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Structure, Sex, and Strength and Knee and Hip Kinematics During Landing

Jennifer S. Howard, PhD, ATC*; Melisa A. Fazio, MS, ATC†; Carl G. Mattacola, PhD, ATC, FNATA*; Timothy L. Uhl, PhD, PT, ATC, FNATA*; Cale A. Jacobs, PhD‡

*Department of Rehabilitation Sciences, University of Kentucky, Lexington; †University of Colorado, Boulder; ‡ERMI, Inc, Atlanta, GA

Context: Researchers have observed that medial knee collapse is a mechanism of knee injury. Lower extremity alignment, sex, and strength have been cited as contributing to landing mechanics.

Objective: To determine the relationship among measurements of asymmetry of unilateral hip rotation (AUHR); mobility of the foot, which we described as relative arch deformity (RAD); hip abduction–external rotation strength; sex; and medial collapse of the knee during a single-leg jump landing. We hypothesized that AUHR and RAD would be positively correlated with movements often associated with medial collapse of the knee, including hip adduction and internal rotation excursions and knee abduction and rotation excursions.

Design: Descriptive laboratory study.

Setting: Research laboratory.

Patients or Other Participants: Thirty women and 15 men (age = 21 ± 2 years, height = 171.7 ± 9.5 cm, mass = 68.4 ± 9.5 kg) who had no history of surgery or recent injury and who participated in regular physical activity volunteered.

Intervention(s): Participants performed 3 double-leg forward jumps with a single-leg landing. Three-dimensional kinematic data were sampled at 100 Hz using an electromagnetic tracking system. We evaluated AUHR and RAD on the preferred leg and evaluated isometric peak hip abductor–external rotation torque. We assessed AUHR by calculating the difference between internal and external hip rotation in the prone position (AUHR = internal rotation − external rotation). We evaluated RAD using the Arch Height Index Measurement System. Correlations and linear regression analyses were used to assess relationships among AUHR, RAD, sex, peak hip abduction–external rotation torque, and kinematic variables for 3-dimensional motion of the hip and knee.

Main Outcome Measure(s): The dependent variables were joint angles at contact and joint excursions between contact and peak knee flexion.

Results: We found that AUHR was correlated with hip adduction excursion ($R = 0.36$, $P = .02$). Asymmetry of unilateral hip rotation, sex, and peak hip abduction–external rotation torque were predictive of knee abduction excursion (adjusted $R^2 = 0.47$, $P < .001$). Asymmetry of unilateral hip rotation and sex were predictive of knee external rotation excursion (adjusted $R^2 = 0.23$, $P = .001$). The RAD was correlated with hip adduction at contact ($R^2 = 0.10$, $R = 0.32$, $P = .04$) and knee flexion excursion ($R^2 = 0.11$, $R = -0.34$, $P = .03$).

Conclusions: Asymmetry of unilateral hip rotation, sex, and hip strength were associated with kinematic components of medial knee collapse.

Key Words: anteversion, valgus, arch mobility, alignment, anterior cruciate ligament, biomechanics, lower extremity, risk factor, regression analysis

Key Points

- Asymmetry of unilateral hip rotation, sex, and hip abduction–external rotation strength were predictive factors for knee abduction excursion during landing.
- Asymmetry of unilateral hip rotation and sex were predictive of knee external rotation excursion during landing.
- Asymmetry of unilateral hip rotation alone was predictive of hip and knee adduction excursions during landing.
- Relative arch deformity was correlated with hip adduction at contact and knee flexion excursion but did not predict medial knee collapse during landing.
- Greater asymmetry of unilateral hip rotation, female sex, and lower hip abduction–external rotation strength might be associated with a medial collapse during landing characterized by greater hip adduction, knee abduction, and knee external rotation.

Medial collapse of the lower extremity during landing often has been proposed and observed as a mechanism for knee injury.[1–3] Medial collapse or knee valgus can be characterized as the combined motions of knee abduction, hip adduction, and hip or knee rotation. It also might be accompanied by pronation. Risk for this mechanism of injury is believed to be heightened during frequently occurring single-limb landings.[1,4,5] Because of this perceived risk, recent injury-prevention protocols have focused on reducing the occurrence of medial collapse during landing and cutting activities.[6–9] Research regarding what factors might place a person at risk for medial collapse and potential knee injury is limited. Factors that warrant consideration might include nonmodifiable factors, such as lower extremity structural alignment and sex, or modifiable factors, such as strength.

Both distal and proximal structural alignment might influence knee motion. Understanding these relationships might help clinicians better identify factors that contribute to joint positions associated with medial collapse and potential knee injury.[1,3] Alignment of the femur in the transverse plane might be influenced by the degree of femoral anteversion present at the femoral head and neck. Whereas many authors[10–13] have suggested that femoral anteversion could influence the landing mechanics associated with anterior cruciate ligament (ACL) injuries, only Loudon et al[12] have attempted to measure femoral anteversion in patients with ACL injuries.

