2019

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Original Publication Citation

Olsen, M. T., Bruening, D. A., Johnson, A. W., & Ridge, S. T. (2019). The role of the midfoot in drop landings. Medicine and Science in Sports and Exercise, 51(1), 114-122. doi:10.1249/MSS.0000000000001765.114-122.

The Role of the Midfoot in Drop Landings

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Abstract

Purpose: The midfoot is instrumental to foot function; however, quantifying its roles in human movement has been difficult. A forceful dynamic activity like landing may help elucidate the midfoot's contribution to foot energetics and function. The main purpose of this study was to measure 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, body mass index = 21.6 ± 1.7 kg/m) 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 14-camera optical motion capture system in conjunction with two in-ground force platforms and a custom kinetic three-segment foot model. Foot structure was measuring using the arch height index (AHI) and the static midtarsal joint angle from motion capture.

Results: Kinematic data revealed an average sagittal plane midtarsal joint range of motion of 27° through the landing phase. Kinetic data showed that between 7% and 22% of the total lower extremity joint, work during the landing was performed by the midtarsal joint. Both standing AHI and static midtarsal joint angle (static MA) were correlated with sagittal plane midtarsal joint range of motion (standing AHI: $r = -0.320$, $P = 0.026$; static MA: $r = 0.483$, $P < 0.001$) and with midtarsal joint work (standing AHI: $r = 0.332$, $P = 0.021$; static MA: $r = -0.323$, $P = 0.025$).

Conclusion: The midfoot contributes substantially to landing mechanics during a barefoot single-leg landing task. Static foot posture measures have limited value in predicting midfoot kinematics and kinetics during sportlike landings.

Keywords: Midtarsal joint, Mulisegment foot model, Power absorption, Static-Dynamic, Female

The foot is a critical structure in human locomotion and athletic performance. 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 (1), stores and returns elastic energy (2), and quickly adapts to external stimuli to maintain and restore balance (3). The complex design of the foot helps accommodate a variety of mechanical stresses through the movement of the 26 bones and associated joints. Excessive loading or movement alterations may contribute to the development of musculoskeletal injuries at the foot and/or up the kinetic chain (4–6).

The midfoot likely plays an important role in modulating the rigidity and elasticity of the foot (7), but its contribution has been difficult to measure. As a result, describing and quantifying midfoot mechanics is understudied, and further investigation may provide insight regarding how the foot functions in both athletics and activities of daily living. With technological advancements, multisegment foot models are becoming more commonly used (8,9) to explore the manner in which individual segments contribute to overall motion and function. The majority of this research has focused on midfoot mechanics in walking and running gait (9–12) but has not been explored in drop landings. Peak vertical ground reaction forces (vGRF) can reach 3.5–7.1 times body weight during drop landings (5), which would likely elicit a greater midfoot range of motion (ROM) when compared with walking and running. The added stress placed on the foot during a drop landing may provide further understanding into midfoot mechanics of high-impact movements consistent with sport-related activities.

Foot function in dynamic activities is often considered to be related to foot structure. For example, static measures of midfoot posture, such as arch height, often influence the prescription of clinical interventions for foot injury prevention and rehabilitation (13,14). However, the relationship between static foot posture and dynamic midfoot function remains unclear in the literature. Most of the literature exploring these relationships has been confined to walking and running gait (14–17). Studying a high-impact dynamic task like landing may help elicit a stronger relationship between static foot posture and dynamic midfoot function. Observing a connection between static measurements of arch structure (such as arch height index [AHI]) and arch deformation or negative work performed by the arch during a landing could be clinically relevant in identifying contributing factors to musculoskeletal injuries in the foot occurring during high-impact activities. Because most clinicians do not have access to 3D motion capture and force plates, greater understanding of any relationship between measures of static and dynamic foot function may help clinicians make more applicable prevention or treatment recommendations for individual patients.

