*Journal of Athletic Training* 2012;47(4):406–413
doi: 10.4085/1062-6050-47.4.17

www.nata.org/journal-of-athletic-training

*original research*

# Lower Extremity Muscle Activation and Knee Flexion During a Jump-Landing Task

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***Context:*** Decreased sagittal-plane motion at the knee during dynamic tasks has been reported to increase impact forces during landing, potentially leading to knee injuries such as anterior cruciate ligament rupture.

***Objective:*** To describe the relationship between lower extremity muscle activity and knee-flexion angle during a jump-landing task.

***Design:*** Cross-sectional study.

***Setting:*** Research laboratory.

***Patients or Other Participants:*** Thirty recreationally active volunteers (15 men, 15 women: age = 21.63 ± 2.01 years, height = 173.95 ± 11.88 cm, mass = 72.57 ± 14.25 kg).

***Intervention(s):*** Knee-flexion angle and lower extremity muscle activity were collected during 10 trials of a jump-landing task.

***Main Outcome Measure(s):*** Simple correlation analyses were performed to determine the relationship between each knee-flexion variable (initial contact, peak, and displacement) and electromyographic amplitude of the gluteus maximus (GMAX), quadriceps (VMO and VL), hamstrings, gastrocnemius, and quadriceps : hamstring (Q : H) ratio. Separate forward stepwise multiple regressions were conducted to determine which combination of muscle activity variables predicted each knee-flexion variable.

***Results:*** During preactivation, VMO and GMAX activity and the Q : H ratio were negatively correlated with knee-flexion angle at initial contact (VMO: $r = -0.382$, $P = .045$; GMAX: $r = -0.385$, $P = .043$; Q : H ratio: $r = -0.442$, $P = .018$). The VMO, VL, and GMAX deceleration values were negatively correlated with peak knee-flexion angle (VMO: $r = -0.687$, $P = .001$; VL: $r = -0.467$, $P = .011$; GMAX: $r = -0.386$, $P = .043$). The VMO and VL deceleration values were negatively correlated with knee-flexion displacement (VMO: $r = -0.631$, $P = .001$; VL: $r = -0.453$, $P = .014$). The Q : H ratio and GM activity predicted 34.7% of the variance in knee-flexion angle at initial contact ($P = .006$). The VMO activity predicted 47.1% of the variance in peak knee-flexion angle ($P = .001$). The VMO and VL activity predicted 49.5% of the variance in knee-flexion displacement ($P = .001$).

***Conclusions:*** Greater quadriceps and GMAX activation and less hamstrings and gastrocnemius activation were correlated with smaller knee-flexion angles. This landing strategy may predispose an individual to increased impact forces due to the negative influence on knee-flexion position.

***Key Words:*** knee injuries, anterior cruciate ligament, biomechanics

**Key Points**

- Small knee-flexion angles were largely influenced by high quadriceps:hamstrings (Q:H) co-activation ratios during the preparatory phase.
- The high Q:H co-activation ratios were largely the result of diminished hamstrings activity rather than excessive quadriceps activity.
- Interventions designed to enhance preparatory hamstrings activity may be helpful in balancing Q:H co-activation ratios and placing the knee in a more flexed position at initial contact, which may reduce anterior cruciate loading and injury risk.

Sports medicine researchers have extensively examined lower extremity mechanics during athletic movements with the goal of identifying factors that could lead to subsequent knee injuries. Much of the recent literature on knee injuries has focused specifically on determining the lower extremity movement patterns that could predispose an individual to noncontact anterior cruciate ligament (ACL) injury. Analyses of videos recorded during noncontact ACL injury events have repeatedly shown the body to be in an erect posture (ie, decreased knee, hip, and trunk flexion), which implicates decreased sagittal-plane motion as a potential risk factor for ACL injury.[1–3]

An increasing amount of evidence suggests that decreased sagittal-plane motion at the knee during jumping, landing, and cutting may contribute to ACL injuries from the influence of knee-flexion angle on impact forces.[4–7] Greater ground reaction forces[6] and proximal anterior tibial shear force[4,5,7] are present when individuals land with a more extended knee compared with a more flexed knee.

These findings suggest that knee-flexion angle may influence ACL injury risk by modifying the magnitude of ACL loading. Although decreased knee-flexion angles have been found to influence impact forces, researchers have yet to determine which neuromuscular characteristics influence knee-flexion angle during landing tasks. This information could improve our understanding of ACL injury mechanisms.

