



Article

The Relationship Between Foot Posture, Dorsiflexion Range of Motion and Lower Extremity Biomechanics During a Drop-Landing Task

Kendra S. Graham  and Joshua T. Weinhandl * 

Department of Kinesiology, Recreation, and Sport Studies, College of Health and Education, Health and Human Sciences, The University of Tennessee, 1914 Andy Holt Ave, Knoxville, TN 37996, USA; kgraha29@vols.utk.edu

* Correspondence: jweinhan@utk.edu

Abstract

Background/Objectives: While restricted dorsiflexion range of motion (DF-ROM) is linked to deleterious sagittal and frontal plane knee and hip kinematics during landing, the literature is conflicted as to whether excessive foot pronation is linked to knee injury. The purpose of this study was to examine the relationship between static foot posture, DF-ROM, and lower extremity biomechanics during a drop-landing task. **Methods:** Fifteen physically active adults (age: 22.6 ± 2.4 years, height: 1.69 ± 0.08 m, mass: 66.40 ± 9.95 kg) volunteered to participate in this study. Static foot posture was measured by the six criteria of the Foot Posture Index (FPI-6) and DF-ROM was measured using the weight-bearing lunge test (WB-LT). Sagittal and frontal plane kinematics and kinetics of the hip, knee, and ankle were captured using a 3D motion capture system and force plate during a drop-landing task. **Results:** FPI-6 scores (4.67 ± 2.94) correlated with knee abduction angle at initial contact ($1.08 \pm 3.30^\circ$, $r = -0.59$, $p = 0.02$), ankle sagittal plane excursion ($39.11 \pm 7.67^\circ$, $r = -0.63$, $p = 0.01$) and knee adduction moment (0.58 ± 0.51 N/kg, $r = 0.60$, $p = 0.017$). DF-ROM correlated with knee adduction moment ($r = -0.59$, $p = 0.02$). The combination of FPI-6 and DF-ROM accounted for 56% of the variance in knee adduction moment ($r = 0.746$, $p = 0.008$). No significant relationships were identified for hip variables ($p > 0.05$). **Conclusions:** Participants with a more pronated static foot posture displayed less knee adduction angle at initial contact and decreased ankle sagittal plane excursion. Those with less DF-ROM and a pronated static foot posture exhibited increased maximum knee adduction moment. Foot and ankle structure influence lower extremity biomechanics.

Keywords: dorsiflexion; foot pronation; weight-bearing lunge; foot posture index; kinematics; kinetics



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1. Introduction

Knee injuries can have significant ramifications to acute health outcomes, in addition to costly surgeries, rehabilitation, and time loss from activity [1]. Furthermore, they may result in long-term consequences such as a decreased health-related quality of life and increased risk for post-traumatic osteoarthritis development [2]. Anterior cruciate ligament (ACL) injuries are especially prevalent in the United States, accounting for about half of all knee injuries and affecting 1 out of every 3500 people [3]. Further investigation into the possible mechanisms for knee injuries is warranted.

Biomechanical analysis of the knee and hip during dynamic activities has been extensively performed to identify movement patterns that may contribute to knee injuries

broadly, and ACL injuries specifically. Previous studies have explored landing tasks' effects on knee biomechanics due to the increased loading demands on joint structures during landing, and these demands' subsequent association with both chronic and acute knee injuries. For example, during landing, larger peak externally applied knee abduction moments, initial and peak knee abduction angle, and decreased sagittal plane knee and hip range of motion have been associated with ACL injury [4–7]. Further exploration of the associated factors with these risky movement patterns will allow for the development of improved knee injury prevention and treatment strategies.

During landing tasks, restrictions in ankle dorsiflexion range of motion (DF-ROM) are considered to have an association with injurious lower extremity biomechanics. A recent systematic review by Lima et al. determined that restricted DF-ROM was associated with increased knee valgus and frontal plane excursion, reduced knee and hip sagittal excursion, and increased peak vertical ground reaction force (vGRF) during landing [8]. Similarly, a meta-analysis by Mason-Mackay et al., assessing DF-ROM in a weight-bearing position, found restricted DF-ROM to be associated with dynamic knee valgus during landing [9]. These syntheses posit an elevated risk of knee injury due to restrictions in DF-ROM, finding clear associations with restrictions in DF-ROM and risky frontal and sagittal plane knee biomechanics during landing. Therefore, weight-bearing DF-ROM measures may be useful for identifying individuals with landing biomechanics that may result in future knee injury.

