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A thesis submitted to the faculty of
Brigham Young University
in partial fulfillment of the requirements for the degree of

Master of Science

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ABSTRACT

The Role of the Midfoot in Drop Landings

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Purpose: The contribution of the midfoot in landing mechanics is understudied. Therefore, the main purpose of this study was to quantify midtarsal joint kinematics and kinetics during a barefoot single-leg landing task. A secondary aim of this study was to explore the relationship between static foot posture and dynamic midfoot function. Methods: In a cross-sectional study design, 48 females (age = 20.4 ± 1.8 yr, height = 1.6 ± 0.06 m, weight = 57.3 ± 5.5 kg, BMI = 21.6 ± 1.7 kg·m^{-1}) performed drop landings from a height of 0.4 m onto split force platforms. Subjects hung from wooden rings and landed on their dominant leg. Midtarsal joint kinematic and kinetic data were recorded using a motion capture software system in conjunction with a custom multisegment foot model marker set. Arch height index (AHI) for both seated and standing conditions was measured using the Arch Height Index Measurement System (AHIMS). Results: Kinematic data revealed an average sagittal plane midtarsal range of motion (ROM) of 27 degrees through the landing phase. Kinetic data showed that between 7% and 22% of the total power absorption during the landing was performed by the midtarsal joint. Standing AHI was correlated negatively with sagittal plane midtarsal ROM (p = 0.0264) and positively with midtarsal work (p = 0.0212). Standing midfoot angle (MA) was correlated positively with sagittal plane midtarsal ROM (p = 0.0005) and negatively with midtarsal work (p = 0.0250). Conclusion: The midfoot contributes substantially to landing mechanics during a barefoot single-leg landing task. Static foot posture may be a valuable measurement in predicting midfoot kinematics and kinetics.

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Introduction

The foot is a critical structure in the production of human locomotion and athletic performance. If problems arise at the foot, they can impair normal biomechanics resulting in injuries up the kinetic chain (1). Despite the many intrinsic structures of reinforcement, problems related to improper foot function are still common. A healthy foot acts as a dynamic base of support, providing rigidity and elasticity during various phases of walking, running, jumping and landing. It provides shock absorption (2), stores and returns elastic energy (3), and quickly adapts to external stimuli to maintain and restore balance (4). The role of the midfoot is of particular interest when considering foot mechanics. It is understudied and may provide greater insight regarding how the foot functions both in athletics and activities of daily living.

With technological advancements, multisegment foot models are becoming a more commonly used method for quantifying midfoot kinematics and kinetics (5,6). Recent studies have explored the multisegment nature of the foot and how individual segments contribute to overall motion and function (6–8). The majority of this research has focused on walking and running gait to observe midfoot mechanics. For example, Dixon et al. used the Oxford foot model to study midfoot power generation in walking (9). De Mits et al. had healthy subjects walk barefoot to analyze midfoot kinematics using the Ghent model, a six-segment foot model used by physical therapists and podiatrists (10). Another study examined midfoot motion in high- and low-arched individuals during running (11). However, few studies have specifically explored the midfoot in drop landings. These studies have shown that peak vertical ground reaction forces (vGRF) can reach 3.5–7.1 times body weight (12) and elicit a greater midfoot range of motion (ROM) when compared to walking and running. These heightened impact forces, repeated over time, affect the musculoskeletal system and can be a source of bone and soft tissue damage (12).

Therefore, a controlled drop-landing task may be a valuable mechanism to stress the foot system beyond walking and running. The extra stress placed on the foot during a drop landing may provide further understanding into the loading response that occurs at the midfoot during high-impact movements consistent with sport-related activities.

Studying a high-impact dynamic task like landing may also help determine the relationship between static foot posture and dynamic midfoot function. Clinical interventions for foot injury rehabilitation and prevention are often prescribed based on static measures of foot posture. Measuring arch height index (AHI) is one static method that is widely used to classify foot type and predict foot function during dynamic activities (13). The ease and convenience of measuring static AHI in both research and clinical settings, as well as its structural representation of the medial longitudinal arch, make AHI a commonly used measurement. However, this convention merits further investigation as the relationship between static foot posture and dynamic midfoot function remains unclear in the literature. Studies exploring walking and running gait have concluded that there is little to no evidence to suggest a relationship between static measures of foot posture and dynamic midfoot kinematics (11,14,15). Conversely, emerging studies that also compared static foot posture to midfoot kinematics in walking gait, contradict this claim and have found significant relationships (13,16). Other studies have found a relationship between static and dynamic foot measures during walking and jogging. However, these have only compared postural and not kinematic measures (17,18). Observing a connection between static foot posture and midfoot kinematics and kinetics would be clinically relevant in identifying contributing factors to poor foot function or foot injury under increased loading conditions. Without a proven relationship, it seems that the use of static measures as the basis for injury rehabilitation and prevention has limited practical application.

