance.

Discussion

The purpose of this study was to develop an effective method for studying multi-segment models of the foot during running. The second purpose was to compare the effect of shoe forefoot stiffness on foot kinematics during running.

Direct comparison between barefoot and shod running were not assessed, as the primary purpose of the barefoot trials was to establish the position of the virtual markers during gait. Therefore, running velocity was not controlled during barefoot trials as it was during the shod conditions.

The use of virtual markers to determine medial longitudinal arch deformation proved successful. This method allowed nonstructural modifications of the footwear to be utilized in order to assess foot function during running. As previously documented, markers placed on footwear may not accurately reflect foot segment motion (Stacoff, Nigg, Reinschmidt, van den Bogert, Lundberg, 2000). By using marker triads we were able to locate markers within the segment reference system during a standing barefoot trial and then remove them and locate them virtually during shod running trials. During the barefoot running trials both virtual and actual markers could be compared.

High-speed video of the midfoot triad indicated substantial oscillation of this wand during barefoot running. Therefore, the midfoot triad was secured with a combination of elastic and athletic tape. It is possible that the tape could have restricted arch motion but it is assumed that it would have the same effect in both shoe conditions. The mean difference between the virtual and actual navicular marker during the

barefoot running was 0.69 mm. The difference was greatest during the pushoff phase, when the skin covering the arch is pulled tight. It was not possible to determine if the actual or the virtual marker was a more accurate representation of the arch movement during this period.

The peak dorsiflexion angle of the calcaneal-phalangeal angle demonstrated a statistically significant reduction during running in the stiff shoe versus the flexible shoe. The difference of 1.65 degrees had a small effect size (0.36). Reductions in minimum and maximum navicular displacement, average maximum eversion, and maximum tibial rotations also occurred. While these parameters all demonstrated nonsignificant differences, and effect sizes less than 0.10, these results correspond to previous research that established that footwear restricted foot motion relative to barefoot kinematics (Morio, Lake, Gueguen, Rao, Baly, 2009). There is also a consistent trend demonstrating reduced range of motion across segments. Lee, Hertel, and Lee (2010) utilized a similar marker set, and established that rearfoot eversion explained 82% of plantar fascia tension (measured by change in navicular height). Furthermore, the tibia articulates atop the calcaneus, which is the anatomical site assessed for rearfoot motion. This suggests that altering the bending stiffness of footwear can influence multiple foot segments

The reduction in the peak calcaneal-phalangeal angle indicates a reduced windlass effect during running. As the angle is reduced at this joint, the windlass occurs to a lesser degree. This would likely place less strain on the plantar fascia and could lead to inefficient foot function, affecting the overall range of motion of the foot during

gait. On the other hand it could provide some relief to runners suffering from plantar fasciitis.

Stiffness of the footwear in this study differed greatly between the flexible and the stiff conditions. The stiffness for the stiff shoe was measured to be 5 times greater than the flexible shoe. However, even the stiff shoe is more flexible than many running shoes on the market. The stiffness of the footwear utilized in the present study is comparable to the stiffness of models used by Oleson, Adler, and Goldsmith (2005), who concluded that the stiffness of the foot is dominant to the stiffness of the footwear. For most movement, the foot should not be affected by the stiffness of the shoe. They suggested that small differences in stiffness could have an effect late in the gait cycle, in the pushoff phase of running. The data of the present study correspond to these findings, as the increased shoe stiffness affected the peak dorsiflexion angle near pushoff, but did not have a significant effect on other variables measured.

The present results reveal a useful method to model the foot with multiple segments while participants are shod. This creates an enhanced ability to assess foot function during physical activity in shoes with varying features. While the shoes utilized in the present study only revealed statistical significance for one parameter measured, multiple other variables revealed small differences between the two articles of footwear. Future research should include footwear with a greater difference in forefoot bending stiffness to induce a greater effect. Utilizing a larger marker set to include the knee and hip joints would also provide a more complete analysis of gait.

Conclusions

The intent of the present study was to develop and assess the usefulness of a new method to assess foot motion during running. The results indicate the proposed method could be used as a tool to distinguish small effects induced by differences in footwear. The second purpose was to evaluate the effect of shoe stiffness on the windlass mechanism. Results indicate some altered function, and agree with previous research indicating restricted movement in stiffer footwear. The effects on performance are not known. Future research should include a greater treatment effect, and a direct comparison to barefoot conditions.

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Tables

Table 1. Means (Standard Deviations) of 13 participant characteristics.