While restricted DF-ROM is linked to deleterious sagittal and frontal plane knee and hip kinematics during landing, the literature is conflicted as to whether excessive foot pronation is a risk factor of knee injury [10,11]. The relationship between static foot posture and ACL injury has been studied retrospectively using the navicular drop test as the primary assessment of foot posture [12–14]. While several studies have found an association between pronated static foot postures and ACL injury history, an equal number of studies have reported contradictory findings. A recent systematic review by Neal et al., 2014, concluded that static measurements of the medial longitudinal arch alone are not sufficient indicators of future knee injury [10]. However, measurements such as the Foot Posture Index (FPI-6), a multifactorial assessment of static foot posture, may provide more insight into a potential association with lower extremity injury. Few studies have examined the relationship between foot posture and lower extremity kinematics and kinetics during landing. However, individuals with a pronated static foot posture have demonstrated greater vGRF when landing from 40 and 60 cm drop heights compared to individuals with a neutral foot posture [15]. Additional investigation is necessary to understand foot posture's role in injurious lower extremity biomechanics.

Examining DF-ROM, foot posture, and the potential interaction of these elements will provide information about lower extremity movement patterns that may contribute to musculoskeletal injuries, specifically knee injuries, such as ACL injury. Exploring these factors and their interactions will assist in bettering knee injury prevention and rehabilitation strategies. Therefore, the purpose of this study was to investigate the relationship between static foot posture, DF-ROM, and drop-landing biomechanics in physically active adults. We hypothesized that static foot posture and DF-ROM would influence landing kinetics and kinematics at the knee, ankle, and hip.

2. Materials and Methods

Fifteen physically active individuals were recruited to participate in this study. All participants provided written informed consent in compliance with the University's institutional review board. Participants were included if they were between the ages of 18–35 and participated in greater than or equal to 90 min of recreational physical activity or sports per week that at some point included both jumping and landing movements. Individuals who

reported a lower extremity injury in the past 6 months, a history of lower extremity surgery, or a health condition that may have affected their landing ability were excluded from the study. Upon inclusion into the study, all participants came to the lab for a single testing session. A single investigator conducted all assessments to maintain inter-rater reliability. All measures were taken on the dominant limb, determined by asking participants their preferred limb for kicking a ball [16].

Static foot posture was measured using the FPI-6, which has demonstrated excellent intra-rater reliability in research by Cornwall et al. (ICC = 0.90) [17]. The FPI-6 was utilized due to its multi-planar approach to measuring foot posture. A multi-planar approach may allow for a more accurate depiction of foot biomechanics relationship with knee biomechanics. To complete the FPI-6, participants stood in a comfortable stance with their feet shoulder-width apart while the investigator evaluated the foot utilizing six criteria. The six criteria include: talar head palpation, supra and infra lateral malleolar curvature, calcaneal frontal plane position, prominence in the region of the talonavicular joint, congruence of the medial longitudinal arch, and abduction/adduction of the forefoot on the rearfoot [17,18]. All six criteria were graded individually on a 5-point Likert-type scale ranging from −2 to 2. All individual scores were summed for a total score between −12 and 12. Higher positive scores indicate a more pronated static foot posture whereas higher negative scores are indicative of a more supinated foot posture.

The WB-LT was used to assess maximal weight-bearing DF-ROM. The WB-LT has demonstrated excellent inter-rater and intra-rater reliability (ICC > 0.90) [19]. During this evaluation, the participant performed a modified knee-to-wall lunge without shoes on [20]. The participant was allowed to place their hands on the wall for support and their non-test limb in a comfortable position behind the tested limb. The big toe of the tested limb was placed in the middle of a tape measure positioned perpendicular to the wall. All participants began in the testing position with their big toe one centimeter from the wall. The participants were instructed to lunge forward until their knee contacted the wall while simultaneously maintaining heel contact with the ground. Foot placement was progressed backwards along the tape measure until the participant could no longer perform the lunge without their heel lifting from the floor or their knee not contacting the wall. Maximal dorsiflexion was measured as the furthest point along the tape measure in which the participant's knee could make contact with the wall and that their heel remained in contact with the floor. Participants performed three practice trials before the three test trials. Test trials total scores were averaged and used for analysis.