Both foot structure and dynamic foot function are influenced by a number of subject characteristics, including age and sex. Previous research has shown that older adults have flatter/more pronated feet, weaker toe plantarflexor muscles, higher prevalence of toe deformities, and reduced ROM (18). Compared with males, females have more flexible arches (19) and greater midfoot motion (20) during walking. Differences in landing patterns have also been studied between young males and females in an attempt to identify plausible causes for increased injury, such as ACL tears, in females (21). Decker et al. (21) found that females tend to use ankle musculature for impact absorption more than males, which caused them to be in increased knee extension and plantarflexion during contact with the ground. Considering the higher incidence of injury during landings in females and the increased reliance on the foot and ankle, a young female population is ideal for studying midfoot mechanics and investigating a relationship between structure and function.

The main purpose of this study was to investigate foot mechanics by exploring midfoot kinematics and kinetics during a barefoot single-leg drop landing in young females. We theorized that the midfoot would perform substantial negative work that is unaccounted for using a traditional single segment foot model. Our secondary aim was to investigate the relationship between static and dynamic measures of the midfoot during a high-impact movement that requires more midfoot involvement than gait. 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, body mass index = 21.6 ± 1.7 kg/m) from the college campus, and surrounding area completed 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 (22). Subjects were screened based on their responses to the questionnaire and excluded if they had a lower extremity injury, deformity, or condition affecting balance at the time of the study or within the past 6 months. Forty-nine subjects were recruited with only one subject withdrawing due to the inability to complete the landing task successfully. All subjects signed an informed consent approved by the university's 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) (19,23). Sitting AHI and standing AHI were calculated as dorsum height divided by truncated foot length, as described by Williams et al. (24). AHI stiffness was calculated according to Zifchock et al. (19).

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 (Fig. 1A). Kinetic data were collected using two in-ground force platforms (Advanced Mechanical Technology, Inc., Watertown, MA) 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. 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. This trial was used to calculate a static midtarsal joint angle (static MA).

Subjects performed single-leg barefoot drop landings from a height of 0.4 m. This height was chosen to emphasize foot and ankle contribution during the landing phase (5,25). Subjects hung from wooden gymnastic rings and were directed to relax their shoulders to obtain an accurate and consistent drop height for each trial (Fig. 1B). A hanging drop landing was used to accurately control drop height for subjects of differing heights, which may be difficult to recreate from a box drop (26). However, this novel method of performing drop landings does require that subjects have a grip strength sufficient enough to hold their own body weight. Drop height 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 verbal cues for when they should let go of the rings and land on their dominant leg. Landing technique was not controlled. Multiple landing attempts were collected until at least three successful trials were acquired (the maximum number of attempts was 9). A successful trial constituted a natural landing in which the navicular and cuboid markers aligned with the split between the two force platforms, effectively resulting in a rearfoot and forefoot impact on separate plates (Fig. 1C). To avoid subjects targeting the correct landing location and potentially interfering with their natural landing technique, ropes attached to the hanging apparatus were used to position subjects directly over the split in the force platforms (Fig. 1B).

Data processing was performed in Visual 3D software (C-Motion, Inc., Germantown, MD). Marker data were low-pass filtered at 6 Hz, whereas 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. The model was adapted from

Bruening et al. (27,28) with only minor modifications in tracking targets (Fig. 1A). This multisegment model partitions the foot into three distinct segments based on specific anatomical landmarks. These include a rear foot (calcaneus and talus), a mid/forefoot (navicular, cuboid, cuneiforms, and metatarsals) and a toe (proximal and distal phalanges) segment, separated by a midtarsal joint (located at the midpoint between navicular and cuboid markers, representative of multiple anatomical articulations) and a metatarsophalangeal (MTP) joint (center of first metatarsal head) (Fig. 1D). Because of the lack of standardized terminology used to describe foot function (29), throughout this manuscript, the following terminology will be used:

1. ''midtarsal joint'' refers specifically to the modeled joint center about which we calculated kinematics and kinetics
2. ''mid/forefoot'' refers specifically to the segment distal to the midtarsal joint (as described above)
3. ''midfoot'' is a more general term that is used in any context that is not specific to this study's midtarsal joint measurements, such as references to measurements from other studies (that used different foot models) and/or discussion of the function of the midfoot region (which includes multiple articulations, muscles, and connective tissue).

