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Increased Hip and Knee Flexion During Landing Decreases Tibiofemoral Compressive Forces in Women Who Have Undergone Anterior Cruciate Ligament Reconstruction

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Background: Those who have undergone anterior cruciate ligament reconstruction (ACLR) have been shown to exhibit increased muscle co-contraction, decreased knee flexion, and elevated tibiofemoral compressive forces. Elevated tibiofemoral compressive forces may be associated with the high risk of developing knee osteoarthritis in this population.

Purpose: To examine whether muscle co-contraction and tibiofemoral compressive forces in women after undergoing ACLR can be reduced through the use of a landing strategy that emphasizes greater hip and knee flexion.

Study Design: Controlled laboratory study.

Methods: Ten female recreational athletes who had previously undergone ACLR participated in this study. Participants performed a single-legged drop-land task before and after a training session that encouraged them to use greater hip and knee flexion during landing. Peak tibiofemoral compressive forces before and after training were estimated using an electromyography (EMG)–driven knee model that incorporated joint kinematics, EMG, and subject-specific muscle volumes and patellar tendon orientation estimated from magnetic resonance imaging. A co-contraction index (CCI) was calculated to quantify the level of co-contraction between knee flexor and extensor muscles.

Results: After training, peak hip and knee flexion as well as hip and knee flexion excursions increased significantly. Additionally, participants demonstrated a significant decrease after training in the areas of muscle co-contraction (CCI [mean ± SD], 0.28 ± 0.10 vs 0.18 ± 0.05; $P < .001$) and peak tibiofemoral compressive force (97.3 ± 8.0 vs 91.3 ± 10.2 $N \cdot kg^{-1}$; $P = .044$).

Conclusion: Increased muscle co-contraction as well as elevated tibiofemoral compressive loads observed in individuals following ACLR can be reduced by using a landing strategy that encourages greater hip and knee flexion.

Clinical Relevance: The findings of the current study provide useful information for the growth of rehabilitation and/or intervention programs aimed to decrease knee joint loading to prevent or delay the development of knee osteoarthritis in those who have undergone ACLR.

Keywords: EMG driven; MRI; muscle volume; patellar tendon orientation; osteoarthritis

Anterior cruciate ligament reconstruction (ACLR) is often performed to reproduce the function of the original ligament and restore functional stability of the knee joint after ACL injury.[24] Despite substantial improvements in surgical techniques and satisfactory outcomes, individuals who have undergone ACLR have been shown to have a higher risk of developing osteoarthritis of the knee.[20,22] To date, the underlying mechanism(s) for the high risk of knee osteoarthritis in persons following ACLR are not fully understood.

In a previous study, Tsai et al[26] reported that women who have undergone ACLR exhibit greater peak compressive forces (quantified using a subject-specific electromyography [EMG]–driven model) when compared with matched controls. The elevated compressive force observed in women who have undergone ACLR was accompanied by

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TABLE 1
Participant Characteristics of the 10 Women Who Have Undergone Unilateral ACLR[a]

Participant	Age, y	Body Mass, kg	Body Height, cm	Time After ACLR, mo	Type of Graft	Side of Surgery
1	25	58	159	43	BPTB	Left
2	24	61	165	60	BPTB	Left
3	26	62	172	56	Allograft	Left
4	21	62	159	14	Allograft	Left
5	26	61	179	14	Allograft	Left
6	28	65	157	40	Allograft	Left
7	28	46	155	22	Allograft	Left
8	24	64	155	52	Allograft	Right
9	23	71	173	14	Allograft	Right
10	28	54	168	47	BPTB	Right
Mean ± SD	25.3 ± 2.4	60.4 ± 6.7	164.2 ± 8.5	36.2 ± 18.5		

[a]ACLR, anterior cruciate ligament reconstruction; BPTB, bone–patellar tendon–bone autograft.

increased muscle co-contraction and decreased knee flexion during the deceleration phase of a drop-land task.[26] Excessive compressive forces may damage articular cartilage and have been associated with the development of osteoarthritis.[3,4,15] As such, it is reasonable to speculate that correcting abnormal neuromuscular strategies after ACLR may be beneficial in reducing knee compressive forces and therefore the risk of early knee osteoarthritis in this population.