Activation of the muscles controlling sagittal-plane motion at the knee and hip is likely to influence the amount of knee flexion that occurs during dynamic tasks such as a jump landing. Greater concentric action of the hamstrings and gastrocnemius muscles may result in larger internal knee-flexion moments and position the knee in more flexion during landing. Conversely, greater eccentric action of the quadriceps and gluteus maximus muscles may create larger internal knee- and hip-extension moments, respectively, and facilitate a more erect body posture (less knee and hip flexion).[7–9] The relative activation between agonist and antagonist muscles may also be a factor influencing knee-flexion position.[10,11] Therefore, quadriceps to hamstrings co-activation (Q:H ratio) may be important because these are the primary muscles controlling the net internal knee flexion-extension moment.

Investigators have not yet studied the association between the activation of sagittal-plane muscles and knee-flexion position during a jump-landing task. Therefore, the purpose of our study was to evaluate relationships between lower extremity muscle activity (quadriceps, hamstrings, gluteus maximus, gastrocnemius, and Q:H ratio) and knee-flexion angle during a jump-landing task. We hypothesized that the relationships between quadriceps and gluteus maximus electromyographic (EMG) amplitude and Q:H ratio and knee-flexion angle during a jump-landing task would be negative. Additionally, we hypothesized that the relationship between hamstrings and gastrocnemius EMG amplitude and knee-flexion angle during a jump-landing task would be positive.

## METHODS

### Participants

Fifteen men (age = 22.20 $\pm$ 1.78 years, height = 1.83 $\pm$ 0.06 m, mass = 82.21 $\pm$ 11.91 kg) and 15 women (age = 21.07 $\pm$ 2.12 years, height = 1.65 $\pm$ 0.07 m, mass = 62.93 $\pm$ 8.91 kg) were recruited from a university population to participate in this investigation. We performed a power analysis for a regression model based on the kinematic and EMG data from Malinzak et al,[9] which indicated that 30 participants would be necessary to attain a priori power of 0.80. Inclusion criteria consisted of (1) participation in physical activity for 30 minutes per day at least 3 days per week; (2) current (intramural or club) or former (at least 1 year of high school varsity) participation in organized soccer, volleyball, basketball, or lacrosse; (3) no history of lower extremity injury to either leg within the 6 months before data collection; (4) no history of surgery in the lower extremity within the previous 2 years; and (5) no history of ACL injury. We chose to include only individuals with previous participation in the aforementioned sports because the jump landing is a common task performed during these sports and one with which these individuals should be familiar.

### Instrumentation

A surface EMG system (Bangoli 8, DelSys, Inc, Boston, MA; amplification factor = 1000 [20–450 Hz], common mode rejection ratio at 60 Hz > 80 dB; input impedance > $10^{15}$//0.2 $\Omega$//pF) was used to record the muscle activity of the gluteus maximus (GMAX), vastus medialis oblique (VMO), vastus lateralis (VL), biceps femoris long head (BF), and lateral gastrocnemius (LG) using differential surface electrodes (DelSys, Inc). A force plate (model 4060-08; Bertec Corporation, Columbus, OH) was used for kinetic data collection to identify phases of the jump-landing task. Force-plate and EMG data were sampled at 1200 Hz.

Each participant was outfitted with reflective markers (Vicon, Centennial, CO) to record kinematic data of the lower extremity during the jump-landing task. Motion of the reflective markers was captured by 7 infrared video cameras (DCR-HC38 MiniDV Handycam Camcorders; Sony, New York, NY) at 120 Hz calibrated for a 2.5-m-long $\times$ 1.5-m-wide $\times$ 2.5-m-high space in which the participant performed the jump-landing task. All data were collected using Vicon Nexus Software (version 1.1; Vicon Motion Systems). A global reference system was defined using the right-hand rule, in which the x axis was positive in the anterior direction, the y axis was positive to the left of each participant, and the z axis was positive in the superior direction.

### Testing Procedures

Participants reported to a research laboratory for a single testing session wearing athletic shoes and spandex shorts and shirt. Upon arrival, all participants read and signed an informed consent form approved by The University of North Carolina at Chapel Hill's Institutional Review Board, which also approved the study. Demographic information was collected for each person, and a health questionnaire was used to confirm inclusion and exclusion criteria. Participants then completed a 5-minute warm-up on a stationary cycle ergometer at a self-selected pace of 25% of their perceived maximum at a resistance equivalent to that on his or her perceived flat road (5/10 on a 1–10 scale).