It should also be noted that our model includes a shank segment that is connected to the rear foot by an ankle joint (combined talocrural and subtalar joints) and a thigh segment connected to the shank by the knee joint. In addition, a traditional single rigid foot segment was created to compare against the multisegment foot model. A typical Euler rotation sequence (1 - sagittal, 2 - frontal, 3 - rotation) was used to calculate joint angles, with reference to the next proximal segment. Joint moments were calculated using inverse dynamics and expressed as internal moments in the proximal segment reference frame. Scalar joint rotational power was calculated as the dot product of the joint moment and joint angular velocity. Joint work was calculated as the integral of the joint power.

Data analysis consisted of time series waveforms and discrete metrics. Waveforms were used only to graphically represent the landing mechanics and were created as follows. First, a representative trial was chosen for each subject (to avoid overly smoothing peaks). Next, each waveform was cut into three phases using the following four events:

1. Release of the rings (start-drop)
2. Initial contact with the force platform at a threshold of 20 N (IC)
3. Peak vertical ground reaction force (peak vGRF)
4. The lowest vertical position of the subject's center of mass (minCOM)

From start-drop to IC was called the drop phase, from IC to peak vGRF was called the impact phase, and from peak vGRF to minCOM was called the final loading phase. Time normalization was done on each phase separately (i.e., 0%–100% of phase) and represented as a percentage of the phase duration. The relative duration of each phase was then adjusted using the mean absolute duration for that specific phase. For example, the drop phase was, on average, 2.25 times as long as the impact phase, and thus the time-normalized duration of the drop phase was also presented as 2.25 times as long as the impact phase. The mean and the SE were calculated across subjects for each normalized time point and displayed as mean waveforms with plus and minus SE bands.

An average of the three successful trials was used to calculate all discrete metrics. Joint ROM was defined as the difference between maximum and minimum angles throughout the landing trial (start-drop to minCOM). ROM was calculated for the ankle, midtarsal, and MTP joints in the sagittal plane and for only the ankle and midtarsal joints in the frontal plane. Joint work (6) was calculated for the hip, knee, ankle, and midtarsal joints as well as for the ankle using the single segment foot model (ankle SS). Work was normalized by body mass. Midtarsal stiffness was calculated as the linear portion of the sagittal plane midtarsal moment and angle plot (Fig. 2D). Metrics were calculated descriptively as mean and SD. Correlations were run between these dynamic metrics and the static foot posture variables (standing AHI, static MA, and AHI stiffness) using Pearson product–moment correlation coefficients (PCC). Statistical analysis was performed in SAS (SAS Institute, Inc., Cary, NC).