Several ACL injury prevention programs have been described in the literature.[10,14] A common component of these programs is an emphasis on increasing hip and knee flexion during landing. Previous investigations have shown that increasing hip and knee flexion during landing reduces ground-impact forces[11,16,18] and is associated with reduced knee valgus moments.[17] Furthermore, encouraging greater hip and knee flexion may serve to unload the knee by reducing the demand of the quadriceps muscles. For example, it has been shown that those who land with greater hip and knee flexion rely more on hip extensors and less on knee extensors to absorb impact forces when compared with those who land with less hip and knee flexion.[21] Whether increasing hip and knee flexion during landing reduces compressive forces at the tibiofemoral joint has yet to be explored.

The purpose of this study was to examine whether the use of a landing technique that emphasizes greater hip and knee flexion decreases tibiofemoral compressive forces in women who have undergone ACLR. We hypothesized that the peak tibiofemoral compressive forces would decrease after a training session that encouraged increased hip and knee flexion during a single-legged drop-land task. We also hypothesized that the decreased compressive forces after training would be accompanied by decreased muscle co-contraction at the knee.

MATERIALS AND METHODS

Participants

Ten recreational female athletes who had previously undergone unilateral ACLR using either an allograft or bone–patellar tendon–bone (BPTB) autograft participated (Table 1). We have previously reported on these same athletes in a separate work.[26] All participants reported that the right leg was their dominant leg (determined as the leg used to kick a ball). Seven of 10 ACL injuries occurred on the nondominant limb, while 3 ACL injuries occurred on the dominant limb (Table 1). Nine of the 10 participants reported that their ACL tear occurred as a result of noncontact mechanisms during sports participation (ie, soccer, basketball, flag football, and dancing). One participant tore her ACL after sustaining a fall during rock climbing.

To be considered for the study, participants must have returned to unrestricted sport activity for at least 6 months and must have been at least 1 year, but no longer than 5 years, after surgery. Participants were excluded from the study if they reported any of the following: (1) current lower extremity injury that resulted in any persistent pain and discomfort at the time of participation; (2) implanted biological devices (eg, pacemakers, cochlear implants, clips) contraindicated for magnetic resonance imaging (MRI) measurements; (3) history of claustrophobia or severe anxiety; (4) pregnancy; (5) concomitant ligament injuries; (6) history of lower extremity injury or surgery on either limb (except that related to the ACL tear and the subsequent reconstructive surgery); and (7) any medical conditions that would impair the participant to perform the tasks described below. Before participation, all procedures were explained to each individual, and informed consent was obtained as approved by the Institutional Review Board of the University of Southern California Health Sciences Campus.

Procedures

Each participant underwent 2 data collection sessions: (1) biomechanical analysis during a single-legged drop-land task (before and after training) and (2) MRI to obtain subject-specific muscle anatomic parameters. Data obtained from both testing sessions were used as input variables to a subject-specific EMG-driven knee model described by Tsai et al[27] to quantify tibiofemoral compressive forces. Briefly, the input variables to the model included MRI-measured muscle volumes and patellar tendon orientation as well as muscle activation and lower extremity joint kinematics (see below for details).

Sagittal and axial MRI scans of each participant's tested leg were obtained using a 3.0-T MRI system (Signa HDx 3.0 T, GE Healthcare, Little Chalfont, United Kingdom). Axial images of the leg (ankle mortise to the iliac crest) were acquired using a spin-echo pulse sequence (repetition time [TR], 2600-3700 ms; echo time [TE], 11.3 ms; slice thickness, 10 mm; matrix, 512 $\times$ 512). Sagittal plane images of the knee were obtained using a spin-echo pulse sequence (TR, 1100 ms; TE, 37 ms; slice thickness, 3 mm; matrix, 512 $\times$ 512). Sagittal plane images were acquired at 0°, 15°, 30°, 45°, and 60° of knee flexion during static partial weightbearing by having participants push against a load of 111 N. This load was provided by a custom-made nonferromagnetic MRI loading device.