The *dominant leg* was defined as the leg used to kick a ball for maximum distance and was used for kinematic and EMG data collection for each person. For EMG preparation, the skin was shaved, abraded, and cleaned with isopropyl alcohol before surface electrodes were applied. A single differential parallel-bar surface EMG sensor (model DE-2.1; DelSys, Inc: interelectrode distance = 10 mm, nickel-silver sensor material) was affixed to each muscle using a double-sided adhesive skin interface (model DE-3.1 Bangoli; DelSys, Inc). The EMG sensor for the GMAX was placed over the greatest prominence of the middle of the buttocks, midway between the sacral vertebrae and the greater trochanter.[12] The EMG sensor for the BF was placed over the measured midpoint (ischial tuberosity to fibular head) of the muscle belly.[12] The EMG sensors for the quadriceps were placed over the VL and 10 cm superior and 7 cm lateral to the superior border of the patella and oriented at approximately 10° laterally with respect to vertical; those for the VMO were placed approximately 4 cm superior to and 3 cm medial to the superomedial border of the patella and oriented 55° medially with respect to vertical.[13] The EMG sensor for the LG was placed over the muscle belly of the lateral head.[12] Sensor placements and the absence of cross-talk were confirmed by evaluating activity of each muscle with manual muscle tests. Once EMG sensor placements were confirmed, the sensors and leads were secured with prewrap and athletic tape to minimize movement artifact.

Participants were then outfitted with reflective markers placed on the following landmarks: right and left acromial processes, right and left anterior-superior iliac spines, S1 joint space, right and left greater trochanters, anterior aspects of the right and left thighs, right and left lateral and medial femoral epicondyles, anterior aspects of the right and left shanks, right and left lateral and medial malleoli, right and left calcanei, and heads of the right and left first and fifth metatarsals. The markers were affixed to the skin and shoes with adhesive tape. Markers used to represent anatomical landmarks on the foot segment were placed over the participant's shoes in estimated locations. After the markers were placed, the participant was asked to stand in the center of the calibration area with each foot on a force plate for collection of a static calibration trial. After the static calibration trial, the medial malleolus and medial epicondyle markers were removed for data collection during the jump-landing task.

The EMG and kinematic data were collected during 10 trials of a jump-landing task. The task was first described and then demonstrated to the participant. A 30-cm-high box was placed at a distance of 50% of the person's height from the force plate. The participant was instructed to jump down from the box directly onto the force plate, landing with the dominant (test) limb on the force plate and the nondominant (nontest) limb off the force plate, and to then jump vertically for maximum height. The jump-landing task was similar to that previously investigated.[14] A maximum of 5 practice trials was allowed before data collection. A 1-minute rest period between test trials was permitted to avoid fatigue. Each participant completed 10 successful jump-landing trials; a *successful trial* was defined as completing the jump-landing task as described above.

After the data-collection trials, each person performed three 5-second maximum voluntary isometric contractions (MVICs) for the GMAX, quadriceps (VL and VMO), BF, and LG. For the GMAX MVIC, the participant was positioned prone on a table with the knee flexed to 90° and a strap placed over the midbelly of the hamstrings. He or she was instructed to contract isometrically into hip extension. For MVIC of the VMO, VL, and BF, the participant was seated on an isokinetic dynamometer chair with the hips and knees at 90° and straps around the legs and trunk. The person was instructed by the lead investigator to kick into the strap, extending the knee for the quadriceps MVIC, and to kick back into the strap, flexing the knee for the hamstrings MVIC. For the LG MVIC, the participant lay prone on a table with the knees fully extended and the ankle in a neutral sagittal-plane position over the edge of the table. A strap was placed around the metatarsal heads of the foot, and he or she was instructed to push into the strap with maximum force. These testing positions are similar to those in the manual muscle tests described by Hislop and Montgomery.[15]

### Data Reduction

The kinematics, kinetics, and EMG data were time synchronized to 1200 Hz using linear interpolation. The EMG data were passively demeaned, bandpass (10–50 Hz) and notch (59.5–60.5 Hz) filtered, and smoothed using a 20-millisecond root mean square sliding-window function. All data were exported for reduction in a custom MATLAB program (MathWorks, Natick, MA.). Mean EMG amplitudes were calculated during the preparatory and deceleration phases. The *preparatory phase* was defined as 200 milliseconds before *initial contact*, which was defined as the time point at which the vertical ground reaction force exceeded 10 N. The *deceleration phase* was defined as initial contact to the time of peak knee flexion. The EMG data were normalized to the mean amplitude during the middle 3 seconds of each MVIC trial averaged across the 3 MVIC trials. The Q:H ratio was determined separately for the preparatory and deceleration phases by calculating the arithmetic mean of the normalized VMO and VL EMG data and then dividing this number by the normalized BF EMG data.

