The Effects of Gluteal Strength and Activation on the Relationship Between Femoral Alignment and Functional Valgus Collapse during a Single-Leg Landing

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ABSTRACT

Context. A bias toward femoral internal rotation is a potential precursor to functional valgus collapse. The gluteal muscles may play a critical role in mitigating these effects. **Objective.** Determine the extent to which gluteal strength and activation mediate the associations between femoral alignment (femoral anteversion and internal and external rotation hip ROM (ROM_{IR} and ROM_{ER})) and functional valgus collapse during a single-leg forward landing. **Design.** Cross-sectional. **Setting.** Research laboratory. **Patients or Other Participants.** Forty-five females (20.1 ± 1.7 yrs, 165.2 ± 7.6 cm, 68.6 ± 13.1 kg) and forty-five males (20.8 ± 2.0 yrs, 177.5 ± 8.7 cm, 82.7 ± 16.5 kg), healthy for six months prior. **Intervention(s).** Femoral alignment was measured prone. Hip extension and abduction maximal voluntary isometric contractions were obtained using a hand-held dynamometer. Three-dimensional biomechanics and surface electromyography were obtained during single-leg forward landings. **Main Outcome Measures.** Forward-stepwise multiple linear regressions determined the influence of femoral alignment on functional valgus collapse and the mediating effects of gluteus maximus and medius strength and activation. **Results.** In females, once accounting for femoral alignment, less hip abduction strength predicted greater peak hip adduction angle (R^2 change=.10, $p=.02$) and greater hip extensor activation predicted greater peak knee internal rotation angle (R^2 change=.14, $p=.01$). In males, lesser hip abduction strength predicted smaller peak knee abduction moment (R^2 change=.11, $p=.03$), and the combination of lesser hip abduction peak torque and lesser gluteus medius activation predicted greater hip internal rotation angle (R^2 change=.15, $P=.04$). No meaningful mediation effects were observed ($v_{adj} < .01$). **Conclusions.** In females, after accounting for femoral alignment, less gluteal strength and higher muscle activation were only marginally associated with kinematic components of functional valgus

24 collapse. In males, less gluteal strength was associated with a more varus posture. Gluteal
25 strength did not mediate femoral alignment in either sex. Future research should determine the
26 capability of females to use their strength efficiently.

27

28 Key words: ACL, anteversion, hip RoM, gluteal, biomechanics, mediation

For Peer Review

Functional valgus collapse is a high-risk biomechanical movement strategy thought to contribute to anterior cruciate ligament (ACL) rupture.¹ Functional valgus collapse describes the combined lower extremity motions of hip adduction, hip internal rotation, knee (tibial) abduction and knee (tibial) internal rotation,^{2,3} and often includes increased joint moments associated with these movements.^{4,5} This pattern is more often observed in females.⁶ To explain this sex disparity, researchers have sought to identify anatomical characteristics which may predispose females to greater valgus collapse. Because a valgus collapse strategy consists of coupled movements between the hip and knee,^{7,8} anatomical characteristics affecting the hip may potentially influence components of functional valgus collapse. Despite the non-modifiable nature of anatomical characteristics, understanding their influence on lower extremity biomechanics will allow clinicians to focus their interventions on individuals at greatest risk of injury, thus leveraging their limited time and resources.

Femoral anteversion and passive hip internal rotation range of motion (RoM) are greater in females⁹ and are separately thought to contribute to greater movement toward functional valgus collapse by predisposing the hip toward greater adduction and internal rotation.^{10–12} Femoral anteversion is characterized by a medially-torsioned femoral neck and is developmental.¹³ In contrast, passive hip RoM is a pliable soft tissue restraint which may be influenced by other factors such as general joint laxity,¹⁴ that has been shown to remain constant throughout maturation,¹⁵ and has been retrospectively associated with ACL injury.¹⁶ Though these factors cannot be directly modified through training interventions, it may be possible to design more effective intervention strategies to *counteract* non-modifiable anatomical factors.

Strengthening the gluteus maximus and medius muscles may represent modifiable intervention targets to mitigate the negative effects of excessive femoral anteversion or passive

hip RoM.¹⁷ As primary extensors, external rotators, and abductors of the hip, these muscles work eccentrically to control dynamic hip internal rotation and adduction, two primary components of functional valgus collapse. This is particularly true in a single-leg stance when more gluteal activation is needed to maintain a stable pelvis. Moreover, greater degrees of femoral anteversion or passive hip internal rotation RoM result in lengthened moment arms for the gluteus maximus and medius,¹⁸ thus potentially affecting their torque generation capabilities. Although an acutely longer moment arm would result in greater force production capacity, chronic muscle lengthening is thought to reduce force generating ability.¹⁹ To this point, research suggests that greater femoral anteversion and internal rotation RoM may each be associated with decreased gluteal efficiency,^{20,21} as evidenced by decreased torque generation (maximal voluntary isometric contraction: MVIC) coupled with increased surface electromyography (EMG) amplitude. This neuromuscular profile is suggestive of an individual with weaker and less effective gluteals, who must use a greater proportion of available strength to complete a given task. In so doing, the muscle could more readily fatigue, and be unable to guarantee safe movement strategies. While addressing gluteal strength and activation may represent a viable intervention target in individuals displaying increased femoral anteversion or passive internal rotation RoM, the extent to which strength and activation of the gluteals directly mediates problematic femoral alignment has not yet been elucidated.

Therefore, the objective of this study was to determine the extent to which gluteal muscle strength and activation mediate associations of femoral anteversion and passive internal and external rotation hip RoM with components of functional valgus collapse (peak hip and knee frontal and transverse plane angles and external joint moments) during a single-leg forward landing task in females and males. We hypothesized that increased femoral anteversion and

internal rotation RoM (RoM_{IR}) and decreased external rotation RoM (RoM_{ER}) would predict greater peak hip and knee angles and moments associated with functional valgus collapse (operationally defined as greater hip adduction and internal rotation, and knee abduction and internal rotation) during a single-leg forward landing task. We further hypothesized that the addition of increased gluteal activation with decreased hip strength (mediating variables) would strengthen the overall prediction of the biomechanical variable, while weakening the relationship between femoral alignment and biomechanics. This would suggest that gluteal strength and activation may represent effective intervention targets specifically in individuals with more anteverted and internally rotated femurs.