For biomechanical testing, muscle activation levels were recorded from the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), semitendinosus (ST), biceps femoris long head (BFL), medial gastrocnemius (MG), and lateral gastrocnemius (LG) using preamplified bipolar-surface EMG electrodes at 1500 Hz (MA300 EMG system, Motion Lab Systems, Baton Rouge, Louisiana). The preamplifiers had a double-differential input design and a signal bandwidth ranging from 20 Hz to 3000 Hz. Before testing, EMG signals from each muscle were obtained while participants performed a series of 3 maximum voluntary isometric contractions. The EMG data obtained from these tests were for normalization purposes.

Kinematic data were recorded at a rate of 250 Hz using an 8-camera motion analysis system (Vicon 612, Oxford Metrics, Oxford, United Kingdom). Ground-reaction forces (GRFs) were collected at a rate of 1500 Hz using an AMTI force plate (AMTI, Watertown, Massachusetts). Reflective markers were attached to the following bony landmarks: distal first toe, first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanters, anterior superior iliac spines, iliac crests, and the L5-S1 junction. Additional noncollinear tracking cluster markers were placed on the heels, lateral shanks, and lateral thighs.

For the single-legged drop-land task, participants started from a single-legged standing position on a platform (height, 25 cm) in front of the force plate. Participants were instructed to land with the tested foot on the force plate and then jump upward as high as possible. Three trials were collected for each participant.

After performing 3 trials of the drop-land task with their self-selected landing strategy, participants were given a short training session that instructed them to perform the drop-land task with greater hip and knee flexion. Specifically, participants were instructed to land with increased flexion of the hip and knee and to land "as softly as possible" to minimize impact forces during landing.[11,16,18] Each participant was given 2 to 3 practice trials to become familiar with the desired landing strategy. Oral instruction to increase hip and knee flexion was given before each practice trial. In addition, participants were instructed to minimize the impact sound when landing on the ground. Visual feedback (video) was provided after each practice trial.

After completing the practice trials, participants were asked to perform 3 single-legged drop-land tasks using the new landing strategy. No feedback was given during data collection. For the purposes of the current study, we were interested in the deceleration phase of the drop-land task, which was defined as the time from initial contact of the foot with the ground (ie, when the vertical GRF exceeded 20 N) to maximum knee flexion.

Data Analysis

MRI Input Variables. The cross-sectional area of each of the quadriceps, hamstring, and gastrocnemius muscles was measured from each axial MRI scan. The muscle volume of each slice was computed by multiplying the cross-sectional area of the muscle by the slice thickness of the image. The sum of the measured muscle volumes from all slices (ie, total muscle volume) was combined with the muscle pennation angle and fiber length[9,29] to calculate the physiological cross-sectional area (PCSA) for each muscle. The PCSA was then multiplied by a specific tension value of 23 ($N \cdot cm^{-2}$)[6,23] to approximate the maximum isometric muscle force for each muscle.

The orientation of the patellar tendon relative to the tibia was estimated from the sagittal MRI scans at each of the 5 knee flexion angles using ImageJ software (National Institutes of Health, Bethesda, Maryland). The orientation of the patellar tendon relative to the tibia was quantified by measuring the angle formed by the patellar tendon and medial tibia plateau. A linear regression line was fit to the 5 data points to estimate the patellar tendon orientation angle from 0° to 150° of knee flexion.