The 3-dimensional local coordinates of the medial and lateral femoral condyles and medial malleoli were estimated from the 3-dimensional coordinates of markers on the tibia in the standing trial. The *knee joint center* was defined as the middle point between the medial and lateral femur condyles, whereas the *ankle joint center* was defined as the middle point between the medial and lateral malleoli. The 3-dimensional coordinates of the hip joint centers in jump-landing trials were estimated from the 3-dimensional coordinates of the reflective markers on the right and left anterior-superior iliac spines and L4-L5 joint using the method described by Bell et al.[16] The 3-dimensional coordinates of the knee and ankle joint centers and medial and lateral malleoli were used to define the tibia reference frame. The 3-dimensional coordinates of the knee and hip joint centers and medial and lateral femur condyles were used to define the femur reference frame. Knee joint angles were calculated as motion of the shank reference frame relative to the thigh reference frame using Euler angles rotated in flexion-extension, valgus-varus, and internal-external order.

Knee-flexion angle at initial contact, peak knee-flexion angle, and knee-flexion displacement during the deceleration phase of the jump-landing task were calculated for each trial. Knee-flexion displacement was calculated by subtracting the knee-flexion angle at initial contact from the peak knee-flexion angle.

### Statistical Analysis

All dependent variables were averaged over the 10 trials of the jump-landing task and used for data analysis. Analyses were performed for men and women separately and combined. Relationships between EMG amplitudes and knee kinematics during the jump-landing task were first assessed using a series of simple correlational analyses. Six separate analyses were conducted for each muscle: 1 for each kinematic variable (knee-flexion angle at initial contact, peak knee-flexion angle, and knee-flexion displacement) during each phase of the task (preparatory and deceleration). The variables that demonstrated significant simple correlations ($P < .05$) were then entered into separate forward stepwise multiple linear regression analyses, using the enter method, to determine the collective influence of preparatory EMG on initial contact knee-flexion angle, deceleration EMG on peak knee-flexion angle, and deceleration EMG on knee-flexion displacement. The EMG variables were entered into the regression model in order of largest to smallest *r* value for the significant simple correlations. Collinearity tests were also performed to determine if the correlation among the predictor variables (ie, correlation between EMG of the VMO and VL) may have negatively influenced each regression model via variable suppression. We examined the variance inflation factor for each regression model to assess for multicollinearity. A variance inflation factor greater than 10.0 is thought to indicate multicollinearity within a regression model.[17] The a priori $\alpha$ level was set at .05. All analyses were performed using SPSS (version 13.0; SPSS Inc, Chicago, IL).

## RESULTS

To substantiate combining males and females for our analyses, separate 1-way analyses of variance were performed to determine if there were sex differences in the kinematic and EMG variables. No sex differences were found for any of the variables (all $P$ values $>$ .05), and we therefore performed analyses on men and women combined. For 6 participants, 1 trial of the 10 test trials was not included in the statistical analysis because of outliers in the EMG or kinematic data (an *outlier* was defined as a value greater than 3 standard deviations beyond the mean). If 1 trial was removed, we averaged the data across 9 trials. The means, standard deviations, and 95% confidence intervals for each dependent variable and the number of participants included in the analyses with the respective variables are presented in Table 1.

**Table 1. Lower Extremity Muscle Activity and Knee-Flexion Angle by Phase**

| Variable | n | Percentage of Maximal Voluntary Isometric Contraction, Mean ± SD (95% Confidence Interval) |
|---|---|---|
| Muscle | | |
| Gluteus maximus | 29 | |
| Preparatory phase | | 0.14 ± 0.11 (0.10, 0.19) |
| Deceleration phase | | 1.36 ± 1.15 (0.92, 1.80) |
| Vastus medialis oblique | 29 | |
| Preparatory phase | | 0.50 ± 0.33 (0.37, 0.62) |
| Deceleration phase | | 2.24 ± 0.84 (1.92, 2.57) |
| Vastus lateralis | 30 | |
| Preparatory phase | | 0.28 ± 0.19 (0.21, 0.35) |
| Deceleration phase | | 2.48 ± 1.34 (1.98, 2.98) |
| Biceps femoris long head | 29 | |
| Preparatory phase | | 0.15 ± 0.09 (0.114, 0.18) |
| Deceleration phase | | 1.115 ± 0.91 (0.811, 1.50) |
| Lateral gastrocnemius | 30 | |
| Preparatory phase | | 0.30 ± 0.18 (0.23, 0.36) |
| Deceleration phase | | 1.04 ± 0.56 (0.83, 1.25) |
| Quadriceps : hamstrings ratio | 29 | |
| Preparatory phase | | 3.49 ± 2.55 (2.52, 4.46) |
| Deceleration phase | | 3.04 ± 2.08 (2.25, 3.83) |
| Knee-flexion angle, ° | 29 | |
| Initial contact phase | | 20.42 ± 8.60 (17.15, 23.70) |
| Deceleration phase (peak) | | 87.42 ± 18.29 (80.46, 94.38) |
| Deceleration phase (displacement) | | 66.99 ± 17.79 (60.23, 73.76) |