Researchers[14–22] have attempted to validate methods for measuring femoral anteversion; however, the findings have been variable. One of the most common methods, the Craig test, or the trochanteric prominence angle test, has been validated in a population with cerebral palsy.[19] However, similar attempts to validate these methods using accepted computed tomography and magnetic resonance imaging have been inconsistent.[14,22] Using an alternative method of measurement, Kozic et al[15] reported that differences in prone internal rotation (IR) and external rotation (ER) of the hip highly correlated ($r = 0.93$) with radiographic measures of femoral anteversion. Their method consisted of subtracting ER from IR range of motion. We define this measurement of IR and ER as asymmetry of unilateral hip rotation (AUHR) (IR − ER = AUHR). People with greater anteversion have been observed as having greater IR relative to ER.[15,16,18,23–25]

Distally, pronation might be associated with medial knee collapse and knee injury. Greater amounts of pronation have been observed in patients with ACL injuries.[12,26–29] In these same patients, no differences were found in mean measures of pronation between sexes.[26,29] These results suggest that pronation might be a risk factor for ACL injury that is unaffected by sex.

The relationship between sex and knee biomechanics during cutting and landing activities has been studied extensively, with numerous authors[30–35] reporting differences between male and female participants. Similarly, less hip abduction strength has been associated with greater medial knee collapse among women.[36,37] Furthermore, femoral anteversion has been proposed to alter muscle mechanics and neuromuscular activity during hip abduction and ER.[13,38,39] Therefore, variations in hip strength and structure might result in a failure to maintain neutral lower extremity alignment during landing. Exploring the relationship among multiple factors that might influence landing mechanics provides additional information to the clinician for assessment and intervention.[6,40]

In addition to the previously studied factors of strength and sex, potential risk factors for medial knee collapse might include hip and foot structure. Therefore, the purpose of our study was to determine the relationship among measurements of AUHR; mobility of the foot, which we described as relative arch deformity (RAD); hip abduction–ER strength; sex; and medial collapse of the knee during a single-leg jump landing. We hypothesized that AUHR and RAD would be positively correlated with movements often associated with medial collapse of the knee, including hip adduction and IR excursions and knee abduction and rotation excursions.

METHODS

Participants

We conducted a power analysis using pilot data from previous research in our laboratory, which demonstrated means for knee abduction displacement for male and female participants of 3.04° ± 3.50° and 7.33° ± 6.08°, respectively. Assuming an α level of .05, the inclusion of 45 participants was expected to result in sufficient statistical power (0.80). We enrolled 30 women and 15 men in the study (age = 21 ± 2 years, height = 171.7 ± 9.5 cm, mass = 68.4 ± 9.5 kg). The participant pool consisted of recreationally active people recruited from the community as a sample of convenience. Inclusion criteria for participation were self-reported regular physical activity, no history of lower extremity injury in the 6 months before the study, no history of lower extremity surgery, age between 18 and 25 years, and willingness to participate in the study. We defined *regular physical activity* as participation in activity for a minimum of 30 minutes 3 times per week.

Shoes (Air Max Challenge; Nike, Inc, Beaverton, OR) were provided for the participants to wear during the study to control for differences in footwear. Participants' height and mass were measured before we instructed them to perform 3 trials of a standing, 2-footed, forward jump landing on 1 lower extremity. The lower extremity on which a participant chose to land in 2 of 3 trials was considered the preferred leg and was the only lower extremity tested during our study. Participants then completed a 5-minute warmup on a stationary exercise bicycle at a self-determined intensity level. All participants provided written informed consent, and the study was approved by the Institutional Review Board of the University of Kentucky.

Evaluation of AUHR

We measured hip range of motion for IR and ER and considered the difference between the measurements to be an estimate of femoral anteversion.[15,16,18,23–25] Rotation was measured with participants lying prone on the examining table with the hips in extension and knees actively flexed to 90°, with the patellae of both limbs lying even with the end of the table and with the knees spaced approximately 14 in (35.6 cm) apart (Figure 1). We used 14 in of spacing, which was novel to our study, to attempt to standardize limb position and control for variations in hip motion resulting from different amounts of abduction. This distance was chosen based on pilot testing in which we observed that participants often self-selected this position for comfort. A standard goniometer modified with the addition of a bubble level to ensure vertical alignment of the reference arm was used to measure IR and ER in degrees. Participants were instructed to actively flex their knees and to allow their hips to rotate internally and then externally under the force of gravity to their passive limit.[25] Both limbs were either internally or externally rotated simultaneously to aid in stabilization of the pelvis.[41] Each measurement was recorded 3 times to calculate an average measure of IR and ER from which AUHR was calculated. The same tester (J.S.H.) performed all hip rotation measurements. The intraclass correlation coefficients (2,1) were 0.99 (SEM = 0.5°) and 0.95 (SEM = 1.9°), respectively, for intratester between-days reliability (n = 10 with 1 day between tests) for measures of IR and ER a priori.