Kinematic and EMG Input Variables. Visual3D software (C-Motion, Rockville, Maryland) was used to compute the segmental kinematics of the tested lower extremity. Raw trajectory data were filtered using a fourth-order zero-lag Butterworth low-pass filter at 6 Hz. Segment mass and center of mass location were approximated from the data of Dempster.[7] Raw EMG signals were band-pass filtered (35-500 Hz), rectified, and smoothed with a 6-Hz low-pass filter. The smoothed EMG data were normalized to the highest EMG value recorded from either the maximum voluntary isometric contractions or the drop-land task. The EMG data were processed using a custom MATLAB program (MathWorks, Natick, Massachusetts).

To quantify the level of co-contraction between the knee flexor and extensor muscles, a co-contraction index (CCI) was calculated for each participant.[1] Specifically, the CCI was calculated as the ratio of the averaged normalized flexor EMG to the averaged normalized extensor EMG multiplied by the averaged normalized EMG of all muscles during the deceleration phase of the drop-land task. Using this convention, a larger CCI value was indicative of greater muscle co-contraction.

Subject-Specific EMG-Driven Knee Model to Quantify Tibiofemoral Compression. SIMM software (MusculoGraphics Inc, Chicago, Illinois) was used to create a generic lower extremity musculoskeletal model.[6] The model included 10 musculotendon actuators: VL, VM, vastus intermedius (VI), RF, ST, semimembranosus (SM), BFL,

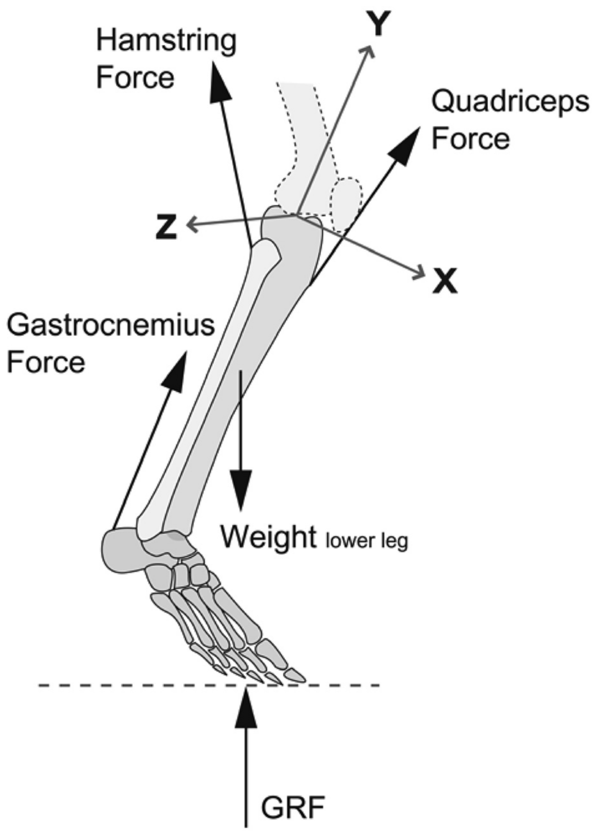

Figure 1. The free body diagram showing the tibial reference frame and the forces used to calculate the tibiofemoral shear and compressive forces. GRF, ground-reaction force. (Reprinted with permission from Tsai LC, McLean S, Colletti PM, Powers CM. Greater muscle co-contraction results in increased tibiofemoral compressive loads following anterior cruciate ligament reconstruction. *J Orthop Res.* 2012; 30(12):2007-2014).

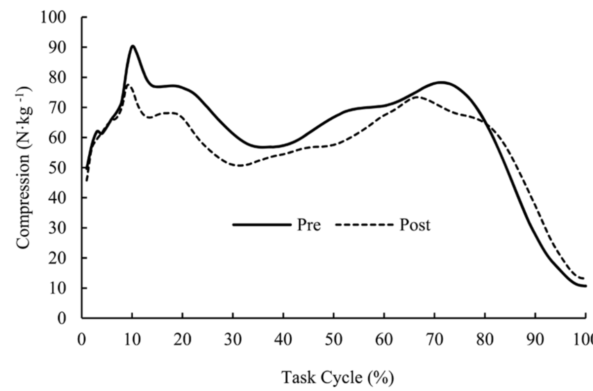

Figure 2. Representative time-series curves of the tibiofemoral compressive forces from 1 participant during the single-legged drop-land task before (solid line) and after (dashed line) the single training session.

