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Effect of knee flexion angle on ground reaction forces, knee moments and muscle co-contraction during an impact-like deceleration landing: Implications for the non-contact mechanism of ACL injury

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ABSTRACT

Investigating landing kinetics and neuromuscular control strategies during rapid deceleration movements is a prerequisite to understanding the non-contact mechanism of ACL injury. The purpose of this study was to quantify the effect of knee flexion angle on ground reaction forces, net knee joint moments, muscle co-contraction and lower extremity muscles during an impact-like, deceleration task. Ground reaction forces and knee joint moments were determined from video and force plate records of 10 healthy male subjects performing rapid deceleration single leg landings from a 10.5 cm height with different degrees of knee flexion at landing. Muscle co-contraction was based on muscle moments calculated from an EMG-to-moment processing model. Ground reaction forces and co-contraction indices decreased while knee extensor moments increased significantly with increased degrees of knee flexion at landing (all $p<0.005$). Higher ground reaction forces when landing in an extended knee position suggests they are a contributing factor in non-contact ACL injuries. Increased knee extensor moments and less co-contraction with flexed knee landings suggest that quadriceps overload may not be the primary cause of non-contact ACL injuries. The results bring into question the counterbalancing role of the hamstrings during dynamic movements. The soleus may be a valuable synergist stabilizing the tibia against anterior translation at landing. Movement strategies that lessen the propagation of reaction forces up the kinetic chain may help prevent non-contact ACL injuries. The relative interaction of all involved thigh and lower leg muscles, not just the quadriceps and hamstrings should be considered when interpreting non-contact ACL injury mechanisms.

1. Introduction

One of the most often injured structures in the knee is the anterior cruciate ligament (ACL) with over 200,000 ACL injuries occurring each year in the United States [1,2]. Non-contact, sudden deceleration, with the knee positioned near full extension, is considered the primary cause of the majority of ACL injuries [3–5]. Sagittal and non-sagittal plane biomechanical factors have been associated with ACL injury [6–9]. Valgus loading is considered by many to be an important mechanism of non-contact ACL injury [6–8] while other research suggests that non-sagittal plane moments alone are incapable of causing isolated ACL injury [9–15]. The results of *in vivo* and *in vitro* research have identified the quadriceps as the principal muscle group that generates anterior tibial translation capable of injuring the ACL within this non-contact mechanism of injury [2,3,5,14,16–19]. Small knee flexion angles [3,9,20,21] and large knee extensor moments are important mechanisms coupled to non-contact ACL strain but little is known about how the extensor moments change with the knee flexion angle during sudden decelerations.

A number of studies using various methodologies have suggested that the hamstrings can provide counterbalancing force to protect against quadriceps induced anterior tibial translation, thereby reducing the potential for ACL injury [16,22–29]. Coordinated recruitment of the quadriceps and hamstrings is considered essential to establishing dynamic stabilization of the knee, thereby reducing reliance on static restraints such as the ACL [30,31]. Electromyography (EMG) is a measure often used to investigate the quadriceps–hamstrings balance of patients with ACL injury and healthy subjects while performing a variety of movement tasks [30–35]. Although some results based on EMG bring into question the potential role of the hamstrings to counterbalance anterior tibial translation induced by the quadriceps [34,35] others have suggested that knee extensor and flexor co-activation is an important strategy for stabilizing the knees of individuals with ACL injury [30–32]. One methodological explanation for the conflicting results is the assumption of a linear relationship between EMG and muscle force. The association of force with EMG measures is complex and dependent on muscle kinematics [36,37]. However, simple EMG-to-force models based on joint angle

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calibrations that incorporate muscle moment arm and length changes have been shown to be useful when quantifying co-contraction for muscles crossing the knee [32]. The role of the hamstrings in counteracting quadriceps force, based on EMG measures, during dynamic movements should consider the effects of muscle kinematics.

Research related to knee stabilization and co-activation generally considers only the thigh musculature and ignores the gastroc–soleus complex [16,17,22,24–27,29,32,33,38,39]. It has been reported that athletes injured during deceleration activities tend to land with a dorsiflexed or flat-foot position leading to abnormal absorption of ground reaction forces [40]. Based on this analysis, Boden et al. speculated that the gastroc–soleus complex may play a protective role in preventing ACL injuries [40]. *In vitro* works suggest that the gastrocnemius force can strain the ACL by eliciting posterior femoral translation relative to the tibia, whereas the soleus can limit anterior translation of the tibia produced by the quadriceps [18,19]. Assessment of the calf musculature is therefore warranted when considering knee stabilization during rapid deceleration movements.