### Simple Correlational Analyses

During the preparatory phase, EMG activity of the VMO and GMAX and the Q:H ratio had significant negative relationships with knee-flexion angle at initial contact (VMO: $r = -0.38$, $P = .04$; GMAX: $r = -0.38$, $P = .04$; Q:H ratio: $r = -0.44$, $P = .01$; Table 2). During the deceleration phase, EMG activity of the VMO, VL, and GMAX had significant negative relationships with peak knee-flexion angle (VMO: $r = -0.68$, $P < .01$; VL: $r = -0.46$, $P = .01$; GMAX: $r = -0.38$, $P = .04$; Table 2). Also, during the deceleration phase, EMG activity of the VMO and VL demonstrated significant negative relationships with knee-flexion displacement (VMO: $r = -0.63$, $P < .01$; VL: $r = -0.45$, $P = .01$; Table 2). No significant correlations were found for BF or LG EMG during the preparatory or deceleration phases.

### Multiple Regression Analyses

For all regression models, the variance inflation factor never exceeded 2.0, and therefore we are confident that multicollinearity did not influence the results. Regression model coefficients are presented in Table 3. To determine the collective influence of lower extremity muscle activation during the preparatory phase on initial contact knee-flexion angle, the following variables were entered in the stated order: Q:H ratio, GMAX, VMO. The Q:H ratio and EMG activity of the GMAX during the preparatory phase predicted approximately 34.7% of the variance in initial contact knee-flexion angle ($R^2 = 0.34$, $P < .01$) (Knee-flexion angle at initial contact = 30.28 + –1.41*Q:H ratio + –27.11*GMAX preparatory). The inclusion of the VMO activity ($R^2$ change = 0.002, $P = .77$) in this model did not cause a significant change in the $R^2$ value, so it was not included in the final model. To determine the collective influence of lower extremity muscle activation during the deceleration phase on peak knee-flexion angle, the following variables were entered in the stated order: VMO, VL, GMAX. The EMG activity of the VMO during the deceleration phase predicted approximately 47.1% of the variance in peak knee-flexion angle ($R^2 = 0.47$, $P < .01$; peak knee flexion = 121.75 + –14.95*VMO deceleration). The GMAX ($R^2$ change = 0.01, $P = .36$) and VL activity ($R^2$ change = 0.02, $P = .31$) did not significantly change the $R^2$ value in the model and therefore were not included in the final model. Finally, to determine the collective influence of lower extremity muscle activation during the deceleration phase on knee-flexion displacement, the following variables were entered in the stated order: VMO, VL. The EMG activity of the VMO during the deceleration phase predicted approximately 39.9% of the variance in knee-flexion displacement ($R^2 = 0.39$, $P < .01$; knee-flexion displacement = 97.57 + –13.42*VMO deceleration). The VL activity ($R^2$ change = 0.035, $P = .22$) did not significantly change the $R^2$ value in the model and therefore was not included in the final model.

**Table 2. Correlations of Muscle Activity and Knee-Flexion Angles**

| Muscle | Knee-Flexion Angle at Initial Contact[a] | | Peak Knee-Flexion Angle[b] | | Knee-Flexion Displacement[b] | |
|---|---|---|---|---|---|---|
| | *r* Value | *P* Value | *r* Value | *P* Value | *r* Value | *P* Value |
| Gluteus maximus | −0.38[c] | .04 | −0.38[c] | .04 | −0.23 | .22 |
| Vastus medialis oblique | −0.38[c] | .04 | −0.68[c] | .001 | −0.63[c] | .001 |
| Vastus lateralis | −0.19 | .32 | −0.46[c] | .01 | −0.45[c] | .01 |
| Biceps femoris long head | 0.21 | .27 | −0.11 | .56 | −0.17 | .36 |
| Lateral gastrocnemius | −0.31 | .09 | −0.16 | .40 | −0.19 | .32 |
| Quadriceps:hamstrings ratio | −0.44[c] | .01 | −0.36 | .05 | −0.27 | .16 |

[a] Correlated with muscle activity during the preparatory phase.
[b] Correlated with muscle activity during the deceleration phase.
[c] Correlation significant at $<.05$.