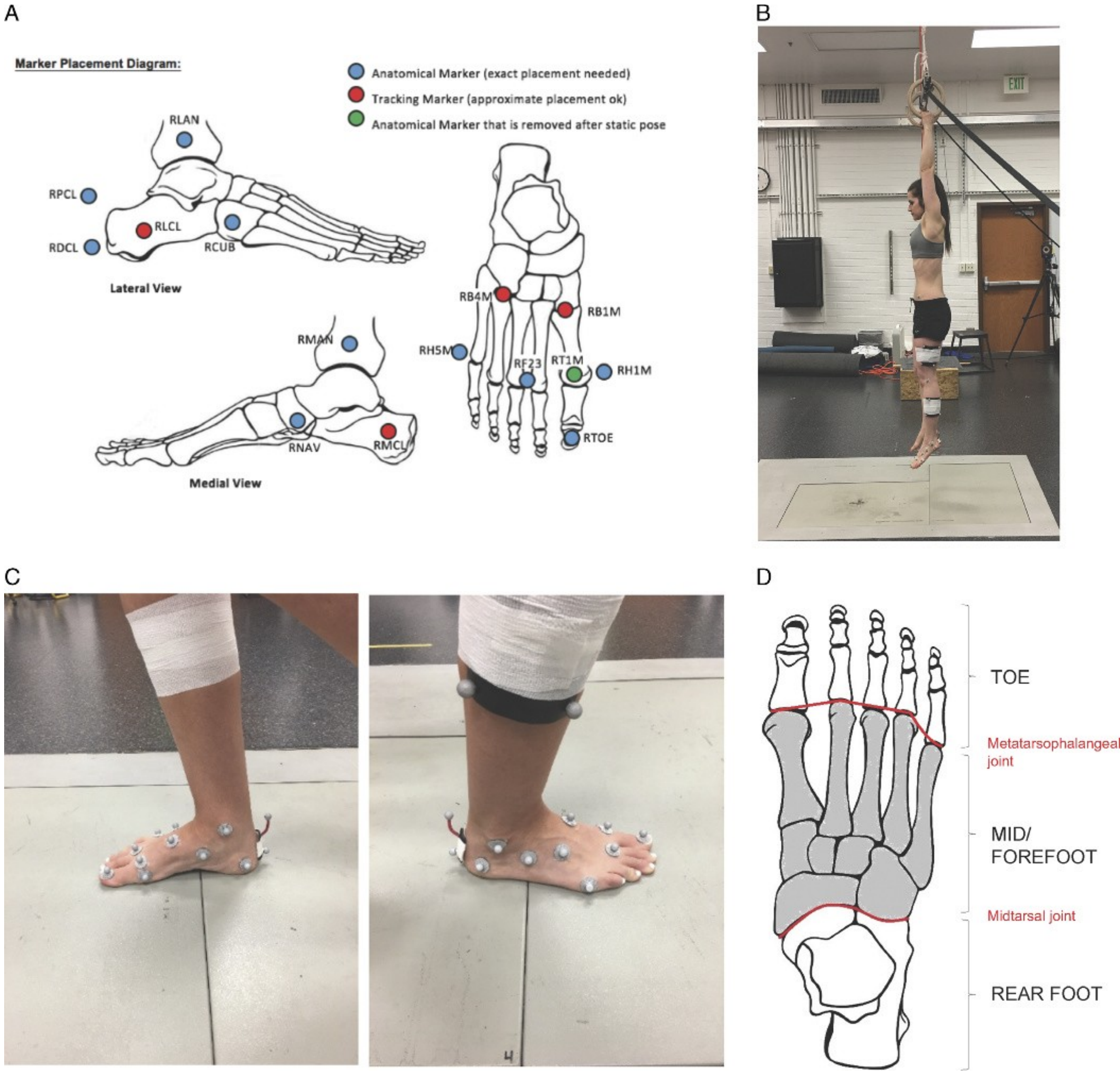

Figure 1—A, Customized kinetic multisegment foot model with slight modifications. B, Hanging apparatus for drop landings. C, Successful landing with the rear foot and forefoot positioned on separate force platforms. D, Foot segments separated by joints of interest.