METHODS

Participants

Forty-five female participants (20.1 (1.7) yrs, 165.2 (7.6) cm, 68.6 (13.1) kg) and forty-five male participants (20.8 (2.0) yrs, 177.5 (8.7) cm, 82.7 (16.5) kg) were recruited for a single session. This sample size was based on an *a priori* power analyses of pilot data in a female cohort and determined to be adequate to detect moderate effect sizes with a power of .80. To ensure a homogenous sample, specific inclusion criteria were: 1) adults between the ages of 18 and 25 and 2) a score of two or more (at least “one time in a week”) on categories 2-4 (“cutting, “decelerating”, and “pivoting”) of the Marx Activity Rating Scale.²² Specific exclusion criteria were: 1) any history of lower extremity surgery, 2) any history of ligamentous or meniscal or labral knee or hip injury, 3) any history of lower extremity injury within the previous 6 months, 4) history or diagnosis of a vestibular condition affecting balance, and, 5) history or diagnosis of any cardiovascular condition precluding exercise. Each participant provided written informed consent as approved by the university’s Institutional Review Board.

98 **Procedures**

99 **Anatomical Measures.** Both femoral anteversion and hip RoM were measured prone
100 with a standard inclinometer with the hip in neutral and the knee flexed to 90°. ^{23,24} To measure
101 femoral anteversion, the examiner rotated the lower leg while palpating the greater trochanter.
102 When the greater trochanter was at its most lateral point, the transverse plane angle formed by
103 the tibial shaft and true vertical was measured as femoral anteversion. To measure hip RoM, the
104 examiner passively internally and externally rotated the lower leg while palpating the sacrum. At
105 the point of initial sacral movement, the transverse angle formed by the tibial shaft and true
106 vertical was measured as RoM_{IR} and RoM_{ER}, respectively (Figure 1). Three trials were taken for
107 each measure and averaged for analysis. The same examiner took all measurements, and had
108 previously established good to excellent inter-day reliability [ICC_{2,3}(SEM); femoral anteversion:
109 .92(1.2°); internal rotation RoM: .97(1.6°); external rotation RoM: .85(3.3°)]. ²⁵

110 **Surface Electromyography Instrumentation.** Each participant was outfitted with
111 reusable double differential 27 x 33mm silver surface electromyography (EMG) electrodes with
112 a 10mm inter-electrode distance (CMRR > 80dB; Overall Channel Noise <0.75uV) (Trigno
113 Wireless Sensors, Delsys, Boston, MA, USA) to acquire signals from the gluteus medius and
114 gluteus maximus during maximal strength testing and performance of the single-leg forward
115 landing task. Prior to sensor placement, the skin was cleaned with an alcohol swab. The gluteus
116 medius electrode was placed one-third the distance from the iliac crest to the greater trochanter.
117 The electrode on the gluteus maximus was placed halfway between the second sacral vertebrae
118 to the greater trochanter. All electrodes were positioned parallel to muscle fiber orientation and
119 secured with tape or prewrap. Proper positioning was verified with manual muscle testing and
120 visual inspection of the EMG signal using EMGWorks (Delsys, Boston, MA, USA). EMG data

were sampled at 1000Hz and were synchronized with kinematic and kinetic data during post-collection processing. Specifically, custom MATLAB script was written to extract the instant of initial ground contact from the sagittal plane acceleration signal integrated in the Delsys Trigno gluteus medius EMG sensor. Ground contact from the GRF data was defined as vGRF exceeding 10N and ground contact from the accelerometer data was defined as the instant at which acceleration crossed zero (became negative). Once initial ground contact was identified for EMG data, all data sources were synced in MATLAB (MathWorks, Inc., Natwick, MA, USA).

Maximal Voluntary Isometric Contractions (MVICs). Each participant completed a five minute warm up on a stationary bike at a self-selected pace. Following warm-up, MVICs of the gluteus maximus (hip extension) and gluteus medius (hip abduction) were obtained and used as representations of maximum torque generation values, as well as for normalization of EMG amplitude. For all MVIC measures, a strap was used to secure the hand-held dynamometer in place and to provide resistance for the participant. For MVIC measurement of the hip extensors, the participant was positioned prone with the knee bent to 90 degrees. With a handheld dynamometer (Lafayette Instruments, Boston, MA, USA) placed over the posterior distal thigh two inches proximal to the joint line, the participant was asked to maximally contract into hip extension.²⁶ Hip abduction MVICs were measured side-lying on the right side, with the left leg up. The left leg was placed in 10-15 degrees of hip extension and slightly externally rotated, thus isolating the gluteus medius. Maximal hip abduction was resisted by placing the lower edge of the dynamometer two inches proximal to the lateral knee joint line.²⁷ Prior to collecting each MVIC, participants were familiarized to the measure and allowed up to three submaximal practice trials. Each condition consisted of three 5-second trials, with 30 seconds rest between

144 trials. To prevent an artificial spike in dynamometer output during collection, each participant
145 was instructed to slowly increase force output, reaching maximum force production during the
146 third second of the five second trial. Prior to the study the investigator established day-to-day,
147 intra-rater reliability for strap-assisted handheld dynamometry [$ICC_{2,2}(SEM)$; hip extension:
148 .76(33.3N); hip abduction: .96(15.7N)].