A sudden deceleration landing with a flexed knee should generate a reduced ground reaction force due to antigravity muscles being in a more advantageous position to absorb kinetic energy at impact, thereby reducing the reaction forces that propagate up the lower limb kinetic chain [41–43]. A flexed knee at landing, however, should be accompanied by higher knee extensor moments due to the distance of the resultant ground reaction force vector from the knee joint axis [44,45]. Presumably, when the knee is flexed at landing, extensor moments increase and co-contraction also increases to limit extensor moment induced anterior tibial translation. Given that most non-contact ACL injuries occur when the knee is near full extension, knee extensors likely dominate and antagonists such as the hamstrings provide insufficient counterbalancing force to stabilize the joint against anterior tibial translation. To our knowledge, a controlled study investigating the effects of knee flexion angle for a rapid deceleration impact-like landing on the knee joint moments, ground reaction forces, co-contraction indices, and independent moments generated by quadriceps, hamstrings and the gastroc–soleus complex has not been reported. The purpose of this study was to investigate the effect of the knee flexion angle at landing on these measures. It was hypothesized that with increased knee flexion at landing the ground reaction forces would decrease, the net knee extensor moment would increase and the co-contraction of muscle crossing the knee would increase. Considering the role of the lower leg muscles and changing knee joint moments relative to knee flexion angle at landing may help clarify why most non-contact ACL injuries occur near full knee extension.

2. Methods

Ten males age (26.4 ± 2.9 years and mass of 76.03 ± 8.12 kg) volunteered to participate in the study. Subjects were recruited from the campus of the University at Buffalo. All volunteers were physically active and or involved in sport participation at a recreational level. Potential subjects were excluded if they reported a history of significant knee injury that required surgery or if they had been diagnosed as having a structurally unstable knee. The study protocol was approved by a Human Subjects Institutional Review Board. Written informed consent was obtained from each subject prior to data collection. All data collections occurred within the University at Buffalo Biomechanics Laboratory and were conducted by one assessor. The study protocol involved subjects undergoing a series of ankle and knee joint moment calibration trials followed by the performance of a series of rapid deceleration impact-like maneuvers with different angles of knee flexion at landing.

The ankle and knee joint moment calibration trials consisted of a series of slow velocity (15°/s) concentric, open chain contractions performed on a Cybex II isokinetic dynamometer. The trials were repeated several times at different levels of effort. The hip angle for calibration trials was always neutral (~15° flexion). Knee flexion and extension trials were performed from full extension through approximately 90°. Ankle plantar flexion trials were performed from (~15°) of dorsiflexion through full plantarflexion. The moment values for all flexion and extension trials were corrected for gravity effects.

EMG data was collected using surface EMG electrodes (Ag/AgCl) placed over the center of the muscle belly of the vastus lateralis, vastus medialis oblique, semitendinosus, biceps femoris, medial gastrocnemius and soleus with the reference electrode placed over the bony portion of the proximal shaft of the tibia [46,47]. Standard skin preparation was used to reduce electrical impedance at the skin–electrode interface [48]. The EMG data was pre-amplified (1000×), high pass filtered at 10 Hz, amplified (variable 10 to 20×) and low pass filtered at 500 Hz prior to analog-to-digital conversion (16 bit) at 2400 Hz.

Moment of force, joint angle and EMG data from the calibration trials were used to create EMG-to-moment regression equations. A 0.5 s range was used to window the moment and EMG data for the variable effort calibration trials. This calibration data was sampled starting 0.25 s prior to knee flexion angles of 15°, 35° and 65°, and 0.25 s prior to an ankle plantarflexion angle of 3°. The knee joint angles were selected to fall within the 0–25°, 25–50°, and 50–75° target angle ranges for the knee flexion landing conditions, while the ankle position was selected to fall within the dorsiflexion to full plantarflexion range. Moment values were averaged over the 0.5 s sample. A root-mean-square (RMS) value was calculated for the EMG data over the 0.5 s range. Linear, best fit regression equations for EMG-to-moment conversions were calculated based on at least 6 variable effort data points for the different calibration angles (Fig. 1). The calibration regression equations were used to convert the muscle EMG recorded for each of the landing trials to muscle moment values. The EMG instrumentation settings, data sampling, processing and analysis were kept identical for the calibration and the different impact-like deceleration landing trials.