Sagittal and frontal plane kinematics and kinetics of the hip, knee, and ankle were examined during a single leg drop-landing task. Kinematics were captured through an 8-camera 3D motion analysis system (Vicon Motion Systems, Denver, CO, USA) at a sampling rate of 200 Hz. To reduce motion artifact and variability during motion assessment, attire was standardized. Females wore spandex shorts with a sports bra and low-cut socks; males were shirtless with spandex shorts and low-cut socks. All participants wore the same model Nike sneakers to complete the drop-landing. Single reflective markers and cluster plates were placed bilaterally on anatomical landmarks on the torso and lower extremity by a single investigator. Single reflective markers were placed on the acromioclavicular joint, anterior superior iliac spine, posterior superior iliac spine, iliac crest, greater trochanter, lateral and medial femoral condyles, lateral and medial malleoli, base of the fifth metatarsal, and base of the first metatarsalphalangeal joint. Cluster plates were attached by Velcro at the heel of the sneaker and on the lower leg, thigh, and mid thoracic region on the back using neoprene wraps. Figure 1 illustrates the marker set used. Once the markers were attached, participants were asked to stand on the force plate with their hands raised above their head for calibration. Following calibration, all markers were removed except the

anterior superior iliac spine, posterior superior iliac spine, and cluster plates before the drop-landing task was performed. Kinetic data were captured synchronously using a force plate (Bertec Corp, Columbus, OH, USA) at a sampling rate of 1000 Hz.

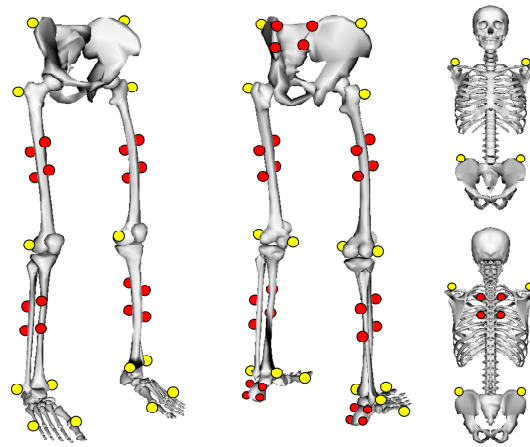


Figure 1. Markerset utilized for data collection. Red markers represent cluster plates, while yellow markers represent the anatomical tracking markers, which were removed prior to data collection.

The box height for the drop-landing task was set to each respective participant's maximum vertical jump height. To determine this, participants performed three maximal effort countermovement jumps on the force plate while ground reaction forces were recorded with a custom software (LabVIEW, v11.0, National Instruments Corporation, Austin, TX, USA). Jump height was calculated for each countermovement jump. This was done using the impulse-momentum relationship, and the box height was set to the maximum of the three trials [21]. Participants then performed 3–5 practice trials of the drop-landing after proper landing instructions. While standing on their non-dominant limb, participants dropped off the box onto the force plate and performed a single-legged landing on their dominant limb. Successful trials occurred when the participants landed on the force plate properly with their whole foot, placed and maintained their hands on their hips for the entire trial, and did not contact the ground with their non-test limb or propel themselves forward off the force plate after landing. Testing concluded once three successful trials were completed.

Drop-landing data was analyzed using Vicon Nexus software (v2.15.0, Vicon Motion Systems, Denver, CO, USA) and Visual3D (v2025 11.1, C-Motion, Inc., Rockville, MD, USA). Calculations of kinematic and kinetic data from the markers and force plate data were performed using Visual3D. Joint moments are reported as internal moments, representing the torques produced as the internal structures, muscles, resist external loads. The right-hand rule convention was used for all joint angles and moments. Sagittal and frontal plane kinematics for the ankle, knee, and hip were reported at initial contact and total excursion range of motion ($^{\circ}$). Maximal joint moments were normalized to body mass (N/kg).