149 ***Biomechanical Instrumentation.*** Prior to digitization for motion capture, participants
150 were outfitted with standardized shoes (Adidas Uraha 2, Adidas AG, Herzogenaurach, Bavaria)
151 to eliminate potential differences in shoe-surface interactions. Participants were adorned with
152 six marker clusters, placed at each of the following locations: lateral aspect of the left foot,
153 lateral aspect of the left lower leg (mid-shaft), medial and lateral proximal tibial flares, the lateral
154 left thigh (mid-shaft), the L5-S1 junction, and the postero-superior thorax (C7-T1 spinous
155 processes). The investigator manually digitized anatomical landmarks on each participant with a
156 stylus using the MotionMonitor software (Innovative Sports Training, Chicago, IL, USA). Joint
157 centers for the knee and ankle were defined as the midway point between the medial and lateral
158 femoral condyles and medial and lateral malleoli, respectively. The hip joint center was
159 determined using the Bell method.²⁸ A segment-based coordinate system was used to define
160 each body segment. The X-axis was defined as the anterior-posterior axis (adduction/abduction),
161 the Y-axis was the distal-proximal axial axis (internal/external rotation), and the Z-axis was
162 defined as the medial-lateral axis (flexion/extension). Motions for each joint were calculated
163 using Euler's equations (Z Y' X"). Kinematic data were obtained with an 8-camera optical LED
164 system (Impulse, Phase Space; San Leandro, CA, USA) at a sampling rate of 240 Hz. Kinetic
165 data were obtained using an embedded Bertec forceplate (Type 4060-130; Bertec Corporation,
166 Columbus OH, USA) and were sampled at a rate of 1000 Hz. Kinematic and kinetic

instrumentation were interfaced with MotionMonitor software and were synchronized by a pulse trigger during each trial. Following acquisition the kinematic data were interpolated to 1000 Hz to allow synchronization of data sources.

Procedure for Single-Leg Forward Landing. Participants were familiarized to the task prior to data collection and were allowed up to three practice trials. Tape was placed on the ground at a distance equal to 40% of each participant's height away from the front edge of the forceplate. A foam barrier equal to 15% of the participant's height²⁹ was placed halfway between the tape and the edge of the forceplate. Instructions to participants were standardized. Participants were instructed to stand behind the tape and jump from two legs over the foam barrier, landing on their left foot (Figure 2). Trials were discarded if the participant double-hopped upon landing, hit the barrier, didn't clear the barrier with both feet, didn't land with the entire left foot on the forceplate, or used their contralateral limb for additional support. Five clean trials were collected and used for analysis.

Data Handling and Processing. Motion capture was collected two seconds before initial ground contact, defined as the point at which the vertical ground reaction force exceeded 10N, and continued for three seconds after initial ground contact. All flexions, adductions, and internal rotations were defined as positive, and all extensions, abductions, and external rotations were defined as negative angles. To determine the appropriate common filter for kinematic and kinetic data, a residual analysis was conducted on the ground reaction force (GRF) in a subset of the trials, according to the procedure described by Winter.³⁰ The noise to signal ratio was determined to be most optimum at 10Hz. Therefore, all kinetic and kinematic data were filtered with a 10 Hz low-pass, zero-lag, 2nd order Butterworth filter. Inverse dynamics were then computed to determine external moments for each joint, which were then normalized to each

participant's height (meters) and mass (kilograms); normalization to both height and mass is recommended to conservatively evaluate between-subject differences and to reduce Type I error.³¹ Kinematic and kinetic data were then exported to MATLAB (MathWorks, Inc., Natwick, MA, USA) for further reduction using custom-written script.

For reduction of MVIC and EMG data, the peak force (N) for each MVIC condition was multiplied by the moment arm length (as determined by Dempster's segmental data³² minus the distance from the respective joint line to the center of the dynamometer's contact pad) then divided by participant body mass, resulting in a normalized torque, $\text{N} \cdot \text{m} \cdot \text{kg}^{-1}$, which is consistent with methods commonly reported in clinical literature.³³ These torque values were used to represent maximum torque generation. To obtain MVIC muscle activation amplitude, the EMG data were filtered in MATLAB using a band-pass 20-350 Hz, 4th order, zero-lag, Butterworth filter with full-wave rectification, and was processed using a root mean squared (RMS) algorithm with a 25-millisecond time constant. The peak RMS EMG amplitude was extracted from each trial's landing phase (initial contact – peak knee flexion) and averaged within each condition, resulting in an averaged peak EMG amplitude for each MVIC direction that was used to normalize the EMG signal (% MVIC) obtained during the landing task. Therefore, peak amplitude was defined as the average peak RMS signal across trials within each MVIC condition.

MATLAB script was written to extract all variables from the exported data. Specifically, initial ground contact (the point at which vertical ground reaction force exceeded 10N) marked the beginning of the landing phase. The end of the landing phase was the point of maximum knee flexion. Peak hip and knee angles were extracted (sagittal, frontal, and transverse planes), and were defined as the maximum respective joint angle occurring during the landing phase.

Peak external moments were obtained for the hip and knee in each cardinal plane, and were defined as the maximum normalized moment during the landing phase.

Statistical Analysis

A stepwise regression procedure was used to determine the direct influence of femoral anteversion and hip RoM on functional valgus collapse and the mediating effect of gluteal strength and activation. Femoral anteversion, RoM_{IR} , and RoM_{ER} were entered on the first step of each separate forward stepwise multiple linear regression analysis predicting frontal and transverse plane hip and knee peak angles and peak joint moments associated with hip adduction and internal rotation and knee abduction and internal rotation (P entry $< .20$). The relative contributions of each independent variable were inspected using semi-partial (part) correlations, which also allowed for observation of interaction between variables. Strength and activation for each gluteal muscle were then entered on the second step of the forward stepwise method, and R squared changes were examined. To thoroughly investigate and quantify any mediating effects of gluteal strength and activation on relationships between anatomy and biomechanics, the recently developed adjusted epsilon (v_{adj}) statistic was calculated for each instance in which the addition of neuromuscular variables weakened or strengthened the part correlations between anatomy and biomechanics. v_{adj} is defined as the percentage of variance in Y that is jointly explained by X and the mediating variable, while also controlling for spurious correlation arising from the ordering of variables. It is interpreted as percent of variance explained. Recommended benchmarks for small, medium, and large effect sizes are .01, .15, and .25, respectively.³⁴ The MBESS package³⁵ in R³⁶ was used to compute v_{adj} values. Lastly, because both empirical evidence and functional anatomy indicate that the amount of potential knee valgus is dependent upon sagittal plane kinematics,³⁷ peak hip and knee flexion were included as control variables in

all analyses. Furthermore, due to the potential sex-specific nature of this mechanism, all analyses were conducted separately for each sex. SPSS (Version 21, IBM Corp, Armonk, NY) was used for all analyses unless otherwise noted.