Figure 1. Measurement of hip range of motion in the prone position.

Evaluation of Longitudinal Arch Mobility

In our study, the foot and medial longitudinal arch were characterized by changes in dorsum height in response to loading.[42] Measurements were accomplished using the Arch Height Index Measurement System (JAK Tool and Model, LLC, Matawan, NJ) (Figure 2). Arch height was recorded in 2 stance conditions: 10% (AH10%) of weight bearing and 90% (AH90%) of weight bearing. To calculate RAD during loading, we used the equation proposed by Nigg et al[43] and modified by Williams and McClay[42]:

$$\mathrm{RAD} = \frac{(\mathrm{AH10\%} - \mathrm{AH90\%})}{\mathrm{AH10\%\ BW}}\ 10^4,$$

where body weight (BW) is expressed in newtons. Participants were weighed on a standard scale, and 10% and 90% of each participant's total weight were calculated. Participants stood with their hands resting on an examination table, which they used to assist in controlling their amount of weight bearing. The height of the examination table was raised or lowered as needed for participants to maintain balance. Participants then placed one foot on the scale and the other foot on an even, adjacent surface. The adjacent surface was the same height as the scale and was positioned just posterior and slightly medial to the stance limb being tested. This position was used to prevent leaning by the participants to one side, possibly influencing the foot height during the 2 weight-bearing conditions. The participants were instructed to control the amount of weight bearing by balancing directly over the scale and either touching the foot of the nontest limb to the adjacent surface or supporting their weight on the examination table until the scale showed that 90% of weight bearing had been achieved. The arch height then was recorded. The process was repeated for 10% of weight bearing. Participants were monitored carefully to ensure that their center of mass remained over the stance limb and that the knee was not bent. Intratester reliability of this instrument for calculating an arch height index using AH10% and AH90% has been reported (intraclass correlation coefficient = 0.94).[42] The same examiner (J.S.H.) performed all measurements.

Kinematic Analysis

Three-dimensional joint kinematics of the hip and knee were collected at 100 Hz using Flock of Birds electromagnetic sensors (Ascension Technology Corporation, Burlington, VT) and The MotionMonitor software (Innovative Sports Training, Inc, Chicago, IL). Electromagnetic sensors were placed on the skin of the preferred landing leg using adhesive pads and tape. One sensor was placed on each participant's sacrum, lateral thigh, and medial tibial plateau. The posterior-superior iliac spine, lateral and medial knee joint lines, and lateral and medial malleoli were digitized per manufacturer recommendations. A standard right-hand coordinate system was used for all joints so that the positive x-axis projected anteriorly and the positive z-axis projected superiorly. The landing task, as described in the jump protocol, was performed by each participant. Time of initial contact was identified using a foot switch placed in the