The main purpose of this study was to investigate the role of the midfoot by exploring its kinematics and kinetics during a barefoot single-leg drop landing. Our secondary aim was to investigate the relationship between static and dynamic measures of the midfoot. We theorized that the midfoot would do substantial negative work that is unaccounted for using a single-segment foot model. We also hypothesized that static measures of foot posture would be significantly correlated with dynamic midfoot kinematics and kinetics.

Methods

Forty-eight healthy females (age = 20.4 ± 1.8 yr, height = 1.6 ± 0.06 m, weight = 57.3 ± 5.5 kg, BMI = 21.6 ± 1.7 kg·m^{-1}) were recruited from the college campus and surrounding area to participate in this study. Subjects were given a questionnaire inquiring about leg dominance, any past or present lower extremity injuries or abnormalities, as well as any medical diagnoses relating to balance. Leg dominance was defined as the preferred leg for kicking a ball [19]. Subjects were screened based on their responses to the questionnaire and excluded if they had a lower extremity injury or abnormality at the time of the study or within the past six months. No subjects reported having a condition affecting balance. Forty-nine subjects were recruited with only one subject being unable to complete the landing task due to physical limitations. All subjects signed an informed consent approved by the university Institutional Review Board before participation.

Sitting and standing foot posture measurements for the dominant foot were obtained using the Arch Height Index Measurement System (AHIMS) [20–22]. Sitting AHI and standing AHI were calculated as dorsum height divided by truncated foot length, as described by Williams et al. [23]. AHI stiffness was calculated as the difference between sitting and standing

AHI divided by the sitting AHI and multiplied by a factor of 10^4 over BW (kg), according to the formula presented by Nigg et al. (24).

Twenty-eight skin surface markers were placed on specific anatomical landmarks of the pelvis, thigh, knee, lower leg (shank), ankle and foot of the dominant landing limb, following the kinetic multisegment foot model created by Bruening et al. (25,26). The retroreflective markers were prepared and attached using double-sided toupee tape safe for skin contact (Fig. 1-A). All landings were performed barefoot.

Kinetic data were collected using two in-ground force platforms (Advanced Mechanical Technology, Inc, Watertown, MA, USA) at a sampling rate of 1000 Hz. A 14-camera motion capture system (Vicon, Motion Capture Systems, Ltd., Oxford, UK) was used to collect kinematic data sampled at 250 Hz. Specific camera placement and video calibration was completed prior to data collection. A static trial was first captured with subjects in equal weight-bearing stance on one force platform with their feet shoulder-width apart and arms crossed over their chest. The static trial was processed and analyzed immediately to ensure that all markers appeared in the camera view before proceeding with further data collection. From this trial, static midfoot angle (MA) was calculated using markers from the first metatarsal head, navicular tuberosity, and posterior base of the calcaneus. Subjects performed all trials facing the researcher in order to effectively respond to verbal and visual cues.

In an attempt to increase foot and ankle contribution during the landing phase, subjects performed single-leg drop landings from a height of 0.4 m (12,27). Subjects hung from wooden gymnastic rings and were directed to relax their shoulders to obtain an accurate and consistent drop height for each trial. A hanging drop landing was employed to better simulate a sports-like landing, which can be difficult to recreate from a box drop. The same drop height was measured

separately for each subject. This distance was measured with a ruler from the force platform to the plantar aspect of the heel directly in line with the lateral malleolus. Subjects received visual and verbal cues for when they should let go of the rings and land in a natural way on their dominant leg. Although landing style was not controlled, all participants landed in a toe to heel pattern. Multiple landing attempts were collected until at least three successful trials were acquired. A successful trial constituted a natural landing in which the navicular and cuboid markers aligned with the split between the two forces platforms, effectively resulting in a rear foot and forefoot impact on separate plates. To avoid subjects targeting the correct landing location and potentially interfering with their natural landing strategy, ropes attached to the hanging apparatus were used to position subjects directly over the split in the force platforms, ensuring a successful landing (Fig 1-A).