RESULTS

Descriptive and bivariate correlation statistics among independent variables are presented in Tables 1 and 2, respectively. Descriptive statistics for biomechanical variables are presented in Table 3. Three aberrant values for gluteus medius activation were observed in the male cohort, which substantially altered regression analyses. These participants were omitted from all analyses, resulting in 42 male participants. Females had greater femoral anteversion and RoM_{IR} and less hip extension and hip abduction peak torque, confirming the need for sex-stratified analyses. Complete stepwise multiple regression results for frontal and transverse plane hip and knee biomechanics are presented in Tables 4 and 5.

Hip Kinematic and Kinetic Peak Values

In females, anatomical variables were not associated with peak hip adduction, though less hip abduction strength independently explained greater peak hip adduction angle during landing (R^2 change = .10, P = .02). Greater RoM_{IR} was associated with greater peak hip internal rotation angle (R^2 change = .12, P = .02) and lesser RoM_{ER} was associated with greater peak hip internal rotation joint moment during landing (R^2 change = .12, P = .02) (Table 4). Gluteal strength and activation did not alter the relationship between femoral alignment and biomechanics.

In males, greater total hip rotation (RoM_{IR} + RoM_{ER}) was associated with greater peak hip adduction angle during landing (R^2 change = .25, P = .01). The combination of greater femoral anteversion and lesser RoM_{ER} was associated with greater peak hip internal rotation angle (R^2 change = .19, P = .01), while greater RoM_{IR} was associated with greater peak hip

internal rotation moment during landing (R^2 change = .16, $P = .01$). The combination of lesser hip abduction peak torque and lesser gluteus medius activation explained additional variance in greater hip internal rotation angle (R^2 change = .11, $P = .05$). However, all mediation effects were negligible ($v_{adj} < .01$) (Table 4).

Knee Kinematic and Kinetic Peak Values

In females, anatomical variables were not associated with knee abduction angles or moments. The combination of less femoral anteversion and greater total hip rotation RoM was associated with greater peak knee internal rotation during landing (R^2 change = .16, $P = .04$). Greater gluteus maximus activation explained additional variance in peak knee internal rotation (R^2 change = .14, $P = .01$), though the mediation effect was negligible ($v_{adj} < .01$) (Table 5).

In males, greater total hip rotation RoM was associated with greater peak knee abduction angle (R^2 change = .26, $P = .01$) and moment (R^2 change = .13, $P = .06$) during landing. The inclusion of greater hip abduction torque explained additional variance in peak knee abduction moment (R^2 change = .11, $P = .03$). Greater RoM_{ER} predicted greater peak knee internal rotation angle (R^2 change = .03, $P = .02$), and greater gluteus medius activation explained additional variance in peak knee internal rotation angle (R^2 change = .18, $P = .01$). However, all mediation effects were negligible ($v_{adj} < .01$) (Table 5).

DISCUSSION

We expected that an anatomical bias towards internal hip rotation, as evidenced by greater femoral anteversion, greater RoM_{IR}, and lesser RoM_{ER}, would precipitate greater movements and moments that comprise functional valgus collapse. We hypothesized that strength and activation of the gluteus maximus and medius would explain additional variance in these biomechanical variables and that inefficient use of the gluteus maximus and gluteus

medius, operationally defined as gluteal muscles displaying lower torque production and higher percentage activation, would mediate the relationships between femoral alignment and biomechanics. Our results partially supported these hypotheses. In females, greater RoM_{IR} and less RoM_{ER} predicted greater hip internal rotation angle and joint moment, respectively. In males, greater RoM_{IR} and RoM_{ER} were associated with greater frontal plane hip and tibial angles and moments. In both sexes, weaker gluteals activating to a higher degree marginally explained additional biomechanical variance without having much effect on the correlations between anatomy and biomechanics, suggesting that the gluteal muscles may act independently of one's femoral alignment, instead of directly mediating the effects of alignment.

Gluteal Influences on Hip and Knee Biomechanics

In females, once accounting for the influence of femoral alignment on lower extremity biomechanics, weaker hip abduction peak torque explained an additional 10% of greater hip adduction angle, and higher hip extension activation explained an additional 14% of greater knee internal rotation angle. These effects were consistent with our hypothesis that weaker muscles activating to a higher degree would predict greater components of functional valgus collapse, as hip adduction and knee internal rotation are primary components of a valgus collapse mechanism.^{3,37} The additional variance explained by gluteal function without impacting the correlations with anatomy is important because it suggests that hip musculature may act independently of femoral alignment, irrespective of one's anatomy.

In males, once accounting for the influences of femoral alignment on hip and knee biomechanics, gluteal function explained up to an additional 18% of the variance in biomechanical variables. Specifically, less hip abduction strength predicted smaller knee abduction moments, greater hip adduction angle, and greater hip internal rotation angle; while

greater gluteus medius activation predicted greater knee internal rotation angle and lesser hip internal rotation angle. This is reasonable, as the hip abductors do work to limit hip adduction and internal rotation. It appears that men are able to effectively use hip abduction strength, whereas women do a poor job of activating weak abductors (Table 5).

The Mediating Effects of Gluteal Muscles on the Relationship between Femoral Alignment and Functional Valgus Collapse

Although we hypothesized that gluteal function would mediate the relationships between one's femoral alignment and lower extremity biomechanics, this effect was not observed in either sex. That is, the inclusion of gluteal muscle strength and activation did not meaningfully alter (neither weakened nor strengthened) the relationships between anatomy and biomechanics. If we had observed meaningful mediation, it would have demonstrated that gluteal function directly counteracts one's femoral alignment, and as such, gluteal function may have been mechanically linked to transverse plane femur alignment. In the present cohort however, zero-order correlations between femoral alignment and gluteal function were low (r range = .00 - .39), indicating gluteal strength and activation are not related to transverse plane femoral alignment. Improving gluteal strength and activation may not be more effective for individuals with increased femoral rotation than it would be for the general athletic population. These data would suggest that the gluteal muscles are able to affect safer movement patterns regardless of one's femoral alignment, though further research is needed to determine the efficacy of neuromuscular interventions in individuals with differing degrees of femoral alignment.