Subjects were required to step off a 10.5 cm high box, landing with the right foot in unilateral stance onto a piezoelectric force plate (Kistler Instrument Inc). Each subject performed a total of nine randomized trials while landing with a knee flexion angle falling within the target ranges. Three trials were completed for each target range, 0–25°, 25–50° and 50–75°. Consistency across trials and conditions was controlled using an audible metronome set at a

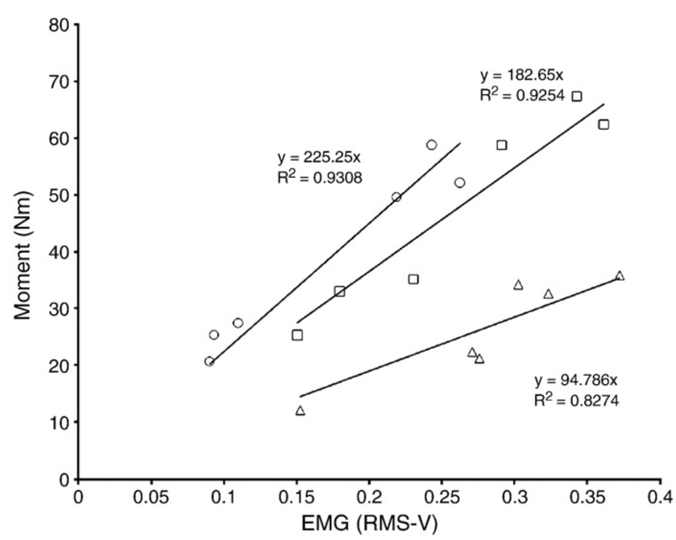

Fig. 1. Medial hamstring moment regression equations for a single subject at 15° (open circles), 35° (open squares) and 65° (open triangles).

constant rhythm. The foot–ground reaction forces at landing were analog-to-digitally converted (16 bit) at 2400 Hz. The force signals were subsequently re-sampled to create a 60 Hz signal by calculating the median force over 40 sample epochs. This re-sampling matched the force plate signal to the spatial data sampling rate collected by video.

The landing trials were captured on S-VHS video tape with the use of a Panasonic 5000 digital video recorder. Segment orientations were determined from five reflective markers attached with adhesive tape to the iliac crest, greater trochanter, tibiofemoral joint line, lateral malleolus and the 5th metatarsal. These markers were required to determine segment anthropometrics and segment endpoints [49]. The spatial coordinates of each marker were determined through conversion from the videotape to digital format at 60 Hz (Peak Motus). EMG, force plate and video data were synchronized. The lower limb of each subject was represented as a 3-segment (thigh, leg, and foot), rigid body model for planar calculation of segment kinematics, and ankle, knee and hip joint moments using inverse dynamics [49]. Segment lengths, knee flexion angles, moments of inertia, segment centers of mass, segment angles, segment angular acceleration, and segmental planar accelerations were determined based on anthropometric tables and methods presented in Winter [49]. Joint moments and reaction forces calculated via inverse dynamics were based on unfiltered data. This approach was taken based on evidence that false moments are created when spatial coordinates and ground reaction force data are filtered at different cut-offs for impact-like landings [50]. The joint moments and forces were filtered after they were calculated using a zero lag, dual pass digital Butterworth filter with the cutoff frequencies determined individually for each subject. The filter cutoff frequencies were calculated by averaging the 98th percentile frequency for the vertical and anterior–posterior ground reaction forces for each individual trial. In order to obtain the 98th percentile frequency, the time domain ground reaction force signals were windowed with an extended cosine bell function and converted to the frequency domain via a Fast Fourier Transform. A power spectral density relationship was calculated and the 98th percentile frequency represented the frequency where the cumulative area of the spectrum was at 98%.