Raw 3D marker coordinate and ground reaction force data were filtered using a fourth-order, low-pass Butterworth filter with a cutoff frequency of 20 Hz [22]. A kinematic model comprising four skeletal segments (pelvis, right thigh, right shank, and right foot) was created using a static trial in Visual3D (v2025 11.1, C-Motion Inc., Rockville, MD, USA). Three-dimensional ankle, knee, and hip angles were calculated using a joint coordinate system [23]. Hip joint centers were located according to Weinhandl and O'Connor [24]. Knee joint center was defined as the midpoint between epicondyle markers according to Grood and Suntay [23] and ankle joint center was defined as the midpoint between malleoli markers [25]. Internally applied, 3D joint kinetics were calculated using a Newton–Euler approach according to Bresler and Frankel [26] and projected to the joint coordinate

system [27]. All kinetic data were normalized to participant body mass. Body segment masses were estimated from Dempster and Gabel [28] and segment moment of inertias were estimated from Hanavan [29]. Positive joint angles and joint moments represented hip flexion and adduction, knee extension and adduction, and ankle dorsiflexion and inversion.

Pearson correlations (r) along with the coefficient of determination (r^2) were performed to examine the relationship between the FPI-6, DF-ROM, and individual landing kinematic and kinetic variables (Tables 1 and 2). In cases where both FPI-6 and DF-ROM significantly correlated to a biomechanical variable, backwards multivariate linear regression was used to model the relationship (Table 3). Pearson correlations were interpreted as weak (0.10–0.40), moderate (0.41–0.69), or strong (>0.70) [21]. Significance was set at $p \leq 0.05$ for all analyses.

Table 1. Means and standard deviation (SD) for all dependent variables.

	Mean	SD
FPI Score	4.67	2.94
WB-LT (cm)	11.84	3.29
Knee abduction angle at IC (°)	1.08	3.29
Hip flexion angle at IC (°)	20.69	7.27
Hip adduction angle at IC (°)	−10.58	5.20
Knee extension angle at IC (°)	−13.87	4.70
Ankle dorsiflexion angle at IC (°)	−19.43	6.64
Hip flexion total excursion (°)	21.45	8.95
Hip adduction total excursion (°)	10.17	4.27
Knee extension total excursion (°)	−45.87	9.15
Knee abduction total excursion (°)	1.84	4.18
Ankle dorsiflexion total excursion (°)	39.11	7.68
Ankle inversion total excursion (°)	−14.91	6.75
Hip extension moment (N/kg)	−9.01	2.40
Hip abduction moment (N/kg)	−4.48	1.00
Knee extension moment (N/kg)	5.29	0.80
Knee adduction moment (N/kg)	0.58	0.51
Ankle plantarflexion moment (N/kg)	−3.85	0.82
Ankle eversion moment (N/kg)	−0.46	0.50

FPI = Foot Posture Index; WB-LT = weight-bearing lunge test; and IC = initial contact.

Table 2. Pearson correlation coefficients (r) between FPI, WB-LT, and all landing kinematics (°).

	FPI	WB-LT
Knee abduction angle at IC	−0.59 *	0.19
Hip flexion angle at IC	0.01	0.12
Hip adduction angle at IC	0.13	0.071
Knee extension angle at IC	0.11	−0.08
Ankle dorsiflexion angle at IC	0.32	−0.40
Hip flexion total excursion	−0.26	0.07
Hip adduction total excursion	0.17	0.28
Knee extension total excursion	0.30	−0.17
Knee abduction total excursion	−0.23	−0.01
Ankle dorsiflexion total excursion	−0.63 *	0.468
Ankle inversion total excursion	0.05	−0.001

FPI = Foot Posture Index, WB-LT = weight-bearing Lunge test, and IC = initial contact. * Denotes significant Pearson correlation between variables ($p < 0.05$).

Table 3. Pearson correlation coefficients (*r*) between FPI, WB-LT, and hip, knee, and ankle landing kinetics (N/kg).