The Relationship between Hip Anatomy and Functional Valgus Collapse

The most consistent theme observed with respect to femoral alignment and lower extremity movement is that individuals with more lax hips, as indicated by greater RoM_{IR} and

RoM_{ER}, were more likely to display hip and knee biomechanics that contribute to functional valgus collapse. This relationship was more pronounced in males; an interesting finding, considering that females tend to demonstrate greater laxity. Though males in the current cohort were generally less lax than the females, it is possible that *if* a male has greater total hip RoM than average, he may be at greater risk. In males, hip RoM is more discriminative than in females. In females, either greater RoM_{IR} or lesser RoM_{ER} contributed to more risky hip biomechanics, while total RoM was only associated with higher peak knee internal rotation angle. Further research should examine the importance of the amount of RoM_{IR} relative to RoM_{ER} versus absolute total RoM values. Excessive RoM_{IR} in the presence of minimal RoM_{ER} could be just as problematic, if not more problematic, than excessive RoM_{IR} and RoM_{ER}, particularly in females. It is possible that this capsular pattern encourages a sex-specific valgus collapse mechanism by increasing the bias toward internal femoral rotation, an idea supported by evidence demonstrating that female sex and a bias toward RoM_{IR} combine to predict greater knee valgus.¹⁰ If this is the case, then it may present an opportunity to intervene upon females' movement strategies. If it is possible to increase RoM_{ER} in females through preventative programs, it may aid in the mitigation of potentially injurious frontal and transverse plane biomechanics.

Also of note, femoral anteversion displayed differential effects in males and females. In males, greater femoral anteversion was indicative of greater hip internal rotation angle; **whereas in females femoral anteversion was marginally associated with greater knee abduction and lesser knee internal rotation angles.** RoM_{IR} and femoral anteversion in females were highly correlated ($r = .76$), potentially masking femoral anteversion effects. In males, RoM_{IR} and femoral anteversion appear to be **more** separate constructs ($r = .54$), each contributing a unique portion of

variance to movement patterns. Furthermore, it suggests that in females, bony alignment (anteversion) may dictate capsular configuration (ROM), while in males, bony structure and capsular limitations are decoupled. Should this be the case, it would be useful for clinicians to consider both femoral anteversion and RoM_{IR} as factors in male patients, but in females it would suffice to obtain either RoM_{IR} or femoral anteversion. Further research needs to be completed to determine specific influencers of RoM_{IR} and the degree to which it is modifiable in males and females.

Limitations

We acknowledge there were limitations in our study. Although our sample of young, active individuals was one in which ACL injuries typically occur, it is possible that our included participants are not among those at-risk for ACL injury. As such, our participants possibly had no need to compensate for sub-optimal femoral alignment, which could have resulted in a dampened effect. We also acknowledge the assumption that peak joint angles represented the time at which participants were most vulnerable to injury. Although this is common practice in lower extremity biomechanical research, the use of discrete peak joint angles may not have fully captured one's biomechanical risk profile. We also assumed that participants put forth maximal exertion during MVIC measurement. Though we did provide practice trials with verbal feedback, we cannot guarantee that obtained MVICs truly represented maximal effort. We also assumed that the instant of ground contact as determined by GRF data is temporally aligned with the instant of ground contact determined from accelerometer data, though this method is preceded in the literature.³⁸ We did not examine the influence of trunk movement on hip and knee biomechanics, which is known to contribute to knee loading,³⁹ and may have helped account for the ~70% of variance in hip and knee biomechanics left unexplained in the current

study. Lastly, we opted to use the recently developed upsilon statistic to quantify mediation. The developers of this statistic acknowledge it may be overly conservative.³⁴ The upsilon statistic was selected for the current study because it is a continuous effect size measure, is interpreted similarly to an R^2 , and the commonly used indirect ratio effect for mediation is flawed and not recommended.⁴⁰ Nevertheless, there remains the possibility of a statistically significant mediation effect.

CONCLUSIONS

In conclusion, this study demonstrated that in females, after accounting for femoral alignment, less gluteal strength and higher muscle activation were only marginally associated with components of functional valgus collapse. In males, less gluteal strength and higher activation were associated with joint moments consistent with a more varus posture. Males appear to be more capable of using their musculature to affect safer biomechanics than are females. In neither sex did gluteal strength and activation directly mediate femoral alignment. Thus, interventions targeting improved gluteal function may not be more appropriate for individuals with femoral torsion than they would be for the general population. Further work is needed to determine why females are unable to use their existing strength as effectively as males.

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REFERENCES

1. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med*. 2005;33(4):492-501.
doi:10.1177/0363546504269591
2. Meyer EG, Haut RC. Anterior cruciate ligament injury induced by internal tibial torsion or tibiofemoral compression. *J Biomech*. 2008;41:3377-3383.
doi:10.1016/j.jbiomech.2008.09.023
3. Ireland ML. Anterior cruciate ligament injury in female athletes: epidemiology. *J Athl Train*. 1999;34(2):150-154.
<http://search.proquest.com/docview/206645539/fulltextPDF?accountid=14604>. Accessed February 21, 2015.
4. Jones PA, Herrington LC, Graham-Smith P. Technique determinants of knee abduction moments during pivoting in female soccer players. *Clin Biomech*. 2016;31:107-112.
doi:10.1016/j.clinbiomech.2015.09.012
5. Sigward SM, Powers CM. Loading characteristics of females exhibiting excessive valgus moments during cutting. *Clin Biomech (Bristol, Avon)*. 2007;22(7):827-833.
doi:10.1016/j.clinbiomech.2007.04.003
6. Boden BP, Torg JS, Knowles SB, Hewett TE. Video analysis of anterior cruciate ligament injury: Abnormalities in hip and ankle kinematics. *Am J Sports Med*. 2009;37(2):252-259.
doi:10.1177/0363546508328107