## DISCUSSION

Our most important findings were that preparatory Q:H co-activation ratio and GMAX activity were significant predictors of knee-flexion angle at initial contact; however, neither Q:H co-activation ratio nor GMAX activity were indicative of knee-flexion kinematics during the deceleration phase. These findings suggest that different neuromuscular characteristics influence knee-flexion kinematics at initial contact and over the deceleration phase of a jump-landing task. Contrary to our original hypothesis, we did not observe a significant relationship between sagittal-plane knee kinematics and either BF or LG muscle activation.

Both Q:H co-activation ratio and GMAX activation were negatively correlated with knee-flexion angle at initial contact, explaining approximately 35% of the variance in knee-flexion angle at initial contact. These findings indicate that a greater Q:H co-activation ratio combined with greater GMAX activity facilitates a more extended knee position at initial contact. Although we examined only knee-flexion angle, it is likely that individuals landing with a decreased knee-flexion angle had a more erect posture. Previous authors[7,8] reported that participants landing with decreased knee-flexion angles also had decreased hip flexion angles. We know that individuals who land with a smaller knee-flexion angle also have greater quadriceps activation.[8,9] However, earlier researchers did not specifically explore the relationship between knee-flexion angle and muscle activation patterns.

Isolated activation of the VMO, VL, BF, and LG did not explain additional variance in knee flexion at initial contact beyond that explained by the Q:H co-activation ratio and GMAX activation. The relative balance between quadriceps and hamstrings activation appears to have the greatest influence on knee flexion at initial contact. This is evidenced by the fact that Q:H co-activation and GMAX activation explained 21% and 13% of the variability, respectively, in knee flexion at initial contact. Although our results show that Q:H ratio and GMAX activation both play a role in modifying knee flexion at initial contact, the relative contribution of Q:H co-activation is higher. The GMAX has been reported to function like the gluteus medius during dynamic tasks[18] and therefore may assist in controlling frontal- and transverse-plane motion at the hip. Knee valgus, a biomechanical factor contributing to ACL loading,[19] results from hip internal rotation and adduction in the closed kinematic chain.[20] Interventions aimed at decreasing GMAX activation may help facilitate greater knee flexion at initial contact; however, this may concomitantly compromise hip rotation stability, allowing for greater hip internal-rotation and knee-valgus motion, which may increase ACL loading.

There appears to be great room for improving the relative balance between quadriceps and hamstrings activation because the average preparatory Q:H co-activation ratio was 3.4 in the current study. The quadriceps were approximately 3.4 times more active than the hamstrings, suggesting a quadriceps-dominant muscle activation strategy in preparation for ground contact. Increased knee flexion at initial contact should occur by promoting balanced activation between the quadriceps and hamstrings, thus reducing Q:H co-activation ratio to near 1.0. Interventions able to successfully decrease Q:H co-activation ratios while maintaining GMAX activation during the preparatory phase may allow for greater knee flexion at initial contact without compromising hip rotation stability. Neuromuscular training programs increase hamstrings activity during dynamic tasks, suggesting that modifying preparatory hamstrings activity may be helpful in future ACL injury-prevention programs.[21,22]

Although the Q:H co-activation ratio had an influence on knee-flexion angle before initial contact, knee-flexion kinematics after initial contact were primarily influenced by quadriceps muscle activation. The VMO and VL activation demonstrated the strongest simple correlations with knee-flexion peak and displacement values. The negative relationships observed between VMO and VL activation and knee-flexion peak and displacement values indicate that increased quadriceps activity, rather than a

**Table 3. Regression Model Coefficients**

| Dependent Variable | Predictor Variable | Unstandardized Coefficients | | Standardized Coefficients | | | | |
|---|---|---|---|---|---|---|---|---|
| | | Beta | SE | Beta | *P* Value | $R^2$ | Model *P* Value | n |
| Knee flexion at initial contact | Quadriceps:hamstrings, preparatory phase | −1.41 | 0.56 | −0.41 | .02 | 0.35 | .006 | 27 |
| | Gluteus maximus, preparatory phase | −27.11 | 12.21 | −0.36 | .03 | | | |
| Peak knee flexion | Vastus medialis oblique, deceleration phase | −14.95 | 3.16 | −0.68 | $<.01$ | 0.47 | $<.01$ | 28 |
| Knee-flexion displacement | Vastus medialis oblique, deceleration phase | −13.42 | 3.23 | −0.63 | $<.01$ | 0.39 | $<.01$ | 28 |

lack of knee flexor activity (ie, hamstrings and gastrocnemius), was a major factor limiting knee-flexion motion during the deceleration phase of the jump-landing task. These findings provide further support for the role of quadriceps dominance as a limiting factor for knee-flexion motion.