	FPI	WB-LT
Hip extension moment	0.06	−0.12
Hip abduction moment	−0.43	0.20
Knee extension moment	−0.06	−0.18
Knee adduction moment	0.60 *	−0.59 *
Ankle plantarflexion moment	0.06	−0.15
Ankle eversion moment	−0.06	0.09

FPI = Foot Posture Index, WB-LT = weight-bearing lunge test. * Denotes significant Pearson correlation between variables ($p < 0.05$).

3. Results

Participant characteristics and average measured values are summarized in Table 4. FPI-6 scores displayed a significant negative correlation with knee abduction angle at initial contact ($r^2 = 0.35$, $p = 0.021$) and ankle sagittal plane excursion ($r^2 = 0.39$, $p = 0.013$), as well as a significant positive correlation with knee adduction moment ($r^2 = 0.36$, $p = 0.017$) (Table 2). DF-ROM was significantly correlated with knee adduction moment ($r^2 = 0.35$, $p = 0.020$). The combination of FPI-6 and DF-ROM accounted for 56% of the variance in knee adduction moment ($r = 0.746$, $p = 0.008$). This relationship indicates that those with a more pronated static foot posture and decreased DF-ROM had an increased adduction moment at the knee. No statistically significant relationships were identified between the FPI-6, DF-ROM, and kinematic or kinetic hip variables ($p > 0.05$).

Table 4. Participant characteristics.

Sample Size	15
Age (yrs)	22.1 ± 2.3
Height (m)	1.70 ± 0.07
Mass (kg)	66.7 ± 9.8
Sex	Male: 5 (33.3%), Female: 10 (66.7%)
Dominant Leg	Right: 14, Left: 1
Jump Height (m)	0.32 ± 0.12

4. Discussion

This study identified moderate relationships between DF-ROM, static foot posture, and landing biomechanics. Specifically, less DF-ROM and more pronated static foot postures were independently related to larger knee adduction moments, explaining 56% of the variance in knee adduction moments. Therefore, participants with any combination of these factors (restricted DF-ROM, more pronated static foot posture) may be experiencing greater loading forces at the knee joint.

Potentially, participants who demonstrated a more pronated static foot posture experienced lesser dynamic foot pronation during landing, limiting force attenuation capacities of the foot. These findings align with previous research suggesting restricted DF-ROM may alter landing mechanics. Compensatory landing mechanics may include increased midfoot pronation and/or increases in knee valgus, both of which are associated with knee injury [9]. Lesser DF-ROM was associated with larger knee adduction moments in the single leg drop landing task performed in this study, further supporting findings from a recent systematic review that concluded restrictions in DF-ROM may alter lower extremity landing mechanics, creating a predisposition for knee injury [9]. Additionally, previous biomechanics research on walking [30,31], squatting [32,33] and stepping [34] have also shown associations between restricted DF-ROM and altered movement mechanics. Due

to the widely observed correlations between restricted DF-ROM and movement alterations, the evaluation of weight-bearing DF-ROM should be performed when discerning mechanisms for potential knee injury.

Ankle dorsiflexion assists in force attenuation during landing and other dynamic tasks. This study identified that deficits in DF-ROM may result in compensations associated with increased valgus load on the knee, possibly increasing requisite force attenuation by musculotendinous and ligamentous structures of the joint. Although deficits in DF-ROM were associated with larger knee adduction moments, no significant relationships were identified with lower extremity kinematics. Participants may have adopted unique movement strategies to compensate for deficits in DF-ROM.

Investigators posit the lack of alteration to hip kinematics may be related to its distance from the ankle joint in the kinetic chain and to the study's small and recreationally active sample size. During closed-chain activities, compensations and movements at one joint can affect the movement and loading at other points in the kinetic chain [35]. Knee mechanic alterations can, at least in part, be attributed to foot strength and ankle range of motion [8,36,37]. Varying degrees of DF-ROM restriction may exhibit differing patterns of movement compensation at each joint. Individual differences in adopted movement strategies should be further explored to fully understand the compensatory phenomena during landing.