7. Hollman JH, Galardi CM, Lin IH, Voth BC, Whitmarsh CL. Frontal and transverse plane hip kinematics and gluteus maximus recruitment correlate with frontal plane knee kinematics during single-leg squat tests in women. *Clin Biomech*. 2014;29(4):468-474. doi:10.1016/j.clinbiomech.2013.12.017
8. Imwalle LE, Myer GD, Ford KR, Hewett TE. Relationship between hip and knee kinematics in athletic women during cutting maneuvers: a possible link to noncontact anterior cruciate ligament injury and prevention. *J Strength Cond Res*. 2009;23(8):2223-2230. doi:10.1519/JSC.0b013e3181bc1a02
9. Nguyen A-D, Shultz SJ. Sex differences in clinical measures of lower extremity alignment. *J Orthop Sports Phys Ther*. 2007;37(7):389-398. doi:10.2519/jospt.2007.2487
10. Howard JS, Fazio MA, Carl G, Uhl TL, Jacobs CA. Structure, Sex, and Strength and Knee and Hip Kinematics During Landing. *J Athl Train*. 2011;46(4):376-385.
11. Nguyen A-D, Shultz SJ, Schmitz RJ. Landing Biomechanics in Participants With Different Static Lower Extremity Alignment Profiles. *J Athl Train*. 2015;50(5):498-507. doi:10.4085/1062-6050-49.6.03
12. Kaneko M, Sakuraba K. Association between Femoral Anteversion and Lower Extremity Posture upon Single-leg Landing: Implications for Anterior Cruciate Ligament Injury. *J Phys Ther Sci*. 2013;25(10):1213-1217. doi:10.1589/jpts.25.1213
13. Kozic S, Gulán G, Matovinovic D, Nemec B, Sestan B, Ravlic-Gulan J. Femoral anteversion related to side differences in hip rotation. Passive rotation in 1,140 children aged 8-9 years. *Acta Orthop Scand*. 1997;68:533-536. doi:10.3109/17453679708999021

14. Fan L, Copple TJ, Tritsch AJ, Shultz SJ. Clinical and instrumented measurements of hip laxity and their associations with knee laxity and general joint laxity. *J Athl Train*. 2014;49(5):590-598. doi:10.4085/1062-6050-49.3.86
15. Shultz SJ, Nguyen A, Schmitz R. Differences in Lower Extremity Anatomical and Postural Characteristics in Males and Females Between Maturation Groups. *J Orthop Sport Phys Ther*. 2008;38(3):137-149. doi:10.2519/jospt.2008.2645
16. Gomes JLE, de Castro JV, Becker R. Decreased hip range of motion and noncontact injuries of the anterior cruciate ligament. *Arthroscopy*. 2008;24(9):1034-1037. doi:10.1016/j.arthro.2008.05.012
17. Nyland J, Kuzemchek S, Parks M, Caborn DNM. Femoral anteversion influences vastus medialis and gluteus medius EMG amplitude: composite hip abductor EMG amplitude ratios during isometric combined hip abduction-external rotation. *J Electromyogr Kinesiol*. 2004;14(2):255-261. doi:10.1016/S1050-6411(03)00078-6
18. Radin EL. Biomechanics of the human hip. *Clin Orthop Relat Res*. 1979;October:28-34.
19. Kendall F, McCreary E, Provance P, Rodgers M, Romani W. *Muscles: Testing and Function, with Posture and Pain*. 5th ed. Philadelphia, PA: Lippincott Williams & Wilkins; 2005.
20. Kaneko M, Sakuraba K. Association between Femoral Anteversion and Lower Extremity Posture upon Single-leg Landing: Implications for Anterior Cruciate Ligament Injury. *J Phys Ther Sci*. 2013;25(10):1213-1217. doi:10.1589/jpts.25.1213
21. Sigward SM, Ota S, Powers CM. Predictors of frontal plane knee excursion during a drop

- land in young female soccer players. *J Orthop Sports Phys Ther*. 2008;38(11):661-667.
doi:10.2519/jospt.2008.2695
22. Marx R, Stump T, Jones E, Wickiewicz T, Warren R. Development and evaluation of an activity rating scale for disorders of the knee. *Am J Sport Med*. 2001;29(2):213-218.
 23. Magee D. *Orthopedic Physical Assessment*. Philadelphia, PA: W B Saunders; 1997.
 24. Ruwe. Clinical Determination of Femoral Anteversion. *J Bone Jt Surg*. 1992:820-830.
 25. Hogg J, Schmitz R, Nguyen A-D, Shultz SJ. A Comparison of Passive Hip Range of Motion Values Across Sex and Sport. *J Athl Train*. 2018;53(6):560-567.
 26. Starkey C, Ryan J. *Orthopedic & Athletic Injury Evaluation Handbook*. Philadelphia, PA: F.A. Davis; 2003.
 27. Krause DA, Schlagel SJ, Stember BM, Zoetewey JE, Hollman JH. Influence of Lever Arm and Stabilization on Measures of Hip Abduction and Adduction Torque Obtained by Hand-Held Dynamometry. *Arch Phys Med Rehabil*. 2007;88(1):37-42.
doi:10.1016/j.apmr.2006.09.011
 28. Bell AL, Brand RA, Pedersen R. Prediction of hip joint centre location from external landmarks. *Hum Mov Sci*. 1989;8:3-16.
 29. Jacobs CA, Uhl TL, Mattacola CG, Shapiro R, Rayens WS. Hip abductor function and lower extremity landing kinematics: sex differences. *J Athl Train*. 2007;42(1):76-83.
<http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=1896084&tool=pmcentrez&rendertype=abstract>.