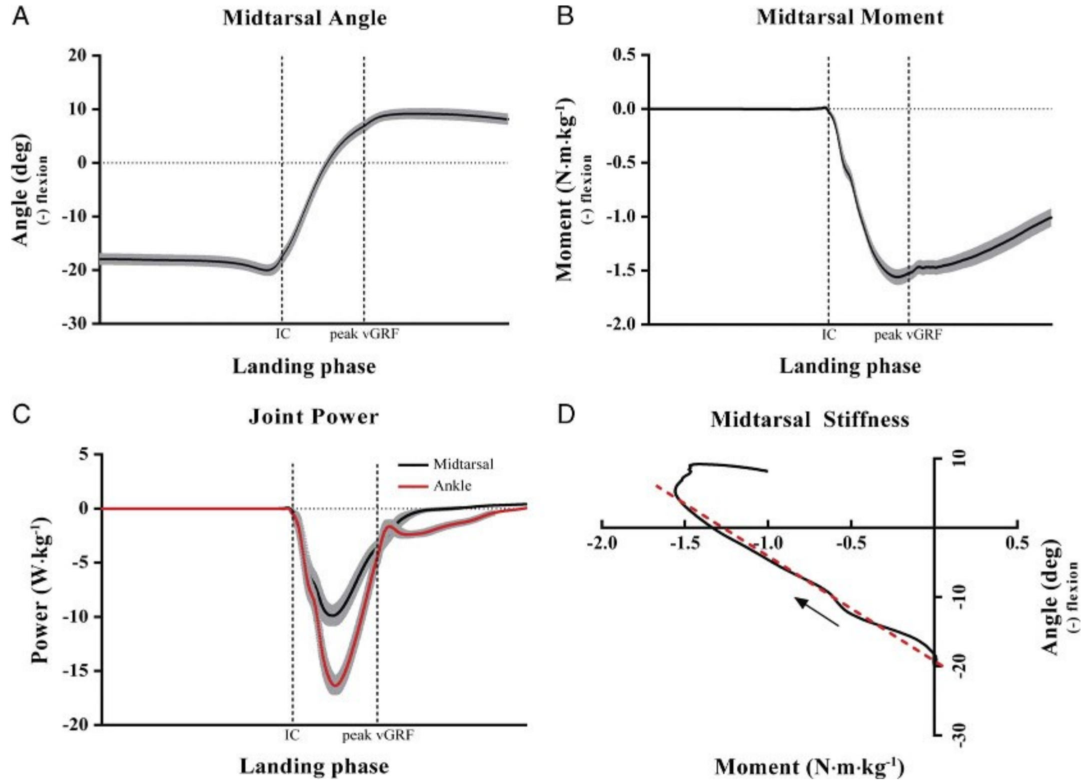

Figure 2—A, Sagittal plane midtarsal joint angles. B, Sagittal plane midtarsal joint moments. C, Midtarsal and ankle joint powers. Mean values represented with a solid black line. Gray bands show SE. D, Midtarsal joint stiffness of a single representative subject throughout the landing phase. The arrow represents the direction of time. All data curves were time normalized for graphical presentation and ease of interpretation.

Results

Sagittal plane midtarsal ROM was large, whereas frontal plane midtarsal joint ROM was relatively small during the landings (Table 1). The midtarsal joint began in flexion and inversion at start-drop and remained in that position until just before IC where it increased in flexion slightly before moving into extension and eversion (Fig. 2A). The midtarsal joint rapidly went through the measured extension excursion beginning with IC, through peak vGRF, and then began to flex and invert again as the subject approached minCOM. Sagittal plane ankle ROM was larger than sagittal plane midtarsal ROM, whereas frontal plane ankle ROM was similar to frontal plane midtarsal ROM (Table 1). Sagittal plane MTP joint ROM was also large through the landings. The MTP joint began in extension and continued extending to almost 20° until just before IC (Fig. 3). 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.

Sagittal plan midtarsal moments peaked at -1.56 ± 0.29 Nm/kg shortly before peak vGRF and slowly began to decline as subjects approached minCOM (Fig. 2B). The sagittal plane ankle moment reached its maximum later, at peak vGRF, with a maximum value of -2.36 ± 0.44 Nm/kg.

Midtarsal power peaked at -9.89 ± 3.69 W/kg between IC and peak vGRF, returning to baseline before minCOM (Fig. 2C). Ankle power peaked at -15.49 ± 5.81 W/kg between IC and peak vGRF similar to but slightly after midtarsal power reached its maximum.

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 4. 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) to decelerate the body during landing followed by the hip extensors (-1.19 ± 0.43 J/kg) and ankle plantar flexors (-1.19 ± 0.26 J/kg) sharing equally the remainder of the work. Multisegment modeling approximated the ankle work at -0.75 ± 0.24 J/kg and midtarsal work at -0.45 ± 0.13 J/kg during landing. A single segment foot model overestimated the ankle's contribution to power absorption on average by 0.44 ± 0.25 J/kg when compared with our multisegment foot model. The variation within subjects was large with a range of ankle power absorption being overestimated by 18%–68%.

A significant and moderately strong inverse correlation was found between standing AHI and static MA ($r = -0.609$, $P < 0.001$) (Table 2). Sagittal plane midtarsal ROM was negatively correlated with standing AHI and positively correlated with static MA. Sagittal plane midtarsal work was correlated positively with standing AHI and negatively with static MA. Static MA was positively correlated with ankle work (Table 2).

Table 1 Raw data with means ± SD

(n = 48)	Mean ± SD
Standing AHI	0.32 ± 0.02
Static MA (deg)	−22.48 ± 5.23
AHI Stiffness	16.85 ± 6.11
Midtarsal Stiffness ($N \cdot m \cdot kg^{-1} \cdot deg^{-1}$)	0.07 ± 0.05
Midtarsal Sagittal ROM (deg)	27.04 ± 6.92
Midtarsal Frontal ROM (deg)	5.81 ± 2.38
Midtarsal Work ($J \cdot kg^{-1}$)	−0.42 ± 0.13
Ankle Sagittal ROM (deg)	34.50 ± 7.47
Ankle Frontal ROM (deg)	12.78 ± 3.33
Ankle Work ($J \cdot kg^{-1}$)	−0.75 ± 0.03

Mean and SD values of static variables (using AHIMS and motion capture technology) and dynamic variables (using motion capture technology) for statistical comparisons.