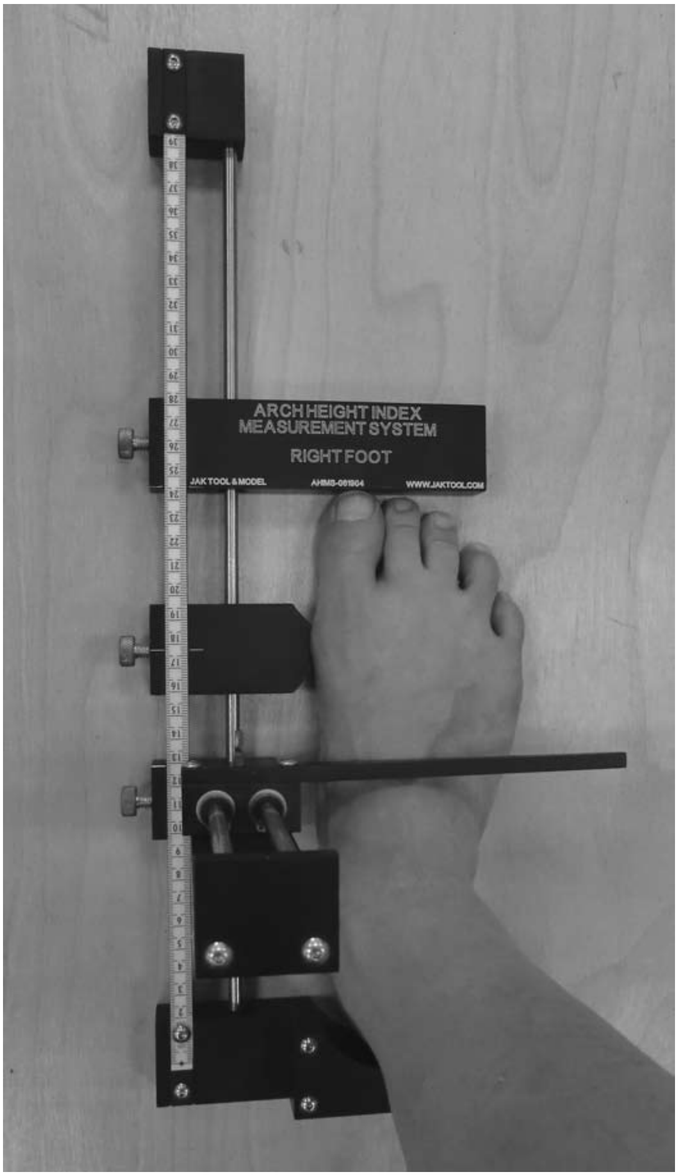

Figure 2. The Arch Height Index Measurement System (JAK Tool and Model, LLC, Matawan, NJ).

participant's shoe. Foot switches contained toe, midfoot, and heel sensors to accurately document ground contact regardless of landing technique. Foot-switch data were collected at 2000 Hz and synchronized with the kinematic data.

Jump Protocol

Participants performed a double-leg jump with a single-leg landing. Each participant started the task at a distance away from a designated target area that was equivalent to 40% of his or her height; the vertical component of the initial jump was equivalent to 115% of the participant's height.[37] To control for the vertical component of the jump, a foam block was hung at the appropriate height, and the participant was instructed to jump so the block just brushed the top of his or her head.[37] The participants were instructed to use a single-leg landing strategy and to stabilize quickly. They also were instructed to perform several familiarization repetitions of the jump-landing task with the preferred leg. A minimum of 3 practice trials were performed to ensure comprehension of the task. Additional trials were permitted if a participant did not feel comfortable with the task. Three test trials from each participant were used for analysis. A trial was discarded and an additional one recorded if a participant did not meet the height or distance requirement or if the single-leg landing was not held stable. To be considered stable, participants had to maintain unilateral stance for a minimum of 5 seconds after landing.

Strength Testing

After completing the jump-landing task, all participants underwent isometric strength testing for a motion of combined hip abduction and ER.[13] Strength of the preferred leg was assessed with the PrimusRS dynamometer (BTE Technologies, Inc, Hanover, MD). Participants were positioned side lying on a table with the preferred leg on top. The trunk was in neutral alignment, with the hips flexed to 45° and the knees flexed to 90° (Figure 3).[13] Participants were allowed to support their heads using the contralateral hand and arm and could stabilize themselves on the table using the ipsilateral hand. Participants remained in this position for the duration of the strength testing.

The pad on the resistance arm of the dynamometer was placed over the lateral side of the preferred knee. Participants abducted and externally rotated the preferred leg against the pad. The foot of the preferred leg was not allowed to touch the other foot to prevent pushing off but was required to remain below the level of the knee to prevent internal hip rotation. Three maximal contractions of 5 seconds each with a 30-second rest between trials were averaged to establish the maximal voluntary isometric contraction torque (newton-meters) for each participant, which was then normalized to body mass (kilograms) for further analysis.

Data Analysis

Initial joint angles for 3-dimensional motion at the hip and knee were measured at the time of ground contact as detected by foot switches. Joint excursions for the hip and the knee were calculated as the difference between the joint angle at initial ground contact and the maximal joint angle occurring between contact and maximal knee flexion (maximal angle − initial contact angle = joint excursion).[33,34] Excursions were considered for each direction of motion within a plane. For example, for frontal-plane hip motion, excursions were calculated in abduction and adduction. All joint angles were calculated using a segmental, local coordinate system such that *hip motion* was defined as movement of the thigh sensor relative to the sacral sensor and *knee motion* was defined as movement of the shank sensor relative to the thigh. Euler angle equations were used to estimate joint angles, and the Leardini method was used to estimate the hip joint center.[44,45] Joint position data were processed using a fourth-order, dual-pass, Butterworth filter with a cutoff frequency of 9 Hz, which was confirmed by residual analysis. Datapac 2K2 (Run Technologies, Mission Viejo, CA) was used for the processing and analyzing all kinematic data.

Statistical Analysis

This was a single-occasion, descriptive laboratory study. Joint angles for 3-dimensional motion at the hip and knee at initial contact and maximal joint excursions were considered

Figure 3. Hip abduction–external rotation strength test.

the response or dependent variables of interest for this study. The AUHR, RAD, sex, and hip strength were the explanatory or independent variables. Pearson product moment correlations were conducted to determine the relationship between continuous explanatory variables and hip and knee kinematics and to evaluate the relationship between AUHR and RAD. For kinematic variables for which a correlation was present, a multivariable linear regression model was used to determine which combination of factors (AUHR, RAD, sex, or hip abduction–ER isometric peak torque) were predictive of lower extremity kinematics of the hip and knee at ground contact and maximal excursion. All explanatory variables were included initially, then the models were backward reduced. The variables that contributed the least were removed at each step until all variables contributed to a concise model at the α level set a priori (.05). All statistical analyses were conducted with SPSS (version 15.0; SPSS Inc, Chicago, IL).