biceps femoris short head (BFS), MG, and LG. The anatomic parameters of the 10 muscles were based on the values reported by Friederich and Brand[9] and Wickiewicz et al.[29] Tendon slack lengths of the 10 muscles were based on the average values reported by Delp[6] and Lloyd and Buchanan.[13]

Normalized EMG and lower extremity kinematics (hip flexion/extension, hip adduction/abduction, hip internal/external rotation, knee flexion/extension, ankle plantar flexion/dorsiflexion) were used as input variables for the EMG-driven model. Lower extremity kinematic data were used to determine individual muscle tendon lengths and contraction velocities for the Hill-type muscle model in SIMM. Normalized EMG data were used to represent the level of muscle activation. Muscle activation of the VI was estimated as the average of the VM and VL normalized EMG amplitudes. The SM was assumed to have the same activation as the ST, and the BFS was assumed to have the same activation as the BFL.[12] A 40-millisecond electromechanical delay was used to adjust for the time difference between the onset of EMG signals and onset of force output.[12]

The estimated maximum isometric muscle forces (derived from MRI-estimated PCSA) and patellar tendon orientation (as measured from MRI) were incorporated into the generic SIMM knee model for each participant. The magnitude and orientation of each muscle force were calculated from this subject-specific model. The calculated muscle force vectors were combined with the anthropometry of the lower leg (foot and shank),[7] GRF, and linear acceleration of the lower leg to calculate the tibiofemoral joint force during drop landing (equation 1):

$$\sum \text{Forces} = \text{muscle forces} + \text{GRF} + \text{weight of the leg} + \text{tibiofemoral joint force} = \text{m} \cdot \text{a}, \tag{1}$$

where m is the mass of the lower leg, and a is the linear acceleration vector of the center of mass of the lower leg. The force and acceleration vectors were referenced to the 3-dimensional reference frame located in the tibia (Figure 1). The tibiofemoral compressive force was the component of the joint force vector along the longitudinal axis (defined by the line connecting the knee and ankle joint centers) and was normalized to the body mass ($N{\cdot}kg^{-1}$) for each participant (Figure 2).

Statistical Analysis

To determine whether the targeted landing pattern was achieved after training, paired t tests were performed (Version 15.0 software, SPSS, Chicago, Illinois) to compare the differences in peak hip and knee flexion as well as hip and knee flexion excursions. Paired t tests also were performed to examine whether the peak tibiofemoral compressive force and CCI value decreased after training. In addition, post hoc analyses (ie, paired t tests) were performed to examine the changes in the compressive components of

TABLE 2
Peak Tibiofemoral Compressive Force, Peak and Excursion of Hip and Knee Flexion, and Co-contraction Index During the Deceleration Phase of the Drop-Land Task[a]

	Pretraining	Posttraining	*P* Value	Effect Size
Peak compression, $N \cdot kg^{-1}$	97.3 ± 8.0	91.3 ± 10.2	.044	0.6
Peak hip flexion, deg	63.9 ± 16.2	78.6 ± 14.2	<.001	1.8
Peak knee flexion, deg	67.6 ± 8.8	80.1 ± 8.3	.001	1.4
Hip flexion excursion, deg	30.7 ± 11.5	45.9 ± 10.0	<.001	2.1
Knee flexion excursion, deg	45.4 ± 8.7	57.6 ± 7.8	.001	1.6
Co-contraction index	0.28 ± 0.10	0.18 ± 0.08	<.001	1.9

[a]Values are indicated as mean ± standard deviation.