Data from the landing trials were labeled and tracked using Vicon Nexus 2.5 and then imported into Visual 3D (C-Motion, Inc., Germantown, MD, USA) for further analysis and calculation of key metrics. Marker data were low-pass filtered at 6 Hz while forces were low-pass filtered at 100 Hz. The kinetic multisegment foot model was built using the static trial and applied to all landing trials. Our model follows the reliable and validated model created by Bruening et al. (25,26) with only minor modifications in tracking targets (Fig 1-B). This multisegment model partitions the foot into three distinct segments based on specific anatomical landmarks (Fig 1-C). These include a rear foot (calcaneus and talus), a midfoot (navicular, cuboid, cuneiforms and metatarsals) and a toe (proximal and distal phalanges) segment. The rear foot and midfoot segments are connected by the midtarsal joint whereas the midfoot and toe segments are connected by the metatarsophalangeal (MP) joint. It should also be noted that this

model includes a shank segment that is connected to the rear foot by the ankle joint, and a thigh segment connected to the shank by the knee joint.

From three successful trials, averages were calculated and used to represent each subject for data analysis. The data time curves were cut to when subjects let go of the rings (start drop) through the minimum vertical point of their center of mass (minCOM), representing the power absorption phase of the landings. For visual aid in interpreting the data, events were made at initial contact (IC) with the force platform (20 N threshold) and peak vGRF. A typical Euler rotation sequence (1-sagittal, 2-frontal, 3-rotation) was used to calculate joint angles between segments for the static trial and from start drop through minCOM for the dynamic trials. Sagittal plane ROM was calculated for the midtarsal and metatarsophalangeal (MTP) joints as the difference between peak flexion and peak extension angles of the respective proximal and distal segments relative to each other. Frontal plane ROM was calculated for the midtarsal joint only and used the difference between peak inversion and peak eversion angles of the respective proximal and distal segments relative to each other. Joint moments, powers, and work were calculated using inverse dynamics. Power absorption was calculated as the integral of joint power over time and represented by negative work (28). Work values were calculated for the hip, knee, and single-segment ankle (ankle SS), as well as for multisegment ankle (ankle MS), and multisegment midtarsal (midtarsal MS) joints. These were used for negative work comparison purposes (Fig 2). Midtarsal stiffness in the sagittal plane was calculated as the sagittal plane midtarsal moment over the sagittal plane midtarsal angle, and one subject's data was chosen to graphically represent this metric for the whole sample size (Fig 3-D). All kinetic variables were normalized by BW.

Data were analyzed, and comparisons of negative ankle work between a traditional single segment foot model and multisegment foot model were made. Pearson product-moment correlation coefficients (PCC) were computed to assess the static-dynamic relationship using SAS (SAS Institute, Inc, Cary, NC). Comparisons were made between the two different methods of measuring static AHI as well as between three static foot posture variables and five dynamic midfoot variables.

Results

Individual joint contributions of the hip, knee, ankle, and midtarsal to total work during the drop landing for both single segment and multisegment model analyses are represented in Figure 2. A single segment analysis of joint work showed the knee extensors were eccentrically doing the majority of the work (-1.33 ± 0.32 J·kg^{-1}) to decelerate the body followed by the hip extensors (-1.19 ± 0.43 J·kg^{-1}) and ankle plantar flexors (-1.19 ± 0.26 J·kg^{-1}) sharing equally the remainder of the work. However, multisegment modeling approximated the ankle MS work at -0.75 ± 0.24 J·kg^{-1} and midtarsal MS work at -0.45 ± 0.13 J·kg^{-1} during landing. A single segment foot model overestimated the ankle's contribution to power absorption on average by 0.44 J·kg^{-1} when compared to our multisegment foot model. The variation within subjects was large with a range of ankle power absorption being overestimated by 18-68%.