RESULTS

Descriptive values for strength and structural measures are presented in Table 1, and means for all kinematic variables are presented in Table 2. Overall, participants demonstrated greater ER than IR, with a mean AUHR of –6° ± 15° (range, –35° to 26°). Correlations between AUHR, RAD, normalized peak isometric torque for hip abduction–ER, and kinematic variables are listed in Table 3. Regression models for which more than one explanatory variable was different are presented in Tables 4 and 5. Asymmetry of unilateral hip rotation, sex, and hip abduction–ER strength were predictive of knee abduction excursion during landing (Table 4). Greater AUHR ($\beta = 0.13$), female sex ($\beta = 5.19$), and lower hip abduction–ER strength ($\beta = -5.41$) were predictive of greater knee abduction excursion. Asymmetry of unilateral hip rotation ($\beta = 0.08$) and female sex ($\beta = 2.77$) also were positively predictive of knee ER excursion (Table 5). Asymmetry of unilateral hip rotation was the only factor that contributed to a model for knee adduction excursion ($R^2 = 0.10$, $P = .03$, $\beta = -0.04$) and to a model for hip adduction excursion ($R^2 = 0.13$, $P = .02$, $\beta = 0.13$). Relative arch deformity was the only factor that contributed to a model for hip adduction at contact ($R^2 = 0.10$, $P = .04$, $\beta = 5.57$) and to a model for knee flexion excursion ($R^2 = 0.11$, $P = .03$, $\beta = -6.59$). Hip abduction–ER strength was the only factor that contributed to a model for knee extension excursion ($R^2 = 0.09$, $P = .04$, $\beta = 0.42$). Finally, both lower hip abduction–ER strength ($\beta = -8.82$) and female sex ($\beta = 4.01$) were predictive of greater knee IR (Table 6). No issues with multicollinearity were present in any of the final models.

During analysis, an extreme statistical outlier (>3 SD above the mean) was discovered within the RAD data (RAD = 3.78) and removed. The exclusion of this participant did not influence any of the previously reported regression equations involving AUHR; therefore, the participant's AUHR and landing kinematic data were included in all analyses that did not involve RAD.

DISCUSSION

The primary purpose of our study was to identify relationships among clinical measures of lower extremity structure, sex, and hip strength with landing kinematics. Our main findings were that AUHR, sex, and hip strength were predictive factors for knee abduction excursion during landing; AUHR and sex were predictive of knee ER excursion; and AUHR alone was predictive of hip and knee adduction excursions. In addition, RAD was predictive of hip adduction at contact and knee flexion excursion, and sex and hip strength were predictive of knee IR at contact. These results suggest that AUHR, sex, and hip abduction–ER strength might be associated with medial collapse during landing characterized by hip adduction, knee abduction, and knee ER.

Sex, Hip Strength, and AUHR

Our results indicated that as AUHR increased (suggesting greater relative femoral anteversion), participants experienced greater knee abduction excursion during landing, with female participants and those with weaker hip abductors and external

Table 1. Descriptive Values for the Structural and Strength Measures

Measure	Mean ± SD
Average hip internal rotation, °	29 ± 11
Average hip external rotation, °	35 ± 7
Asymmetry of unilateral hip rotation[a], °	−6 ± 15
Arch height in 10% of weight bearing, cm	6.58 ± −0.58
Arch height in 90% of weight bearing, cm	6.16 ± −0.51
Relative arch deformity[b], N^{-1}	0.96 ± 0.52
Peak hip abduction–external rotation torque[c], Nm/kg	1.07 ± 0.30

[a] Calculated by subtracting average external rotation from average internal rotation.
[b] Calculated as [(AH10% − AH90%)/(AH10% BW)] * 10^4, where AH90% is arch height in 90% of weight bearing, AH10% is arch height in 10% of weight bearing, and BW is body weight.
[c] Indicates normalized to body weight.