30. Winter D. *Biomechanics and Motor Control of Human Movement*. 2nd ed. New York, NY: John Wiley & Sons, Inc.; 1990.
31. Norcross MF, Johnson ST, Pollard CD, Chang EW, Hoffman MA. Normalization influences knee abduction moment results: Could it influence ACL-injury research, too? *J Sci Med Sport*. 2017;20(4):318-321. doi:10.1016/j.jsams.2016.10.005
32. Dempster W. Space requirements of seated operator. *WACD Tech Rep*. 1955:TR-55-59.
33. Franco N, Bittencourt N, Ribeiro T, et al. Reference values of hip abductor torque among youth athletes: influence of age, sex, and sports. *Phys Ther Sport*. 2015;(2016). doi:10.1016/j.ptsp.2015.12.005
34. Lachowicz M, Preacher K, Kelley K. A novel measure of effect size for mediation analysis. *Psychol Methods*. 2018;23(2):244-261. doi:10.1037/met0000165.1
35. Kelley K. MBESS: The MBESS R Package. 2017.
36. R Core Team. R: A language and environment for statistical computing. 2017. <https://www.r-project.org/>.
37. Fukuda Y, Woo SL-Y, Loh JC, et al. A quantitative analysis of valgus torque on the ACL: a human cadaveric study. *J Orthop Res*. 2003;21(6):1107-1112. doi:10.1016/S0736-0266(03)00084-6
38. Marotta N, Demeco A, de Scorpio G, Indino A, Iona T, Ammendolia A. Late Activation of the Vastus Medialis in Determining the Risk of Anterior Cruciate Ligament Injury in Soccer Players. *J Sport Rehabil*. 2020;29:952-955.

39. Zazulak BT, Hewett TE, Reeves NP, Goldberg B, Cholewicki J. Deficits in neuromuscular control of the trunk predict knee injury risk: a prospective biomechanical-epidemiologic study. *Am J Sports Med.* 2007;35(7):1123-1130.
doi:10.1177/0363546507301585
40. MacKinnon D, Warsi G, Dwyer J. A simulation study of mediated effect measures. *Multivariate Behav Res.* 1995;(30):37-41.

For Peer Review

Figure 1. Measurement Position of Anatomical Variables.

For Peer Review

Figure 2. Terminal Position for the Single-Leg Forward Landing Task.

For Peer Review

Table 1. Descriptive Statistics for Independent Variables in Males and Females.

	Fem. ant. (°)	RoM _{IR} (°)	RoM _{ER} (°)	HEX _{PTQ} (N·m·kg ⁻¹)	HABD _{PTQ} (N·m·kg ⁻¹)	HEX _{%MVIC} (% MVIC)	HABD _{%MVIC} (% MVIC)
Females <i>N</i> = 45	9.4 (4.5)	30.4 (10.3)	47.8 (7.6)	.86 (.21)	1.15 (.24)	37.2 (31.4)	42.2 (23.7)
Males <i>N</i> = 42	3.0 (3.5)	19.8 (8.5)	49.6 (6.4)	1.04 (.27)	1.48 (.29)	28.2 (26.2)	52.3 (41.2)
Totals	6.3 (5.1)*	25.3 (10.8)*	48.7 (7.0)	.95 (.26)*	1.31 (.31)*	32.9 (29.2)	47.1 (33.5)

Fem. Ant. = femoral anteversion; RoM_{IR} = internal rotation RoM; RoM_{ER} = external rotation RoM; HEX_{PTQ} = hip extension peak torque during MVIC testing; HABD_{PTQ} = hip abduction peak torque during MVIC testing; HEX_{%MVIC} = hip extension EMG activation amplitude during landing; HABD_{%MVIC} = hip abduction EMG activation amplitude during landing; *significant between-sex difference at $P < .05$.

Table 2. Within-Sex Bivariate Correlations Among Independent Variables.

	Fem. ant.	ROM _{IR}	ROM _{ER}	HEX _{PTQ}	HABD _{PTQ}	HEX _{%MVIC}	HABD _{%MVIC}
Fem. Ant.	--	.76*	-.42*	.15	-.26*	-.00	-.21
ROM _{IR}	.54*	--	-.42*	.27*	-.28*	-.28*	-.39*
ROM _{ER}	-.43*	-.32*	--	.01	.00	.25	.00
HEX _{PTQ}	-.17	-.18	.10	--	.58*	-.33*	-.05
HABD _{PTQ}	.04	-.32*	-.08	.32*	--	-.15	-.22
HEX _{%MVIC}	-.21	-.03	.05	.14	-.22	--	.53*
HABD _{%MVIC}	.24	.33*	-.18	.07	-.44*	.38*	--

Bolded values signify female correlations; unbolded values signify male correlations; Fem. Ant. = femoral anteversion; ROM_{IR} = internal rotation ROM; ROM_{ER} = external rotation ROM; HEX_{PTQ} = hip extension peak torque; HABD_{PTQ} = hip abduction peak torque; HEX_{%MVIC} = hip extension activation amplitude; HABD_{%MVIC} = hip abduction activation amplitude; *significant correlation at $p < .05$

Table 3. Descriptive Statistics (mean(SD)) for Dependent Biomechanical Variables in Males and Females.

	Males	Females	Total
Peak Hip Adduction Angle (°)	-1.81±7.19	0.89±5.41	-0.41±6.44
Peak Hip Adduction Joint Moment (Nm*kg ⁻¹ *m ⁻¹)	0.145±0.047	0.16±0.070	0.154±0.060
Peak Hip Internal Rotation Angle(°)	-1.28±4.91	1.06±6.21	-0.07±5.72
Peak Hip Internal Rotation Joint Moment(Nm*kg ⁻¹ *m ⁻¹)	0.010±0.016	0.012±0.017	0.011±0.017
Peak Knee Abduction Angle(°)	0.51±6.62	-3.05±5.70	-1.33±6.38
Peak Knee Abduction Joint Moment(Nm*kg ⁻¹ *m ⁻¹)	-0.015±0.028	-0.018±0.031	-0.017±0.029
Peak Knee Internal Rotation Angle(°)	13.15±8.86	11.96±12.67	12.54±10.95
Peak Knee Internal Rotation Joint Moment(Nm*kg ⁻¹ *m ⁻¹)	0.025±0.016	0.020±0.015	0.022±0.015

Joint moments are reported as external moments.

Table 4. Final GLM Forward Stepwise Regression Models Detailing the Influence of Control Variables, Anatomical Variables, and Neuromuscular Variables on peak hip joint angles and external moments.