Sagittal plane midtarsal ROM averaged 27.04 ± 6.92 degrees whereas frontal plane midtarsal joint ROM averaged 5.81 ± 2.38 degrees (Fig 3-A). The midtarsal joint at start drop began in flexion and inversion and remained in that position until just prior to initial contact (IC) where it increased in flexion slightly before moving into extension and eversion. The midtarsal joint rapidly went through the full extension ROM beginning with IC and past peak vGRF then began to flex and invert again as the subject approached minCOM.

Sagittal plane midtarsal moments, representing the net torque at the joint, are shown in Figure 3-B. Midtarsal moments increased to a maximum shortly before peak vGRF and slowly began to decline as subjects approached minCOM. Midtarsal power averaged -10.79 ± 3.89 $W \cdot kg^{-1}$ and peaked between IC and peak vGRF, returning to baseline before minCOM (Fig 3-C). Sagittal plane midtarsal stiffness showed a linear trend as the midtarsal joint moved from flexion into extension and as midtarsal moments increased (Fig 3-D).

Sagittal plane MTP joint ROM averaged 24.14 ± 8.74 degrees through landing (Fig 4). The MTP joint began in extension and continued extending to almost 20 degrees until just before IC. To prepare for contact, the toes went into further extension before touching down onto the force platforms. The peak extension angle occurred just after IC and then moved quickly toward maximal MTP flexion at peak vGRF before returning to a neutral position by minCOM.

Raw data with means and standard deviations for all static and dynamic variables can be referenced in Table 1. Several comparisons of static and dynamic foot measures were made and PCCs with p-values are listed in Table 2. A significant inverse correlation was found between standing AHI and static MA ($r = -0.6087$, $p < 0.0001$). Significant positive and negative relationships were found between standing AHI and both sagittal plane midtarsal ROM and midtarsal work. Static MA was also significantly positively and negatively correlated with the same dynamic variables.

Discussion

The midtarsal joint, also known as the transverse tarsal or Chopart's joint, refers to two distinct articulation sites between the talus and navicular (talonavicular) and the calcaneus and cuboid (calcaneocuboid) and forms an s-shaped joint. Although motion at this joint is convenient to observe and quantify, it should be understood that all of the tarsal bones contribute to

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complete midfoot motion. However, due to the limitation in surface marker technology and the inability to separate out all the individual joint movements between the tarsals, we classified the collective motion between the tarsal bones in the midfoot as midtarsal joint motion. Studies agree that more of the functional motion of the midfoot takes place proximally rather than distally (29,30). Therefore, our representation of midfoot motion will focus on the articulations of the midtarsal joint. The rationale for using a multisegment foot model was to be able to capture a more comprehensive representation of midtarsal joint motion and the contribution of the midfoot region to landing kinematics and kinetics.

To our knowledge, this is the first study to specifically explore midtarsal joint kinematics and kinetics during a barefoot single-leg landing. Many studies have examined midfoot motion and function during walking and running (5,9–11). However, research beyond these dynamic activities is scarce in the literature. As such, the main purpose of this study was to observe the role of the midfoot during drop landings. A secondary aim was to explore the relationship between static foot posture and midfoot function during a barefoot single-leg landing as this comparison has also not been reported in previous literature.

Our results suggest that there is a substantial amount of motion in the midfoot during a barefoot single-leg landing. We observed that sagittal plane midtarsal ROM averaged 27 degrees throughout the landing phase, which is significantly higher than reported in other studies. One study measured midfoot joint ROM and reported 5-8 degrees in the sagittal plane (10). However, these measures were calculated during walking where vGRF are much lower and may not produce the need for higher ROM at the midfoot. We found an average frontal plane midtarsal ROM of 3-9 degrees, which is more comparable to data from walking and running studies (9,10). At start drop, most subjects began with the ankle in nearly maximal plantar flexion, the midtarsal

joint flexed, and the toes slightly extended. This style of landing was not controlled but adopted naturally by subjects, likely in attempt to provide the greatest amount of ankle ROM possible during the landing. The observed increase in midtarsal joint flexion prior to IC may be due to activation of the tibialis posterior, tibialis anterior, and fibularis longus muscles, as these are the primary movers of the midtarsal joint (31,32). Muscle activation, with the purpose of further midtarsal joint flexion before impact, could be a proprioceptive mechanism to effectively increase the amount of available ROM needed for the landing. Our results showed that midtarsal joint mobility began before IC when the midfoot was flexed and inverted in preparation for landing. Midtarsal joint motion continued through peak vGRF, where it became increasingly mobile and capable of rapid decelerated extension assisted by the extrinsic and intrinsic muscles of the lower leg and foot (32,33). These findings are supported by recent research questioning the midtarsal locking theory, in reference to the midtarsal joint being rigid and restricted in motion during inversion (30,34). The midtarsal joint did not reach peak extension until after peak vGRF, potentially signifying continued absorptive efforts as the subject approached minCOM.