Table 2. Mean Joint Angles at Contact and Excursion from Contact to Peak Knee Flexion°

Motion	Mean Angle at Contact ± SD	Mean Excursion[a] ± SD
Hip flexion	18.40 ± 6.97	18.40 ± 6.97
Hip extension	NA	0.07 ± 0.24
Hip adduction	NA	10.58 ± 5.32
Hip abduction	7.49 ± 5.41	0.91 ± 1.76
Hip internal rotation	NA	7.59 ± 4.65
Hip external rotation	12.14 ± 6.88	1.29 ± 1.87
Knee extension	NA	0.12 ± 0.41
Knee flexion	4.04 ± 5.43	41.29 ± 6.34
Knee adduction	3.90 ± 5.87	1.32 ± 2.08
Knee abduction	NA	8.48 ± 5.77
Knee internal rotation	1.66 ± 6.68	4.39 ± 4.38
Knee external rotation	NA	4.92 ± 3.91

Abbreviation: NA, not applicable.
[a] Calculated by subtracting peak angle from contact angle.

rotators experiencing the most knee abduction excursion. Femoral anteversion often is cited as a possible risk factor for ACL or other knee injuries[10–13]; however, we were the first to attempt to include an estimate of femoral anteversion as a predictor of knee abduction, which is a kinematic position linked to ACL injury.[1,2,5,46]

As hypothesized, hip adduction excursion also was correlated with AUHR ($R = 0.36$). Although not a primary aim of our study, the high correlation ($R = 0.70$) observed between knee abduction excursion and hip adduction excursion was not surprising. Clinically, these combined motions are often called *knee valgus*.[47] When they occur in unison, these motions represent a medial collapse of the lower extremity.[47] Therefore, it would be functionally challenging for one to occur without the other in such a manner that balance between the hip and knee can be maintained.

To identify the functional influence of femoral anteversion, Nyland et al[13] studied the relationship between femoral anteversion and electromyographic (EMG) activation of hip musculature during an isometric exercise of combined hip abduction and ER. They reported that participants with greater IR had lower gluteus medius/hip abductor and vastus medialis/hip abductor EMG ratios than did participants with less rotation. Although we did not collect EMG data, our results demonstrated a potential relationship between AUHR and the frontal-plane movements of hip adduction and knee abduction. Based on the findings of Nyland et al,[13] it might be theorized that the positive correlation observed between AUHR and hip adduction is a result of less relative gluteus medius activation. A decrease in relative gluteus medius activation in those with increased femoral anteversion (a deformity in the transverse plane) might result in a loss of frontal-plane hip control, sending the hip into adduction and a corresponding knee abduction position.[13] Although this pattern is plausible, further investigation involving structural measures, kinematics, and electromyography is needed to elucidate this potential biomechanical relationship.

The observed negative relationship between hip abduction–ER strength and knee abduction excursion also supports a potential link between AUHR and hip muscular strength and function. These results are supported by the work of Arnold et al,[39] in which computer modeling demonstrated that increasing femoral anteversion decreases the abduction moment arm of the gluteus medius. This decreased moment arm could contribute to both variations in activation and a loss of frontal-plane control. This potential biomechanical relationship between AUHR and frontal-plane knee motion is further supported by the observed negative relationship between AUHR and knee adduction excursion ($R = -0.33$). This result suggests that lower AUHR might have a protective effect against medial knee collapse by increasing knee adduction during landing.

The results presented in Tables 3 and 4 demonstrate that participants with greater AUHR went into more hip adduction and that as AUHR increased and hip abduction–ER strength decreased, participants experienced greater knee abduction excursion. Extreme valgus position (composed, at least in part, of knee abduction) has occurred in landing and cutting actions, resulting in subsequent injury to the ACL.[1,3] Our results showed a relationship among greater AUHR, female sex, lower hip strength, and greater knee abduction, which is an established position of risk.

Our observed trend of women experiencing greater knee abduction excursion during landing is consistent with the results reported by several researchers.[30–35] The role of hip strength during landing has not been studied extensively and is less understood. In women, a negative correlation between eccentric hip abductor peak torque and knee abduction ($R = -0.61$) has been reported during a landing activity similar to the one we used.[36,37] We observed a similar finding with hip abduction–ER strength and sex as predictors of knee abduction excursion. These same factors also were predictive of knee IR at contact. We found no differences in normalized hip strength between sexes ($P = .7$) (data not shown), which is a finding that other investigators[36] also have observed. However, this finding is in contrast to findings reported by other authors[37,48,49] documenting differences in hip abduction–ER strength between sexes. More study is needed to fully understand the role of hip strength and its relationship with sex and the possible implications of these factors for medial knee collapse.