The Q:H co-activation was significantly related to knee flexion at initial contact, but no such relationship was seen with knee-flexion peak or displacement values. These results suggest that knee-flexion motion after initial contact is more influenced by the activation of the quadriceps musculature than the activation of the knee-flexor muscles (eg, hamstrings and gastrocnemius). More specifically, preparatory-phase quadriceps dominance may occur through diminished hamstrings activation, which produces greater Q:H co-activation ratios. Deceleration-phase quadriceps dominance appears to be influenced primarily by increased recruitment of the quadriceps musculature instead of Q:H co-activation. Using a quadriceps-dominant muscle activation strategy is a hypothesized risk factor for noncontact ACL injury.[11,23] Therefore, designers of ACL injury-prevention programs should identify successful strategies that could minimize quadriceps dominance during landing tasks by increasing recruitment of the hamstrings musculature.

Additional theorized risk factors for ACL injury include increased proximal anterior tibial shear force, knee valgus moment, and knee rotational moments. Each of these is theorized to be magnified when the knee-flexion angle is reduced.[19,23] Thus, increased preparatory-phase Q:H co-activation ratio and deceleration phase quadriceps activity may contribute to greater ACL loading and injury risk through their influences on knee-flexion position. This is further evidenced by the fact that noncontact ACL injuries are frequently described as occurring with the knee in a more extended position.[1–3,14]

These findings suggest that ACL injury-prevention programs should attempt to increase preparatory hamstrings activity, resulting in a more flexed knee position throughout motion. In addition to our results, authors[24] of a recent meta-analysis of neuromuscular intervention programs reported that the incidence of ACL injuries was reduced after injury-prevention programs that focused on teaching individuals to land in a more bent-knee posture. This more flexed position would likely reduce ACL loading by limiting the anterior tibial shear force imparted by quadriceps action and enhancing the posterior shear force imparted by hamstrings action.[25,26] Furthermore, a more flexed knee position at initial contact would minimize the potentially negative effects of heightened quadriceps activity during the deceleration phase of landing.[25,26]

We acknowledge that the current study has several limitations. First, the participants tested were recreationally active individuals, not necessarily athletes; therefore, our conclusions cannot be applied directly to athletes (the individuals at greatest risk of ACL injury) or to the subset of jumping athletes. Also, the jump-landing task was performed in a controlled laboratory setting, and as such, it may not mimic real-world events. We further acknowledge that placing reflective markers over the shoes as well as having participants perform the jumping tasks in shoes may be a limitation to this investigation; however, we were only interested in knee-flexion kinematics and do not feel that these procedures invalidate our findings. In addition, the calculation of Q:H ratio was based on muscle activity for 2 quadriceps muscles and 1 hamstrings muscle. Future researchers should attempt to analyze muscle activation for the 4 muscles that constitute the quadriceps musculature and the 3 muscles that constitute the hamstrings musculature to provide a true representation of Q:H ratio. Another limitation of our study was our method of collecting MVIC data, as it does not directly mimic the actions of the muscles throughout our task. Future investigators should attempt to collect MVIC data in a manner more consistent with the testing activity. Another limitation of our investigation is that all participants performed the testing procedures in the same order, which may have influenced our results by introducing an order bias. One final limitation is that EMG activity levels are not a measurement of muscle forces being produced; thus, they cannot be used to directly infer the magnitude of ACL loading.

We also acknowledge the lack of sex differences in our independent and dependent variables. Although some research supports sex differences in lower extremity kinematics and muscle activation during dynamic tasks, other research[7–9,27] contradicts these findings. Authors[27] have theorized that differences in participant populations across studies may help to explain the inconsistencies in the findings of sex differences in kinematics and muscle activation during dynamic tasks. Furthermore, the lack of sex differences may also be supported by the theory that only a percentage of females display kinematics and muscle activation strategies that are different from males.[27] Future studies should investigate the relationships between muscle activation and knee-flexion position in the specific group of females who display the altered kinematics and muscle activation patterns.

Although there were no sex differences among our outcome variables, we decided to perform a preliminary sex-specific analysis to help guide future research in this area. Our study was underpowered for performing the additional regression analyses separately for men and women, so we caution the interpretation of the results presented in Tables 4 and 5. During the preparatory phase, EMG activity of the VMO predicted approximately 39.3% of the variance in initial contact knee-flexion angle in women ($R^2 = 0.39$, $P = .02$). In men, the Q:H ratio during the preparatory phase predicted approximately 35.8% of the