The amount of work done by the midfoot during landing warrants clinical attention. Work has generally been measured at the ankle, knee, and hip joints during landing and classified as power absorption (35). However, we were interested in understanding the contribution of the midtarsal joint to total power absorption. The midtarsal joint contribution to negative work continued past peak vGRF until minCOM, signifying active shock absorption throughout the entire impact phase of landing. Sagittal plane midtarsal moments showed a maximum joint torque occurring just prior to peak vGRF showing active midfoot flexors. Midtarsal powers were negative, confirming that midfoot flexors were eccentrically contracting to slow down midtarsal extension and absorb impact forces from the landing. Using a

multisegment foot model, we were able to calculate midtarsal work during the landing phase. Our results showed that the negative work done by the midtarsal joint was 52% of that done by the ankle, 31% of the knee, and 34% of the hip. Multisegment analysis showed that the midtarsal joint accounted for, on average, 11% of total work performed by the other lower extremity joints, including the ankle, knee, and hip. However, the midtarsal joint contribution to total negative work reached up to 22% is some subjects. This amount of absorption seems substantial enough to not be overlooked and may be clinically significant when prescribing rehabilitation and injury prevention interventions. While we only tested a single drop height of 0.4 m, this was chosen near the lower end of the range suggested by Zhang et al. to magnify ankle and midtarsal joint contribution to power absorption (28). From our single segment model approach, the ankle and hip contributed equally to total negative work performed, supporting the concept that a reduction in landing height would increase the contribution of the ankle joint. Further comparisons using a single segment foot model to calculate joint kinetics resulted in an overestimation of ankle work by an average of 38%. These results confirm claims made by other researchers that single segment foot models grossly overestimate ankle power absorption (9,26,36). Ankle power differences between models were due entirely to the differences in angular motion and velocity of the rear foot segment. This finding is supported by Dixon et al. who reported similar results when using the multisegment Oxford foot model for kinetic analysis of the foot and ankle during walking gait (9). While our split force platform approach may not be practical for in-field studies, accurate midfoot kinematics and kinetics should be calculated using a multisegment foot model as differences in landing mechanics may be undetectable with a single segment foot model (37).

The toes influence midfoot function during landing. This was seen as the toes assisted in midtarsal ROM when the MTP joint rapidly extended prior to IC. The MTP joint extended close to 20 degrees and caused the plantar fascia to tighten, engaging the windlass mechanism and pulling the midtarsal joint into further flexion prior to landing (26). During landings, the windlass mechanism may operate in conjunction with extrinsic and intrinsic foot musculature to engage the midfoot and provide a method for increased absorptive function of the midtarsal joint. In walking and running this mechanism acts as a means of power generation during the toe-off phase of gait (9). However, we observed the windlass mechanism, initiated by the toes, assisted in power absorption at the midtarsal joint. Understanding the involvement of the toes during the windlass mechanism and how they affect midtarsal motion may help clarify the role of the midtarsal joint in not only power generation during propulsion but as a means of increasing power absorption during landings.

Our dynamic landing task identified a potential relationship between static foot posture and dynamic midfoot function. In addition to our static-dynamic comparisons, we chose to compare two different types of static measures—standing AHI using the AHIMS device (7,20–22) and static MA from motion capture technology. The strong correlation between the two different measures of static foot posture suggests that either method is appropriate for use in clinical and research settings. However, static MA showed a stronger correlation to sagittal plane midtarsal ROM than standing AHI. It would seem that when comparing static and dynamic foot measures, it may be beneficial to use the same technology to reveal any existing relationships that are present. We also found a relationship between static foot posture and both dynamic midfoot variables (sagittal plane midtarsal ROM and midtarsal work). These findings are consistent with previous research which showed a significant relationship between static foot