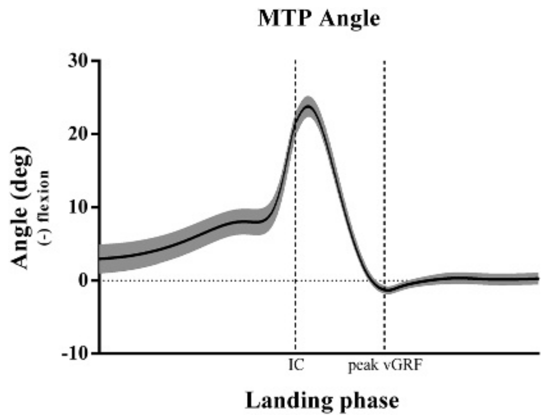

Figure 3—Sagittal plane MTP joint angles. Mean represented with a solid black line. Gray bands show SE. The data curve was time normalized for graphical representation and ease of interpretation.

Discussion

The main purpose of this study was to investigate 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. Although several studies have examined midfoot motion and function during walking and running (8,10,11), to our knowledge, this is the first study to specifically explore midtarsal joint kinematics and kinetics during a barefoot single-leg landing.

Our results suggest that there is a substantial amount of motion in the midtarsal joint in females during a barefoot single-leg landing. Sagittal plane midtarsal joint ROM averaged 27° throughout the landing phase. This is higher than the 5°–20° of sagittal plane motion previously measured in the midfoot during walking (11,27,30,31), likely due to lower vGRFs and midfoot moments measured during gait. At start-drop, subjects began with the ankle in plantarflexion, the midtarsal joint flexed, and the toes slightly extended. Midtarsal joint motion continued through peak vGRF, where it became increasingly mobile and decreased extension velocity assisted by the intrinsic and extrinsic muscles of the lower leg and foot (32,33). The midtarsal joint did not reach peak extension until after peak vGRF, potentially signifying continued absorptive efforts as the subjects approached minCOM. This style of landing was not controlled but adopted naturally by subjects.

The observed increase in midtarsal joint flexion before IC is likely due to activation of both intrinsic and extrinsic foot muscles. Several intrinsic muscles with their origins at the calcaneus and insertions at the toes influence arch flexion. For instance, previous research on forefoot strike running (which involves similar landing mechanics) has shown activation of the abductor hallucis before foot contact (34). Extrinsic muscles may also contribute slightly to arch flexion, even considering their relatively smaller moment arms. For example, during subtalar inversion, co-contraction of the tibialis anterior (35) and posterior would stabilize sagittal plane ankle motion and potentially result in arch flexion (33,36). In addition, the toe extensor muscles (both intrinsic and extrinsic) may contribute to arch flexion through the windlass mechanism (see below). These muscle activations before landing may prepare the foot for impact 1) by sensitizing neuroreceptors as a potential feedforward mechanism to prepare the eccentric control of the midtarsal joint upon impact, 2) by allowing for a greater ROM during landing to assist with energy absorption, and/or 3) by allowing for earlier and greater contact time to respond to proprioceptive feedback.

Toe extension before and during landing may have contributed to midfoot function. Before landing, the MTP joint extended close to 20°, increasing contact area as well as engaging the windlass mechanism and causing the plantar fascia to tighten, pulling the midtarsal joint into further flexion (28). 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 (10). Loading of the foot during landing is sequentially reversed (from toe to heel, rather than heel to toe in walking); therefore, the windlass mechanism, initiated by active extension of the toes, may assist in power absorption at the midtarsal joint. Without sufficient muscular control of the midtarsal joint, the absorption of landing forces is left to the plantar fascia and ligaments (37). With excessive repetition and/or overloading, stress injury to the plantar fascia may occur. However, at manageable loads, loading of the foot during toe to heel landings may induce eccentric strengthening and lead to hypertrophy of the intrinsic and extrinsic foot muscles. During dynamic activities, intrinsic and extrinsic foot muscles support the MLA, with the abductor hallucis and posterior tibialis muscles being major contributors (38,39). To control medial–lateral forces and achieve balance, coordinated contractions of the antagonistic muscles (fibular muscles, tibialis anterior, and tibialis posterior) must occur. Intrinsic foot muscles, such as the abductor hallucis, flexor digitorum brevis, and quadratus plantae, have the capacity to slow deformation of the MLA, control midtarsal joint motions, and protect the foot from excessive strain on tissues such as the plantar fascia (33). Stronger muscles may have protective characteristics for the foot, which may reduce injury rates, particularly in females who have greater midfoot motion in walking (20).