We found a predictive relationship among AUHR, sex, and knee ER excursion (Table 5). Participants with greater AUHR went into more knee ER during landing, with women experiencing the greatest ER excursion. The observed ER excursion might be a function in part of the degree of knee IR at contact. Sex and hip strength were predictive of knee IR at contact, with women who had less hip strength landing in the most internally rotated position (Table 6). Landing in an internally rotated position might have contributed to the greater ER excursion associated with the female sex. It is important to note that the 2

Table 3. Relationship Between Structural or Strength and Kinematic Variables

Variables	*R* Value	*P* Value
Asymmetries of unilateral hip rotation[a] and hip adduction excursion	0.36	.02
Asymmetries of unilateral hip rotation[a] and knee abduction excursion	0.51	<.001
Asymmetries of unilateral hip rotation[a] and knee adduction excursion	–0.33	.03
Asymmetries of unilateral hip rotation[a] and knee external rotation excursion	0.41	.005
Relative arch deformity and hip adduction at contact	0.32	.04
Relative arch deformity and knee flexion excursion	–0.34	.03
Peak hip abduction–external rotation torque and knee extension excursion	0.31	.04
Peak hip abduction–external rotation torque and knee abduction excursion	–0.37	.02
Peak hip abduction–external rotation torque and knee internal rotation at contact	–0.41	.005

[a] Calculated by subtracting average external rotation from average internal rotation.

Table 4. Regression Model[a] Predicting Knee Abduction Excursion

Variable	Parameter Estimate, β	Standard Error	95% Confidence Interval	Standardized Estimate	*P* Value
Asymmetries of unilateral hip rotation	0.13	0.05	0.04, 0.22	0.336	.006
Peak hip abduction–external rotation torque	–5.41	2.17	–9.79, –1.03	–0.278	.02
Sex					
Male	Reference				
Female	5.19	1.39	2.39, 8.83	0.374	.001

[a] Knee abduction excursion = 11.63 + 0.13 (asymmetries of unilateral hip rotation) – 5.39 (peak hip abduction–external rotation torque) + 5.09 (sex = female). Adjusted $R^2 = 0.47$, $P < .001$.

Table 5. Regression Model[a] Predicting Knee External Rotation Excursion

Variable	Parameter Estimate, β	Standard Error	95% Confidence Interval	Standardized Estimate	*P* Value
Asymmetries of unilateral hip rotation	0.08	0.04	0.01, 0.15	0.313	.03
Sex					
Male	Reference				
Female	2.77	1.12	0.50, 5.04	0.338	.02

[a] External rotation excursion = 3.58 + 0.08 (asymmetries of unilateral hip rotation) + 2.77 (sex = female). Adjusted $R^2 = 0.23$, $P = .001$.

Table 6. Regression Model[a] Predicting Knee Internal Rotation at Contact

Variable	Parameter Estimate, β	Standard Error	95% Confidence Interval	Standardized Estimate	*P* Value
Peak hip abduction–external rotation torque	–8.82	3.01	–14.90, –2.73	–0.39	.006
Sex					
Male	Reference				
Female	4.01	1.88	0.22, 7.79	0.29	.04

[a] Knee internal rotation at contact = 8.46 – 8.82 (peak hip abduction–external rotation torque) + 4.01 (sex = female). Adjusted $R^2 = 0.214$, $P = .002$.

kinematic variables were only moderately correlated ($R^2 = 0.27$) (data not shown), suggesting that in addition to contact position and sex, other factors, such as AUHR, might have contributed to knee ER excursion. Overall, this movement from a more internally rotated position at contact to a more externally rotated position during landing is consistent with a medial collapse knee injury mechanism, which might be exaggerated among women, especially those with lower hip ER–abduction strength.

In our study, greater anteversion resulted in greater knee external excursion during landing. In a previous study,[50] greater anteversion resulted in less tibial IR excursion (tibial IR excursion = 22.009 + 1.09 [navicular drop] – 1.083 [body mass index] – 0.0771 [anteversion]). Although the terminology is different, the motions described are similar. The results of our study and the study by Carcia and Houglum[50] suggest that femoral anteversion might influence transverse-plane knee motion during landing. Despite these similarities, several differences exist between our study and that of Carcia and Houglum.[50] For example, the landing task was different, the method for evaluating anteversion was not reported, and although the overall model (including femoral anteversion) was predictive of tibial IR, the contribution of femoral anteversion to the model was not at the α level established a priori ($P = .06$).[50] Each study provided additional information for better elucidating the roles of structure, strength, and function in neuromuscular patterns, such as landing, but additional research clearly is needed to identify consistent relationships between hip asymmetry/structure and knee position during landing tasks.

Despite these findings, the clinical applications of our results are influenced by the limitations of the instrumentation used to measure biomechanical movement. In our experimental design, a segmental local coordinate system was used such that knee IR was identified as rotation of the shank relative to the thigh. With this design, IR of the femur relative to a stable tibia would be reported as external knee rotation. This type of kinematic assessment makes drawing distinct conclusions regarding transverse-plane knee motion difficult because which segment of the kinetic chain is moving is not known. This method might explain our lack of findings in the transverse plane at both the hip and knee despite examining factors thought to primarily influence transverse kinematics.