posture and joint work in high and low arched females during landing (38). Our results showed that static foot posture was able to predict 10-32% of variation associated with dynamic midfoot function. McPoil et al., who showed similar findings in walking and running, suggest that clinicians could use static measures of foot posture to understand midfoot function without having to administer dynamic testing accompanied by complex collection processes and analyses (17). A significant static-dynamic relationship may have implications in injury prevention, especially for athletes who participate in high impact, repeated loading activities. Using static arch height measurements to predict midfoot landing mechanics could also help clinicians and researchers identify individuals that are at higher risk for injuries. Fraser et al. studied the association between lateral ankle sprains in people with chronic ankle instability and midfoot kinematics (39). They found that the midfoot was essential in force transmission and often injured in conjunction with lateral ankle sprains (39). Therefore, a static measure of foot posture that has the ability to predict midfoot kinematics and kinetics could potentially be used as a screening tool for lateral ankle sprains. Lastly, we did not find any significant relationship with AHI stiffness and any dynamic variables. We also noted the same lack of correlation with midtarsal stiffness and all dynamic variables, which was unexpected. Static and dynamic foot stiffness measures were included as variables of interest because arch stiffness is a clinical measure of foot function in response to a vertical load and may relate to injury risk (7,22). Zifchock et al. compared standing AHI to arch stiffness among 145 individuals and found a weak but significant relationship (22). We did not find a significant relationship between standing AHI and arch stiffness, but our sample size was much smaller. Regardless, our findings suggest that either arch stiffness is measuring something other than what we intend, or there needs to be new methods explored for quantifying arch stiffness.

Limitations

As with all research, there were several limitations to our study. First, our study was a cross-sectional design which did not allow us to draw specific causal conclusions but allowed us to identify relationships present within our specific population. We chose to use only females for our subject base which does not represent the whole the population. However, we chose this population because studies showed that compared to males, females exhibited more contribution to total power absorption from the ankle joint in landings (40). We did not control for landing strategy or allow subjects familiarization trials during data collection. We did collect multiple landing attempts until three successful trials were completed. Lastly, we used a multisegment foot model to measure sagittal plane midtarsal ROM, which is an improvement from traditional models but is still limited in identifying individual tarsal articulations.

Conclusion

This study focused on highlighting the role of the midfoot during a barefoot single-leg landing task through multisegment model analysis. Our observations suggest that the midtarsal joint experiences a large sagittal plane ROM and does contribute to power absorption during barefoot single-leg landings. Additionally, ankle power absorption is greatly overestimated when using a single segment foot model. Further analysis of the midtarsal joint motion should be done using a multisegment foot model. We also explored the theory of static foot posture predicting dynamic midfoot function. We found that both methods of measuring static arch height were moderate predictors of midtarsal kinematics and kinetics during a landing task. Static foot posture may be a valuable clinical tool in assessing midfoot function and injury risk in pathological and athletic populations.