Our results show that the midtarsal joint contribution to negative work continued past peak vGRF until minCOM, signifying active shock absorption throughout the impact phase of landing. Sagittal plane midtarsal joint moments showed a maximum joint torque occurring just before peak vGRF, possibly the result of active midfoot flexors. Midtarsal joint powers were negative, suggesting that midfoot flexors were eccentrically contracting to slow down midtarsal joint extension and absorb energy from the landing.

Impact absorption during landing can be quantified by calculating work at the major joints of the lower extremity (i.e., ankle, knee, and hip) (40). However, it is clear that work is also being performed by the structures of the foot, although this has been more difficult to measure. Our results showed that the 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 an average of 11% ± 3.5% (range: 5%–22%) of the total work performed by all lower extremity joints. These findings support previous research, which suggests that the neurophysiological and mechanical function of the midfoot play a larger role than previously thought (41). The amount of power absorption measured at the midtarsal joint seems substantial enough to not be overlooked and may also be clinically meaningful when prescribing injury prevention and rehabilitation interventions.

From our single segment model approach, the ankle and hip contributed equally to total negative work performed. Our analysis showed that this overestimated ankle work by an average of 38%, similar to previous research on walking and running (10,28,42). Ankle power differences between models were likely due almost entirely to the differences in angular motion and velocity of the rear foot segment, as the inertial properties of the foot are small (10). Thus, the combined midtarsal and ankle moments are roughly equal to a single segment ankle moment.

These results highlight the importance of using a multisegment foot model to more completely and accurately understand lower extremity mechanics. The overestimation of ankle work and power when calculated using a single rigid foot model may result in assumptions of greater activity of the larger ankle plantarflexors and underestimate the role of the smaller intrinsic and extrinsic foot muscles. In addition, the use of a multisegment foot model allows researchers and clinicians to differentiate the timing of power absorption and generation across the joints of the foot (43).

The secondary aim of this study was to identify a potential relationship between static foot posture and dynamic midfoot function during the landing task. We chose to compare two different types of static measures—standing AHI using the AHIMS device and static MA from motion capture technology. The moderately strong correlation between the two different measures of static foot posture suggests that either method may be appropriate for use in clinical and research settings. However, static MA showed a stronger correlation to sagittal plane midtarsal joint ROM than standing AHI. Studies that compare static–dynamic relationships generally use measurements acquired via different techniques, such as comparing AHI from the AHIMS device to motion capture based joint angles. Calculating the midtarsal joint angle the same way under both static and dynamic conditions could lead to increased accuracy for static-dynamic comparisons.

Clinicians routinely rely on assumed static–dynamic relationship to prescribe interventions to improve foot function in high and low arched individuals (13,14). Although static arch measurements are generally thought to be indicative of passive motion assessed by clinicians (19), there is little evidence of a relationship when the movement is active (14–17). Studies that have found correlations between static and dynamic measurements focused on postural measures (peak eversion angle or navicular height at mid-stance) (44,45) rather than functional measures (ROM or joint work) (14–17). Examining these functional measures under higher loading conditions suggests that there is a significant but relatively weak relationship between static foot posture and sagittal plane midtarsal joint ROM and work in a normal arched population. Applying similar methodology (high forces and kinetic multisegment foot model) to a more diverse subject sample that includes high and low arched individuals may reveal a stronger static–dynamic relationship that could be used in a clinical setting (46). For example, a simplified motion capture setup using a few markers on the foot and a high speed camera could allow clinicians to make simple assessments of midfoot mobility during dynamic movements. Further research is needed to determine simple, cost-effective solutions for clinicians to use to assess dynamic foot function.