Mass (kg)	Height (m)	Age (y)	Weekly Mileage
67.3 (6.3)	1.77 (0.05)	22 (5.0)	37.5 (23.3)

Table 2. Characteristics of the footwear utilized in the study.

Shoe	Manufacturer	Model	Rearfoot height (mm)	Forefoot height (mm)	Rearfoot impact (g)	Forefoot impact (g)	Forefoot flexibility (Nm/degree)
1	New Balance	Flexible	23.70	14.8	13.50	20.90	0.73
2	New Balance	Stiff	23.90	16.1	11.90	24.80	0.45

Table 3. Means (Standard Deviations) of lower limb kinematics during running.

Variable	Barefoot	Shod - Flexible	Shod - Stiff
Running Velocity (m/s)	3.18 (0.64)	3.58 (0.55)	3.63 (0.64)
Maximum Calcaneal-Phalangeal Angle (deg)	32.21 (5.91)	26.18 (4.48)*	24.52 (4.74)*
Minimum Navicular Displacement (mm)	23.72 (5.89)	22.55 (7.68)	21.77 (7.12)
Maximum Navicular Displacement (mm)	31.73 (6.45)	30.03 (7.61)	29.83 (7.50)
Rearfoot Eversion (deg)	-5.13 (3.16)	-6.41 (4.03)	-6.21 (3.86)
Rearfoot Inversion (deg)	9.12 (2.88)	8.11 (3.89)	7.54 (3.93)
Tibial Rotation (deg)	1.97 (2.00)	2.08 (1.82)	1.86 (1.83)
Stance Time (s)	0.229 (0.03)	0.233 (0.03)	0.236 (0.03)

*Denotes a significant difference between shod conditions.

Figures

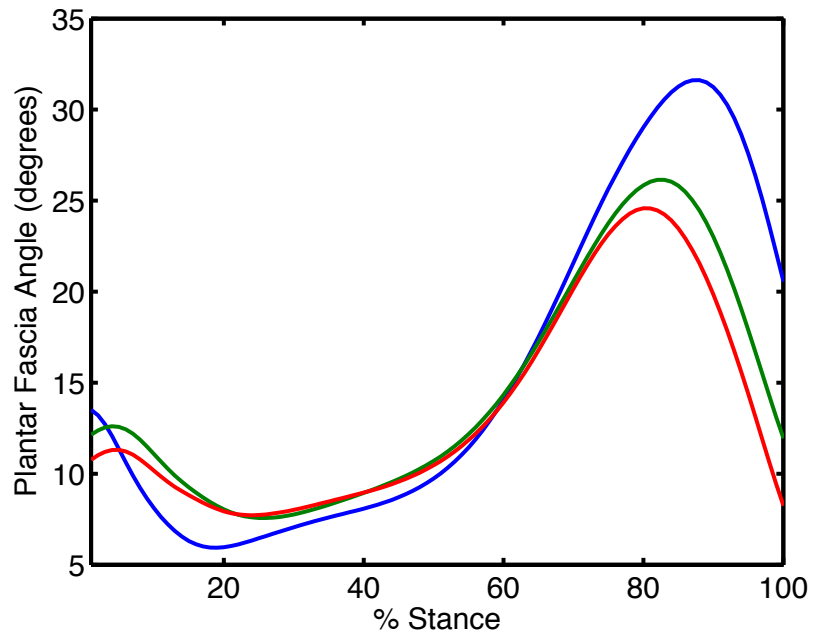


Figure 1. Ensemble curve of the calcaneal-phalangeal angle during stance for barefoot (blue), flexible (green), and stiff (red) conditions.