Arch Mobility

Our hypothesis that RAD would predict medial collapse during landing was not supported by our results. We observed a correlation between RAD and hip abduction at contact. This negative relationship is consistent with the potential for medial collapse of the lower extremity.[51] However, the absence of a relationship with hip adduction or knee abduction excursion suggests that RAD does not relate to a medial collapse actually occurring during landing. No factors associated with medial collapse during landing were correlated with RAD. Although RAD might have influenced frontal-plane hip position at contact, this relationship did not continue throughout the landing process.

We observed a negative correlation between RAD and knee flexion excursion, suggesting that those experiencing greater arch deformation moved through a smaller range of knee flexion during landing. This relationship might represent a tradeoff in force absorption between knee flexion and arch mobility. If the impact of landing can be attenuated by arch collapse, knee flexion excursions might be lessened. Gross and Nelson[52] suggested that attenuation processes during landing might be learned muscular responses. However, Hargrave et al[53] did not report lower ground reaction forces or differences in knee flexion among participants with increased pronation. Further research with kinetic analysis is needed to determine whether this relationship is consistent and how it might influence ground reaction forces and subsequent risk for injury during landing activities.

Limitations

Our results are presented with the assumption that AUHR is an appropriate estimate of femoral anteversion. Although this method has been validated successfully in the literature, it was tested predominantly among 8- and 9-year-old children.[15] It is well documented that femoral anteversion decreases with growth from birth to age 16 years.[24] The smaller variation in degree of anteversion seen in normal, healthy adults possibly limits the accuracy of a test originally designed to evaluate the larger angles of anteversion seen in youth. In addition, this method was validated using biplane radiography, which is a technique documented as being less valid and more variable than computed tomography imaging.[54] We examined screening measures that could be performed in a clinical setting without extensive training or expensive equipment. We believe this is a reasonable method of examining the proposed risk factor for knee injury because greater IR relative to ER has been observed consistently in participants with above-average femoral anteversion.[15,16,18,23–25]

The strength of any conclusions drawn from our study is limited by the inability to fully understand what factors influence static measures of hip rotation. We have not found documentation of the percentage of hip range of motion determined by skeletal structure compared with soft tissue laxity. The ability to differentiate soft tissue laxity from bony structure at the hip or to control for soft tissue laxity at the hip could improve the use of AUHR as an estimate of femoral anteversion. Our use of 14 in (35.6 cm) as a standard spacing between knees during range-of-motion assessment similarly might have influenced hip rotation; in subsequent research, this spacing might need to be normalized to participant height or thigh length.

We defined AUHR as IR – ER. The mean of our sample groups was –6°, suggesting that, on average, they demonstrated more ER. Swanson et al[55] suggested that 30° of asymmetry is necessary to produce abnormal alignment. Why our participants demonstrated greater ER than IR is not clear. It might be a function of the age of our participants. Femoral anteversion and AUHR are documented to decrease with age.[19] Although we anticipated greater IR than ER, reports of rotational differences have varied in the literature, with AUHR (as defined in our study) reported as low as –21.9°.[56] The use of the term *AUHR* is specific to the equation and does not necessarily imply pathologic asymmetry in our sample.

Finally, an additional limitation of our study was the size and health of our sample. The sample size did not allow a robust comparison between sexes, which might have been helpful to fully eliminate sex as a confounding factor in our results. Because sex-related differences have been reported for hip strength and femoral anteversion, fully controlling for the effect of sex on these variables is difficult. However, we have accounted for sex by including it in the regression model and believe that review of the standardized parameter estimates supports our conclusion that the relationship among AUHR, strength, and landing kinematics is not due to sex alone. One limitation with investigating predictive relationships is that the variability of the data for moderate to small samples is compressed. In our sample, the variables of interest might not have been varied enough to detect some relationships. Although our standard deviations did not suggest this was the case with AUHR or RAD, it might have been the case for hip strength (Table 1). Despite these potential limitations of our sample, we believe our participants represented a physically active population that is clinically relevant.

CONCLUSIONS

Changes in AH in response to loading (RAD) were weakly correlated with hip adduction at contact and knee flexion excursion. These results suggested that participants with greater arch mobility made contact with the ground in greater hip adduction and experienced less knee flexion during landing. The RAD was not correlated with any kinematic motions associated with medial collapse of the hip or knee (hip adduction excursion and knee abduction and rotation excursion) during landing.

The factors of being a woman, having greater AUHR, and having less hip abduction–ER strength were linked with various movements associated with medial collapse during landing, including hip adduction, knee abduction, and knee ER.

Further research is warranted to evaluate the possible relationships among femoral anteversion, hip range of motion, and dynamic knee abduction and to evaluate how these values might be incorporated prospectively into a multifactorial approach for knee injury screening and prevention.

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Address correspondence to Jennifer S. Howard, PhD, ATC, Rehabilitation Sciences, 210 Wethington Health Sciences Building, 900 South Limestone Street, University of Kentucky, Lexington, KY 40536. Address e-mail to j.s.howard@uky.edu.