Previous studies that examined the relationship between DF-ROM and landing mechanics used a standardized box height ranging from 30 to 72 cm [8]. Box height for the drop landing task in this study was based on each participant's maximal vertical jump height. This approach normalizes the task to the individual's ability, theoretically making the task equally challenging for all participants. Previous research utilizing a box height normalized to the maximal vertical jump height of participants or greater-than-maximal vertical jump height exhibited stronger relationships between DF-ROM and frontal plane biomechanics [32,38–40]. This indicates that drop-landing heights normalized to, or higher than, one's maximal vertical jump height require greater levels of force attenuation from lower extremity structures. Thus, one's DF-ROM has greater relevance to landing biomechanics.

Foot posture also may play a role in altered biomechanics during landing. Static foot posture demonstrated relationships with knee abduction angle at initial contact, ankle sagittal plane excursion and knee adduction moment. Static foot posture alone accounted for 37% of the variance in knee adduction moments. The exact mechanism by which static foot posture alters landing mechanics is unclear. Potentially, participants with a greater pronated static foot posture moved through less dynamic foot pronation during landing, limiting the amount and rate of force attenuation that can be accomplished by the foot. The results of this study indicate that static foot pronation may have a relation to valgus collapse of the knee during a single leg drop landing task. Further investigation is warranted as valgus collapse of the knee during drop landing is associated with knee injury, namely to that of the ACL [41].

Findings related to foot posture and DF-ROM provide evidence that the structures of the foot and ankle have a significant influence on lower extremity biomechanics. In the present study, DF-ROM and static foot posture were not significantly correlated. This suggests that there is a unique relationship between the two variables and knee adduction moment during landing. Lesser DF-ROM and a pronated static foot posture are associated with an elevated maximum knee adduction moment during drop landing, indicating an increase in valgus knee loading, a widely accepted mechanism for knee injury. This association rationalizes the further exploration of the role DF-ROM and foot posture play

in knee injuries, and the inclusion of their assessment, at the bare minimum, during rehabilitation and injury prevention interventions.

There are multiple limitations to consider when interpreting findings from this study. First, this study is cross-sectional and correlational in nature. While the results offer valuable insights into foot posture, DF-ROM and landing mechanics, the study does not assess causation between static foot posture, DF-ROM and injurious movement. Further research is necessary to confirm static pronated static foot posture and restricted DF-ROM's role on deleterious movement strategies during landing. Additionally, this study included a small, recreational group of participants, limiting the generalizability of findings across populations. The small sample size further limits the interpretation and usability of derived correlational relationships; issues such as increased outlier sensitivity and increased likelihood of type I error will be fully prevented by exploring DF-ROM and static foot posture in a larger sample. Furthermore, while footwear was standardized, the familiarity and comfort of standardized shoes may have affected participant landing mechanics. Future research should consider studying landing mechanics without shoes on participants to account for this effect.

The present study suggests DF-ROM and foot pronation are important to evaluate and consider for knee injury prevention and rehabilitation in similar populations and/or for those who participate in landing tasks. Future research into the relationship between these factors and across different populations is essential, as is the development of interventions addressing and preventing limitations in DF-ROM and excessive foot pronation.

To conclude, the structures of the foot and ankle have an influence on drop landing biomechanics. Participants with a more pronated static foot posture displayed a lesser knee adduction angle at initial contact and a decreased ankle sagittal plane excursion compared to participants with a less pronated static foot posture. Participants with lesser DF-ROM and a less pronated static foot posture exhibited an increased maximum knee adduction moment compared to their counterparts, indicating an increase in valgus knee loading. These findings are important to consider as increases in knee valgus loading may put these individuals at greater risk for knee injuries, such as ACL injury. Assessments such as the FPI-6 and WB-LT are suggested for use to identify individuals at risk for possible excessive knee valgus loading to prevent these deleterious injuries.

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Abbreviations

The following abbreviations are used in this manuscript:

ACL	Anterior Cruciate Ligament
DF-ROM	Dorsiflexion Range of Motion
vGRF	Vertical Ground Reaction Force
FPI-6	Foot Posture Index
WB-LT	Weight-Bearing Lunge Test
IC	Initial Contact

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