Muscle moments at landing were calculated from the EMG-to-moment regression equations based on a 0.5 s EMG RMS window starting 0.25 s prior to the peak vertical ground reaction force at landing. Co-contraction indices were then calculated from the muscle moments using the Falconer and Winter [49] method:

$$\text{Co-contraction index } (CCI) = 2 \times (M_{\text{antag}} / (M_{\text{agon}} + M_{\text{antag}})) \times 100\%$$

where M_{antag} are the moments of force of the antagonists (hamstrings) and M_{agon} represents agonist (quadriceps and gastrocnemius) muscle groups. Perfect co-contraction would be 100%.

Three repetitions of each condition were collected for each subject. Mean values for ground reaction forces and knee extensor moments were normalized to subject body mass for each condition. Within subjects effects were assessed using repeated measures ANOVA (SPSS). Simple contrast was utilized to determine significance ($p<0.05$) between conditions for each of the dependent variables. Changes in the mean value of muscle moments as a function of landing angle were calculated for each of the muscles monitored and presented graphically (Fig. 2).

3. Results

The vertical (Fy) and posterior (Fx) shear force decreased significantly with an increase in knee flexion at landing, (Table 1). Relative to the 0–25° knee flexion target angle at landing, there was an 11.0% decrease in Fy when landing at 25–50° ($p=0.016$) and a 15.0% decrease when landing at 50–75° knee flexion ($p=0.002$). Fx decreased 20.0% when landing at 25–50° ($p=0.011$) and 20.5% when landing at 50–75° ($p=0.022$). There were no significant differences in Fy or Fx between the 25–50° landing and the 50–75° landing conditions.

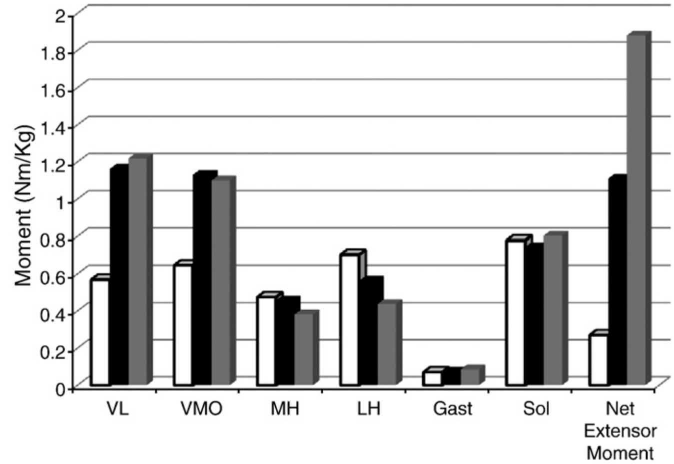

Fig. 2. Average muscular moment output normalized to body weight for the vastus lateralis (VL), vastus medialis oblique (VMO), medial hamstrings (MH), lateral hamstrings (LH), gastrocnemius (Gast), soleus (Sol) and net knee extensor moment for each of the 3 landing conditions; 0–25° (white), 25–50° (solid black) and 50–75° (solid grey).

The net knee extensor moment increased significantly with increasing degrees of knee flexion at landing, (Table 1). Relative to the 0–25° knee flexion target angle, there was a 75.0% increase in the net knee extensor moment when landing at 25–50° ($p=0.000$) and an 85.0% increase at 50–75° knee flexion ($p=0.000$). There was also a significant increase, 41.1%, in the extensor moment between the 25–50 and 50–75° conditions ($p=0.001$).

The co-contraction index decreased significantly with increasing degrees of knee flexion at landing, (Table 1). Relative to the 0–25° knee flexion target angle at landing, there was a 35.0% decrease in the CCI when landing at 25–50° ($p=0.000$) and a 43.0% decrease at 50–75° knee flexion at landing, ($p=0.000$). There was no difference between the 25–50 and 50–75° conditions.

The hamstring moments (average of medial and lateral groups) decreased with increasing degrees of knee flexion at landing, (Fig. 2). Relative to the 0–25° knee flexion target angle there was a 13.4% decrease when landing at 25–50° and a 30.3% decrease at 50–75° knee flexion at landing. A 19.6% decrease was noted in going from the 25–50° to the 50–75° condition. The quadriceps moments (average of vastus medialis and lateralis) increased with increasing degrees of knee flexion at landing, (Fig. 2). Relative to the 0–25° knee flexion target angle there was a 47.0% increase when landing at 25–50° and a 47.6% increase at 50–75° knee flexion at landing. A 1.1% increase was noted in going from the 25–50° landing to the 50–75° landing. The gastrocnemius moment was low relative to the other muscles and increased with increasing degrees of knee flexion at landing, (Fig. 2). Relative to the 0–25° knee flexion target angle there was a 3.2%

Table 1
Mean landing angle and dependent measures for the three target landing conditions.