**Table 4. Regression Model Coefficients (Women Only)**

| Dependent Variable | Predictor Variable | Unstandardized Coefficients | | Standardized Coefficients | | | Model | |
|---|---|---|---|---|---|---|---|---|
| | | Beta | SE | Beta | *P* Value | $R^2$ | *P* Value | n |
| Knee flexion at initial contact | Vastus medialis oblique, preparatory phase | −9.31 | 3.34 | −0.63 | .02 | 0.39 | .02 | 14 |
| Peak knee flexion | Vastus medialis oblique, deceleration phase | −18.99 | 4.67 | −0.77 | <.01 | 0.60 | <.01 | 13 |
| Knee-flexion displacement | Vastus medialis oblique, deceleration phase | −16.37 | 4.10 | −0.76 | <.01 | 0.57 | <.01 | 14 |

**Table 5. Regression Model Coefficients (Men Only)**

| Dependent Variable | Predictor Variable | Unstandardized Coefficients | | Standardized Coefficients | | | Model | |
|---|---|---|---|---|---|---|---|---|
| | | Beta | SE | Beta | *P* Value | $R^2$ | *P* Value | n |
| Knee flexion at initial contact | Quadriceps:hamstrings, preparatory phase | −2.71 | 1.09 | −0.60 | .03 | 0.36 | .03 | 13 |
| Peak knee flexion | Vastus medialis oblique, deceleration phase | −15.53 | 4.70 | −0.71 | <.01 | 0.50 | <.01 | 13 |

variance in initial contact knee-flexion angle ($R^2 = 0.36$, $P = .03$). During the deceleration phase, EMG activity of the VMO predicted approximately 60.0% of the variance in peak knee-flexion angle in women ($R^2 = 0.60$, $P < .01$). Similar to the women, EMG activity of the VMO during the deceleration phase predicted approximately 49.8% of the variance in peak knee-flexion angle in the men ($R^2 = 0.50$, $P < .01$). The final regression model in women revealed that VMO activation during the deceleration phase predicted approximately 57.1% of the variance in knee-flexion displacement ($R^2 = 0.57$, $P < .01$). A regression analysis predicting knee-flexion displacement in men was not performed because of the lack of significant correlations with lower extremity EMG activity knee-flexion displacement in this group.

Based on these results, knee-flexion angle at initial contact is not influenced by the same muscle activation characteristics in men and women. In women, a quadriceps-dominant strategy influenced knee-flexion angle at initial contact, whereas Q:H co-activation influenced knee-flexion angle at initial contact in men. As men and women continued into the deceleration phase, quadriceps dominance seemed to have the largest influence on knee-flexion kinematics. Future researchers should determine if these results hold true in a larger sample of men and women. However, as stated previously, based on our analyses independent of sex, ACL injury-prevention programs should focus on increasing hamstrings recruitment to minimize a quadriceps-dominant landing strategy. In addition, the moderate $r^2$ values we report suggest that factors other than muscle activation are influencing knee-flexion kinematics during a landing task. Therefore, future study is warranted to determine the additional neuromuscular factors influencing sagittal-plane landing kinematics in order to develop the most effective strategies to increase knee-flexion angle with the use of an ACL injury-prevention program.

## CONCLUSIONS

Improved understanding of the neuromuscular factors that influence knee-flexion angle may allow for more effective ACL injury-prevention interventions. Although our results indicate that different neuromuscular factors influence knee-flexion angle at initial contact and during the deceleration phase, quadriceps dominance consistently influenced individuals with limited knee flexion. Small knee-flexion angles and quadriceps dominance during the preparatory phase were largely influenced by high Q:H co-activation. Furthermore, it seems the high Q:H co-activation ratios were largely influenced by diminished hamstrings activity rather than excessive quadriceps activity during the preparatory phase. This notion is supported by supplementary analyses indicating that the Q:H co-activation ratio was significantly and negatively correlated with hamstrings activity ($r = -0.466$, $P = .01$) but was not correlated with quadriceps activity ($P > .05$ for VL, VMO, and VL/VMO mean). These findings suggest that interventions designed to enhance preparatory hamstrings activity may be beneficial in minimizing the Q:H co-activation ratio and placing the knee in a more flexed position at initial contact, which is consistent with reduced ACL loading and injury risk.[22] Increased quadriceps activation, however, was the primary factor limiting knee-flexion motion and facilitating quadriceps dominance during the deceleration phase. Decreased knee flexion throughout motion can result in greater ACL loading and may place individuals at greater risk for sustaining noncontact ACL injuries.[23] Zebis et al[27] demonstrated that an intervention focused on improving awareness and neuromuscular control was successful in increasing activation of the hamstrings musculature throughout a dynamic task; yet quadriceps activation did not change significantly. Research on ACL injury-prevention programs should continue to investigate strategies to modify preparatory phase Q:H co-activation and decreasing quadriceps activation after ground contact.

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