TABLE 3
Compressive Components of the Ground-Reaction Force and Estimated Muscle Forces That Contributed to the Peak Tibiofemoral Compressive Force as Well as the Averaged Electromyography Values During the Deceleration Phase of the Landing[a]

	Pretraining	Posttraining	*P* Value	Effect Size
Forces at peak compression, $N \cdot kg^{-1}$				
Ground-reaction force	28.5 ± 6.9	26.8 ± 5.6	.213	0.3
Quadriceps	53.2 ± 9.1	48.2 ± 4.7	.099	0.4
Hamstring	3.2 ± 1.7	3.1 ± 1.5	.419	0.1
Gastrocnemius	12.9 ± 3.3	13.9 ± 5.5	.128	0.2
Average normalized electromyography values (% of maximum)				
Quadriceps	36.1 ± 5.0	37.3 ± 5.1	.239	0.2
Hamstring	32.7 ± 12.2	25.7 ± 9.5	.003	1.2
Gastrocnemius	27.5 ± 7.0	21.2 ± 6.7	<.001	2.5

[a]Values are indicated as mean ± standard deviation.

the GRFs and muscle forces that contributed to the peak tibiofemoral compressive force as well as the average normalized quadriceps, hamstring, and gastrocnemius EMG that were used for CCI calculation. All significance levels were set as $P \leq .05$.

RESULTS

Peak knee and hip flexion and hip and knee flexion excursions significantly increased after training (Table 2). After training, the peak tibiofemoral compressive force significantly decreased (97.3 ± 8.0 vs 91.3 ± 10.2 $N \cdot kg^{-1}$; $P = .044$) (Table 2). The decreased peak compressive force after training was accompanied by a significant reduction in the CCI value (0.28 ± 0.10 vs 0.18 ± 0.05; $P < .001$) (Table 2). Post hoc analyses revealed a significant decrease in the average hamstring normalized EMG (32.7 ± 12.2 vs 25.7 ± 9.5 [% of maximum]; $P = .003$) and average gastrocnemius normalized EMG (27.5 ± 7.0 vs 21.2 ± 6.7 [% of maximum]; $P < .001$) after training (Table 3). No significant differences were found for the other variables of interest (Table 3).

DISCUSSION

The purpose of this study was to examine whether tibiofemoral compressive forces in women who have undergone ACLR decrease when utilizing a landing strategy that encourages greater hip and knee flexion. Consistent with our hypothesis, participants demonstrated a decrease in the peak tibiofemoral compressive force after training. The decreased tibiofemoral compressive force was accompanied by an increase in the hip and knee flexion as well as decreased muscle co-contraction at the knee.

Elevated loading at the knee has been hypothesized as a potential mechanism underlying the increased risk for knee osteoarthritis in those who have undergone ACLR.[2,26] In animal models, exposure to excessive compressive forces has been shown to result in damage and degenerative changes in the articular cartilage.[3,4,15] On average, participants in the current study demonstrated a decrease in the peak tibiofemoral compressive force of 6.0 $N \cdot kg^{-1}$ (6.2%; effect size = 0.6) after training. Given the average body mass of 60.3 kg of the participants in the current study, this represents on average a reduction of 362 N (61% body weight). Furthermore, the peak compressive forces of our ACLR participants after training were comparable with the values previously reported in healthy women without ACLR during the same drop-land task (91.3 ± 10.2 vs 88.8 ± 9.8 $N \cdot kg^{-1}$).[26] Although the loading threshold to damage human tibiofemoral cartilage is not known, we believe that the observed decrease in the peak tibiofemoral compression represents a meaningful change in joint loading and may serve to protect the knee joint in the long term.

Post hoc analysis revealed that the reduction in the peak tibiofemoral compressive force after training was primarily the result of a decrease in the compressive component of the quadriceps muscle force followed by a decrease in the compressive component of the GRF, although the differences were not statistically significant (Table 3). This finding is consistent with previous studies that have shown that landing with greater hip and knee flexion reduces impact forces.[11,16,18] As such, using greater hip and knee flexion during landing would not only attenuate the impact force acting on the joint but also would reduce the external knee flexion torque. In turn, less quadriceps muscle force would be necessary to control knee flexion during landing. Given that muscle forces are a primary determinant of joint loading, any reduction in muscle forces would be expected to translate into decreased joint compression.