Target knee flexion landing condition	0–25 (°)	25–50 (°)	50–75 (°)
Knee flexion landing angle (°)	17.1 ± 4.1	35.4 ± 2.6	64.9 ± 2.1
Peak vertical (Fy) GRF (N/kg)	19.3 ± 5.0	17.1 ± 4.5[a]	16.4 ± 3.6[a]
Peak posterior shear (Fx) GRF (N/kg)	4.30 ± 1.9	3.4 ± 1.1[a]	3.4 ± 1.1[a]
Peak knee extensor moment (Nm/kg)	0.27 ± 0.25	1.1 ± 0.34[b]	1.9 ± 0.6[b,c]
Knee co-contraction index (%)	84.0 ± 35.8	54.2 ± 33.6[b]	47.7 ± 24.6[b]

Notes: Mean ($N=10$) ± standard deviation. Superscripts [a] indicates significantly ($p<0.05$) less than 0–25° value; [b] indicates significantly greater; [c] indicates significantly greater than 25–50° value.

increase when landing at 25–50° and a 17.7% increase at 50–75° knee flexion at landing. A 14.9% increase was noted in going from the 25–50° position to 50–75° knee flexion at landing. The soleus moment indicates that this muscle was consistently active with little variability across the different knee flexion angles at landing, (Fig. 2).

4. Discussion

The present study quantified the effect of different degrees of knee flexion at landing during an impact-like lunge deceleration on: knee joint moments, ground reaction forces, muscle co-contraction indices and the role of lower extremity muscles. Consistent with our hypothesis, ground reaction forces decreased and the net knee joint extensor moment increased with increasing angles of knee flexion at landing. The hamstrings moments did not increase to generate a counterbalancing moment to stabilize the knee when the knee extensor moment increased; hence, there was less co-contraction of the knee flexors and extensors with increasing knee flexion at landing. Results from the present study suggest that a flexed knee landing is more likely to result in quadriceps overload induced strain on the ACL. Yet, most non-contact ACL injuries occur near full knee extension [3–5]. Theories attributing ACL strain to quadriceps overload [2,5] and insufficient hamstrings stabilization [16,17,19,22,24–27,29,38,39] need further examination.

The magnitude of the ground reaction forces was greatest when landing within the 0–25° knee flexion range during the rapid deceleration task. An extended knee is more likely to result in a rapid change in kinetic energy at landing yielding greater foot–ground impact forces. This was the result found in the present study and it implies a greater risk of injury in an extended knee landing position, which is consistent with the non-contact mechanism of ACL injury [3–5]. With increased knee flexion at landing, antigravity muscles are in a more advantageous position to absorb kinetic energy at impact, thereby reducing reaction forces that propagate up the lower limb kinetic chain [41–43]; however, the consequence is an increased net knee extensor joint moment [44,45]. The increase noted in the present study was due to the orientation of the resultant reaction force vector relative to the knee joint center. With an extended knee at landing, this force vector was closer to the knee joint axis resulting in a smaller moment arm. Hence, the internal, net knee extensor moment required to counterbalance the net external knee flexor moment was less when landing closer to an extended knee position. Given that ground reaction forces are more likely to be greatest and knee extensor moments smallest when landing in an extended knee position; it is possible that ACL strain from non-contact deceleration may be related to rapid translational joint forces that propagate up the kinetic chain rather than resulting from quadriceps overload induced anterior tibial translation. Boden et al. [40] also proposed that a lack of absorption of ground reaction forces at landing may be a factor in ACL injury. They hypothesize that ACL injuries were due to a lack of energy dissipation by the ankle plantar flexors when impacting the ground heel first or with a flat-foot position. In their video analysis of non-contact ACL injuries, athletes who were injured landed with a mean knee flexion angle of 17.6° compared to uninjured controls that landed with a more plantar flexed ankle and had a knee flexion position of 39.3°. The position of the injured athlete's knee is analogous to the 0–25° condition of the present study. Within this 0–25° landing position, we noted the smallest knee extensor moments with the greatest amount of co-contraction at the knee and the largest ground reaction forces. The results from the present study and the work of Boden et al. [40] suggest that the propagation of reaction forces when landing with the knee near full extension could be an important component of non-contact ACL injuries.