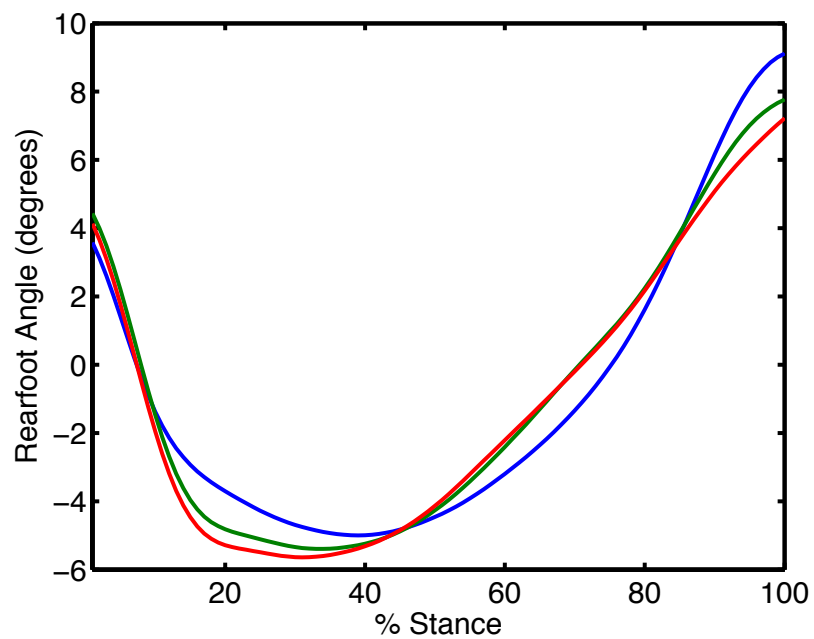


Figure 2. Ensemble curves of the rearfoot angle during stance for barefoot (blue), flexible (green), and stiff (red) conditions.

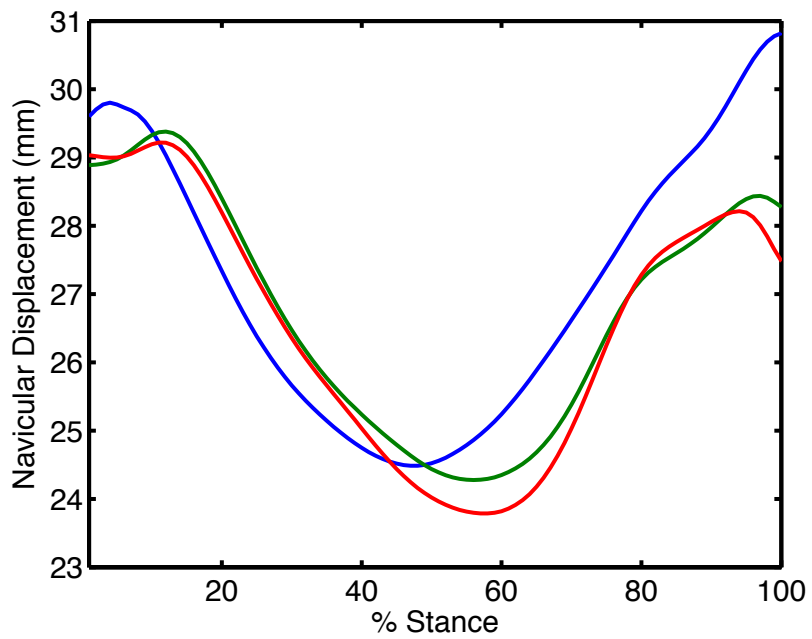


Figure 3. Ensemble curves of the average navicular displacement for the barefoot (blue), flexible (green), and stiff (red) conditions.

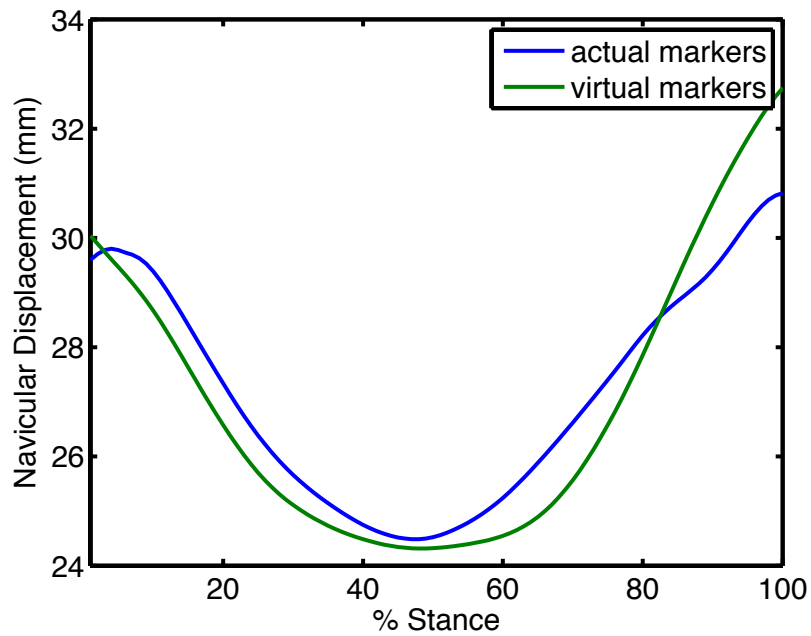


Figure 4. Ensemble curve of the navicular displacement during the barefoot condition.