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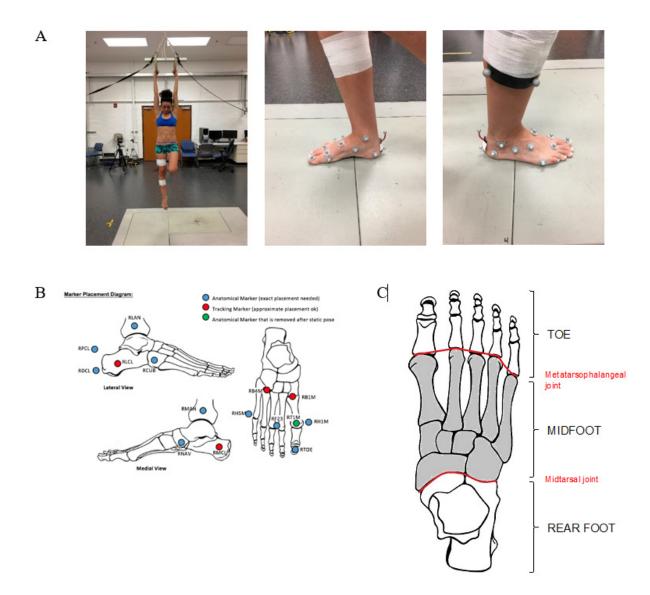
Figure 1. A) Hanging apparatus and force platforms for drop landings B) Customized kinetic multisegment foot model developed by Bruening et al., with slight modifications C) Foot segments separated by joints of interest.
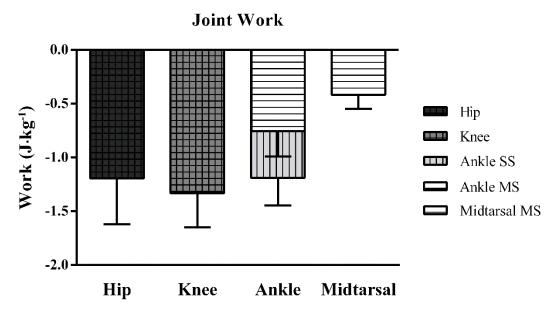
Figure 2. Joint work means and SD (error bars) for both single segment (vertical lines) and multisegment (horizontal lines) foot models.
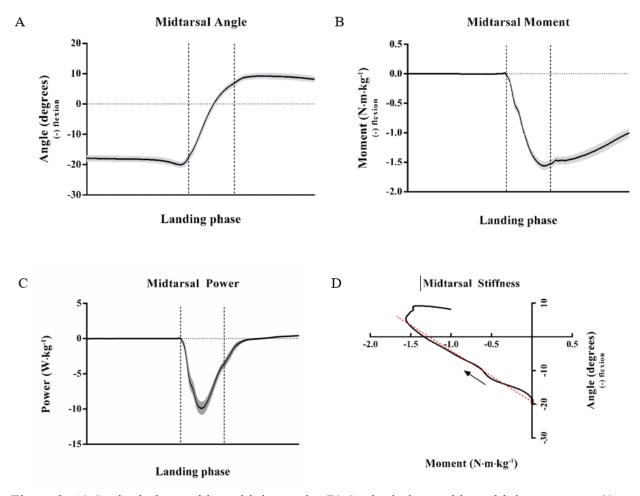
Figure 3. A) Sagittal plane midtarsal joint angles B) Sagittal plane midtarsal joint moments C) Midtarsal joint powers. Means represented with a solid black line. Gray error bands show standard error. Dashed vertical lines signify the events IC and peak vGRF. D) Midtarsal joint stiffness of a single representative subject throughout the landing phase. The arrow represents the direction of time. All data curves were time normalized for graphical presentation and ease of interpretation.

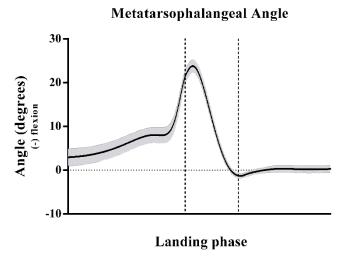**Metatarsophalangeal Angle**

Landing phase

Figure 4. Sagittal plane MTP joint angles. Mean represented with a solid black line. Gray error bands show standard error. Dashed vertical lines represent the events IC and peak vGRF. The data curve was time normalized for graphical representation and ease of interpretation

Table 1 Raw data with means ± SD (n = 48)

	Standing AHI	Static MA (deg)	AHI Stiffness	Midtarsal sagittal ROM (deg)	Midtarsal frontal ROM (deg)	Midtarsal Work $(J \cdot kg^{-1})$	Midtarsal Stiffness $(N \cdot m \cdot kg^{-1} \cdot deg^{-1})$
Mean ± SD	0.324 ± 0.019	−22.475 ± 5.229	16.845 ± 6.109	27.041 ± 6.916	5.812 ± 2.382	−0.415 ± 0.131	0.0651 ± 0.051

Table 2 Correlation coefficients and (p values) for static-dynamic variables (n = 48)

	Midtarsal sagittal ROM (deg)	Midtarsal frontal ROM (deg)	Midtarsal Work $(\text{J} \cdot \text{kg}^{-1})$	Midtarsal Stiffness $(\text{N} \cdot \text{m} \cdot \text{kg}^{-1} \cdot \text{deg}^{-1})$
Standing AHI	−0.32032 (0.0264)*	−0.08899 (0.5475)	0.33180 (0.0212)*	0.06821 (0.6450)
Static MA (deg)	0.48336 (0.0005)*	0.12346 (0.4032)	−0.32321 (0.0250)*	−0.16734 (0.2556)
AHI Stiffness	0.04686 (0.7518)	−0.07582 (0.6085)	−0.10313 (0.4855)	0.11016 (0.4560)

*significant at $p < 0.05$