Co-contraction index (CCI) values decreased with increased knee flexion angles at landing (Table 1). It was expected that CCI values would increase due to the need for hamstring moments to counterbalance the quadriceps and gastrocnemius activity and subsequent anterior translation of the tibia. Based on the muscle moments calculated from the EMG, the quadriceps moment, (vastus lateralis and vastus medialis oblique) increased while moment values for the hamstrings (medial and lateral) decreased with increasing knee flexion at landing. Various methods for determining co-contraction based on EMG have been reported in the literature [31–33,36,49]. Methods that make use of isometric calibrations at a single joint angle and assume a linear relationship between EMG and muscle force do not account for the effect of changing muscle kinematics under dynamic conditions [30,33]. Within the present study we utilized a simple calibration method which considered the relationship between EMG and muscle force to be a function of joint angle (Fig. 1). Utilization of this approach allowed for co-contraction to be based on the formula proposed by Falconer and Winter [49]. The 0–25° condition CCI value is the closest to 100% muscular balance, indicating that hamstrings were providing significant counteraction to the quadriceps and gastrocnemius moments. Hamstring moments did not increase in proportion to quadriceps moments at deeper knee flexion landing angles. These results are consistent with others who have reported similar findings during incremental, isometric knee extensor moment tasks [34,35]. It has been proposed that ACL injury is due in part to a lack of counter-balancing stabilization by the hamstrings [2,3,5,16,17,19,38,51]. Our co-contraction results and those of others suggest that the potential role of the hamstrings in the protection of ACL during dynamic activities requires further investigation.

The potential stabilizing role of the soleus muscle was also assessed in the present study. The soleus is capable of producing posterior tibial translation in closed chain movement based on its origin and insertion, whereas the gastrocnemius' posterior pull on the femur would strain the ACL [19,51]. Muscle moments calculated for the soleus were considerable during the deceleration tasks. The 0–25° landing produced a value of 0.78 Nm/kg with values of 0.74 and 0.80 Nm/kg noted in the 25–50 and 50–75° landings respectively. These soleus moments are generated at the ankle but they will also elicit a posterior directed force on the tibia when the foot is fixed [19]. It may well be that the soleus and other lower leg muscles crossing the ankle provide a supportive role in protecting the ACL by stabilizing the tibia at landing.

There are limitations to the present work. The reaction force and moment analysis was restricted to the sagittal plane. Frontal plane, valgus loading is considered by many to be a critical factor in non-contact ACL injuries [6–8]. The present study was limited to an anterior–posterior loading task with targeted knee flexion landing angles. Using knee flexion as a controlled dependent measure is unique in the investigation of ACL injury mechanisms but our protocol may not be realistic relative to the loading and landing conditions of rapid deceleration, non-contact ACL injury circumstances. Notwithstanding these limitations, the present study accounted for the effect of joint angle changes in the EMG calibrations, and our inverse dynamic calculation of knee moments avoided false moments created when marker coordinates and ground reaction forces are filtered at different cut-offs [50]. Further investigation of this nature, incorporating gender and age comparison as well as subjects who are ACL deficient, ACL reconstructed and at risk of ACL injury based on sport activity may now be warranted. Furthermore, comparison of soleus activity across these populations may be interesting with respect to clinical implications.

In summary, quadriceps overload may not be the primary cause of non-contact ACL injuries when landing with an extended knee. Knee extensor joint moments, traditionally thought to be a main cause of ACL injury were actually smallest within the 0–25° knee flexion condition. In the present study, ground reaction forces were greatest in the extended knee landing condition and could be an important component of non-contact ACL injuries that needs further investigation. Co-contraction indices imply that the hamstrings provide a substantial counter-moment to the quadriceps moment with the knee near full extension but their stabilizing role at deeper positions of knee flexion is questionable. Soleus may play an important stabilizing role at landing, which should also be explored further. The relative interaction of all involved thigh and lower leg muscles, not just the quadriceps and hamstrings should be considered in the interpretation of non-contact ACL injury mechanisms and when devising prevention strategies.

5. Conflict of interest

Neither of the authors has any financial or personal relationships with organizations or individuals that could inappropriately influence this work.

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