In the current study, neither passive nor active measures of stiffness showed significant correlations to any other measured variable. Passive stiffness (difference between AHI sitting and AHI standing) is dependent on passive structures and is assessed at low loads (body weight), which may not be sufficient to elicit a strong functional response in a normal arched population. The lack of correlations using active midtarsal joint stiffness was unexpected as this measure includes both loading and ROM. One possible explanation is the variability due to the complex interplay between active and passive tissues and their responses to loading. Although one person may rely primarily on passive structures (ligaments, bone structure, and fascia), another may rely more on muscular contractions to control the arch. Whether a stronger response in either stiffness measure would occur in a more flexible arch or overpronating population is worthy of investigation.

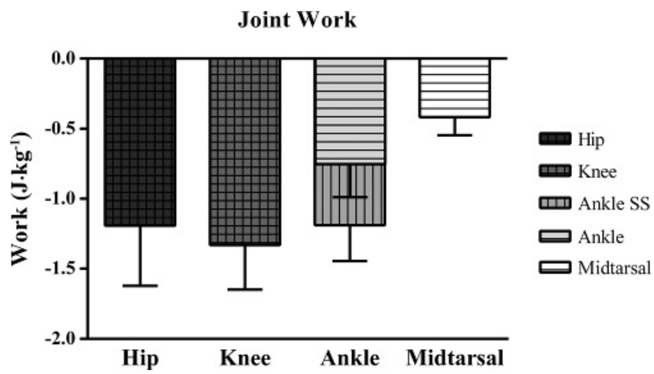

Figure 4—Joint work mean and SD (error bars) for both single segment (vertical lines) and multisegment (horizontal lines) foot models.

Table 2. Correlation coefficients and (*P* values) for static–dynamic variables.

(n = 48)	Standing AHI	Static MA (deg)	AHI Stiffness
Midtarsal Stiffness ($N{\cdot}m{\cdot}kg^{-1}{\cdot}deg^{-1}$)	0.068 (0.645)	−0.167 (0.255)	0.110 (0.456)
Midtarsal sagittal ROM (deg)	−0.320 (0.026)*	0.483 (<0.001)*	0.047 (0.752)
Midtarsal frontal ROM (deg)	−0.089 (0.548)	0.123 (0.403)	−0.076 (0.609)
Midtarsal work ($J{\cdot}kg^{-1}$)	0.332 (0.021)*	−0.323 (0.025)*	−0.103 (0.486)
Ankle sagittal ROM (deg)	−0.168 (0.253)	0.056 (0.703)	0.029 (0.845)
Ankle frontal ROM (deg)	0.036 (0.808)	−0.272 (0.061)	−0.159 (0.845)
Ankle work ($J{\cdot}kg^{-1}$)	−0.252 (0.081)	0.498 (<0.001)*	−0.017 (0.911)

*Indicates a significant relationship between static and dynamic variables at $p<0.05$.

Limitations

It should be noted that 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. In addition, our study participants may have experienced ankle/foot injuries more than 6 months before participation in the study. Therefore, it is possible that some subjects may have residual alterations in midfoot mechanics. We did not control for landing strategy or allow subjects familiarization trials during data collection. However, we did collect multiple landing attempts until three successful trials were completed. The use of a multisegment foot model to measure midtarsal joint ROM, although an improvement from traditional models, is still limited in identifying individual tarsal articulations. Although our split force platform approach may not be practical for in-field studies, more accurate calculations of midfoot kinematics and kinetics using a multisegment foot model are important as differences in landing mechanics may be undetectable with a single segment foot model (47).

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 substantially to power absorption during barefoot single-leg landings. In addition, ankle power absorption is greatly overestimated when using a single segment 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 only mildly correlated with midtarsal joint kinematics and kinetics during a landing task. Static foot posture may have utility, but appears to be a weak tool in assessing midfoot function and injury risk in normal arched populations.

The authors declare no conflict of interest. The results of the present study do not constitute endorsement by the American College of Sports Medicine and are presented clearly, honestly, and without fabrication, falsification, or inappropriate data manipulation.

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