The decrease in the tibiofemoral compressive force after training was accompanied by a 36% decrease in the level of muscle co-contraction at the knee. After training, the CCI values of the participants after ACLR were similar to those previously reported in healthy women during the same single-legged drop-land task.[26] Post hoc analyses revealed that the decrease in co-contraction after training was a result of decreased activation of the hamstring and gastrocnemius muscles, as no significant difference was found in the average normalized EMG of the quadriceps muscles (Table 3). Besides being a knee flexor, the hamstring and gastrocnemius muscles also serve as hip extensors and ankle plantar flexors, respectively. The decreased hamstring and gastrocnemius activation after training suggests that participants may have relied more on the single joint muscles (ie, gluteus maximus and soleus) to control hip and ankle plantar flexion during landing. However, further investigation would be required to test this hypothesis.

The observed increase in hip and knee flexion and corresponding decrease in muscle co-contraction after training suggest that participants with ACLR were capable of modifying their muscle recruitment strategies in response to the single training session. The immediate changes observed after the short training session imply that muscle strength may not be a critical determinant of the neuromuscular strategies employed by individuals following ACLR when performing a drop-land task. This premise is consistent with the findings of Mizner et al,[17] who reported that improvements in landing mechanics after technique instruction were independent of muscle strength. While muscle strengthening typically is emphasized in the rehabilitation of athletes after ACLR, the findings of the current study demonstrate the benefit of incorporating movement re-education during rehabilitation given its potential to minimize loading at the tibiofemoral joint.

While decreased tibiofemoral compressive forces were observed immediately after training, it is unclear whether the observed changes would persist beyond the time frame evaluated in the current study. In addition, it is not clear whether the observed changes during landing can be generalized to other functional activities. For those who have undergone ACLR to benefit from reduced tibiofemoral compressive forces, long-term changes across all functional activities would be necessary. Future studies with a longer training period and follow-up testing are warranted to develop interventions that can result in permanent changes in movement patterns to reduce tibiofemoral loading for those who have undergone ACLR.

Several limitations of the current study need to be acknowledged. Although subject-specific muscle volumes and tendon orientation as quantified using MRI were incorporated to improve the accuracy of the model,[5,25] the use of generic values for muscle contractile or activation parameters (eg, tendon slack length) may have limited the accuracy of our muscle force estimations. While incorporating subject-specific MRI-measured muscle parameters significantly improves the accuracy of an EMG-driven model,[25] this approach does not completely account for rotational equilibrium, as errors still existed in knee joint moment predictions. Given the within-subject design of the current study, however, any errors resulting from the use of these generic parameters would be consistent for both the pretraining and posttraining assessments.

Furthermore, the sample size of the current study was relatively small, and information of a concomitant meniscal injury was not obtained from the ACLR participants. As such, the generalization of our findings to the entire ACLR population may be limited. In addition, the current study was conducted using a simple repeated-measures design without a control group and blinding process to eliminate potential tester bias. Moreover, the ACLR group consisted of individuals with mixed graft types (7 allografts and 3 BPTB autografts). While increased muscle co-contraction has been observed in persons who have undergone ACLR using various graft types,[8,19,28] the type of surgical method may influence the response to training. Future studies that utilize a larger sample size and compare different types of ACLR grafts are needed to better understand how individuals who have undergone ACLR modify their neuromuscular strategies and joint-loading profiles in response to training.

CONCLUSION

The results of the current study suggest that tibiofemoral compressive forces and muscle co-contraction observed in women who have undergone ACLR can be reduced by instructing athletes to land with greater hip and knee flexion. These findings provide useful information for establishing rehabilitation and/or intervention programs aimed to decrease knee joint loading to prevent or delay the development of knee osteoarthritis in those who have undergone ACLR.

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