Dependent variable	Step	Sex	Part correlations										
			P value		Control variables		Anatomical variables			Neuromuscular variables			
			R ²	R ² change	Peak knee flexion	Peak hip flexion	Fem. ant.	ROM IR	ROM ER	HEX PTQ	HABD PTQ	HEX %MVIC	HABD %MVIC
Peak hip add angle (+)	1	F	.18 (.02)		-.22	.37							
	2		.18 (.02)	--	-.22	.37	--	--	--				
	3		.28 (.003)	.10 (.02)	-.17	.42	--	--	--	--	-.32	--	--
	1	M	.00 (.99)		.00	-.02							
	2		.25 (.03)	.25 (.01)	.01	-.08	--	.41	.40				
	3		.25 (.03)	--	.01	-.08	--	.41	.40	--	--	--	--
Peak hip add mom (+)	1	F	.24 (.003)		.01	.49							
	2		.31 (.002)	.06 (.06)	-.01	.50	--	.25	--				
	3		.33 (.002)	.03 (.20)	-.04	.46	--	.29	--	--	.17	--	--
	1	M	.02 (.67)		-.12	-.06							
	2		.09 (.29)	.07 (.09)	-.16	-.05	--	--	.27				
	3		.16 (.15)	.07 (.09)	-.12	-.03	--	--	.29	-.26	--	--	--
Peak hip IR angle (+)	1	F	.11 (.10)		-.04	.32							
	2		.23 (.01)	.12 (.02)	-.07	.33	--	.35	--				
	3		.23 (.01)	--	-.07	.33	--	.35	--	--	--	--	--
	1	M	.05 (.36)		-.20	-.08							
	2		.24 (.04)	.19 (.01)	-.09	-.14	.21	--	-.26				
	3		.39 (.01)	.15 (.04)	-.17	.01	.21	--	-.34	--	-.27	--	-.38
Peak hip IR mom (+)	1	F	.00 (.94)		.03	.04							
	2		.13 (.14)	.12 (.02)	-.01	-.04	--	--	-.35				
	3		.18 (.08)	.06 (.10)	-.05	-.04	--	--	-.36	.24	--	--	--
	1	M	.02 (.71)		.10	.08							
	2		.17 (.06)	.16 (.01)	.15	-.01	--	.39	--				
	3		.29 (.02)	.11 (.05)	.09	.04	--	.42	--	.29	--	-.22	--

p entry<.20; Add = adduction; Mom = moment; Fem. Ant. = Femoral Anteversion; ROMIR = internal rotation range of motion; ROMER = external rotation range of motion; HEXPTQ = hip extension peak torque; HABDPTQ = hip abduction peak torque; HEX%MVIC = hip extension EMG activation amplitude; HABD%MVIC = hip abduction EMG activation amplitude; Control variables were entered on first step of regression; anatomical variables were entered second; neuromuscular variables were entered third

Table 5. Final GLM Forward Stepwise Regression Models Detailing the Influence of Control Variables, Anatomical Variables, and Neuromuscular Variables on peak knee joint angles and external moments.

Dependent variable	Step	Sex	P value		Part correlations								
			R ²	R ² change	Control variables		Anatomical variables			Neuromuscular variables			
					Peak knee flexion	Peak hip flexion	Fem. ant.	ROM IR	ROM ER	HEX PTQ	HABD PTQ	HEX %MVIC	HABD %MVIC
Peak knee abd angle (-)	1	F	.17 (.02)		.32	-.26							
	2		.21 (.02)	.04 (.15)	.36	-.26	-.20	--	--				
	3		.21 (.02)	--	.36	-.26	-.20	--	--	--	--	--	--
	1	M	.00 (1.00)		-.02	.00							
	2		.26 (.02)	.26 (.01)	-.03	.07	--	-.46	-.35				
	3		.26 (.02)	--	-.03	.07	--	-.46	-.35	--	--	--	--
Peak knee abd mom (-)	1	F	.02 (.72)		-.02	-.12							
	2		.02 (.72)	--	-.02	-.12	--	--	--				
	3		.07 (.36)	.06 (.11)	.02	-.12	--	--	--	-.24	--	--	--
	1	M	.04 (.42)		-.11	.19							
	2		.17 (.13)	.13 (.06)	-.11	.23	--	-.29	-.29				
	3		.28 (.03)	.11 (.03)	-.17	.21	--	-.39	-.34	--	-.32	--	--
Peak knee IR angle (+)	1	F	.12 (.07)		.34	-.05							
	2		.28 (.02)	.16 (.04)	.40	.04	-.18	.31	.31				
	3		.42 (.001)	.14 (.01)	.44	.04	-.25	.42	.27	--	--	.37	--
	1	M	.24 (.01)		.42	.18							
	2		.27 (.01)	.03 (.02)	.39	.19	--	--	.18				
	3		.39 (.001)	.18 (.01)	.44	.00	--	--	.25	--	--	--	.43
Peak knee IR mom (+)	1	F	.12 (.07)		.32	-.13							
	2		.12 (.07)	--	.32	-.13	--	--	--				
	3		.12 (.07)	--	.32	-.13	--	--	--	--	--	--	--
	1	M	.05 (.35)		.18	.11							
	2		.05 (.35)	--	.18	.11	--	--	--				
	3		.05 (.35)	--	.18	.11	--	--	--	--	--	--	--

p entry < .20; Abd = abduction; Mom = moment; Fem. Ant. = femoral anteversion; ROMIR = internal rotation range of motion; ROMER = external rotation range of motion; HEXPTQ = hip extension peak torque; HABDPTQ = hip abduction peak torque; HEX%MVIC = hip extension activation amplitude; HABD%MVIC = hip abduction activation amplitude; Control variables were entered on first step of regression; anatomical variables were entered second; neuromuscular variables were entered third

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Figure 1. Measurement of Anatomical Variables.

54x42mm (220 x 220 DPI)

Figure 2. Terminal Position for the Single-Leg Forward Landing Task.
28x76mm (220 x 220 DPI)