Influence of Relative Hip and Knee Extensor Muscle Strength on Landing Biomechanics

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ABSTRACT

STEARNS, K. M., R. G. KEIM, and C. M. POWERS. Influence of Relative Hip and Knee Extensor Muscle Strength on Landing Biomechanics. *Med. Sci. Sports Exerc.*, Vol. 45, No. 5, pp. 935–941, 2013. **Purpose**: This study aimed to determine whether the tendency of women to exhibit higher knee extensor moments relative to hip extensor moments during the deceleration phase of landing can be explained by the relative strength of the knee and hip extensors. **Methods**: Forty recreationally active individuals participated in this study (20 men and 20 women). The maximum isometric strength of the knee extensors and hip extensors was assessed using a load cell and custom testing setup. Lower extremity kinematics and kinetics were collected during a double-leg drop-jump task. **Results**: When compared with men, women demonstrated a significantly higher knee–hip extensor moment ratio during the deceleration phase of landing (1.42 ± 0.6 vs 1.12 ± 0.3, $P < 0.001$). In addition, the knee–hip extensor isometric strength ratio was significantly higher in women compared with men (1.01 ± 0.2 vs 0.89 ± 0.2, $P < 0.001$). The Pearson partial correlation (controlling for sex) revealed a significant positive association between the knee–hip extensor isometric strength ratio and the knee–hip extensor moment ratio ($r = 0.41$, $P = 0.005$). **Conclusion**: The tendency of women to exhibit higher knee extensor moments relative to hip extensor moments may be explained, in part, by the relative strength of the hip and knee extensors. **Key Words:** ACL INJURY, MOMENT RATIO, SEX DIFFERENCES, LOWER EXTREMITY

Tears of the anterior cruciate ligament (ACL) are the most common ligamentous injury of the knee joint. Although ACL tears are observed in both men and women, a disproportionate number of injuries occur in women (1). Biomechanical studies analyzing sex differences during sports-specific maneuvers have demonstrated that women exhibit a lower extremity biomechanical profile that is thought to place them at increased risk for ACL injury. Specifically, women tend to land with less hip and knee flexion and exhibit greater knee extensor moments and lower hip extensor moments when compared with men (4,6,8,22,23). In addition, women exhibit higher knee abduction (valgus) angles and internal knee adductor (valgus) moments during landing and side-step cutting when compared with men (7,10,13,18).

Although women consistently have been shown to exhibit increased knee joint loading in the sagittal and frontal planes, the underlying reason for this biomechanical tendency is unclear. Several authors have suggested that deficits in hip abductor strength may contribute to the greater knee abduction angles and knee adductor moments observed in women (5,10,12,14,25). Pollard et al. (20) hypothesized that the tendency of women to exhibit higher knee extensor moments relative to hip extensor moments (increased knee–hip extensor moment ratio) may result from strength deficits of the hip extensors. The weakness of the hip extensors is thought to result in a compensatory strategy whereby women adopt an overreliance on their quadriceps and passive restraints in the frontal plane (ligaments) to absorb impact forces.

Although studies have demonstrated that women exhibit deficits in hip extensor muscle performance compared with men, no study has examined the relationship between hip extensor muscle performance and sagittal plane kinematics and kinetics during the deceleration phase of landing (2,3,5). In addition, no study has examined the interrelationship of knee and hip extensor strength to sagittal plane biomechanics. Given the importance of the hip extensors to decelerate the center of mass during landing, it is plausible that diminished hip extensor strength relative to knee extensor strength may underlie the tendency for women to exhibit higher knee moments relative to hip moments during landing.

The purpose of the current study was threefold. First, we compared lower extremity biomechanics between men and women during the deceleration phase of a drop-jump task. Second, we quantified sex differences in the maximum isometric strength of the knee and hip extensors. Finally, we evaluated the relationship between the knee and hip extensor

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strength and the knee and hip extensor moments during a drop-jump task. On the basis of previous literature, we hypothesized that women, when compared with men, would exhibit a biomechanical profile that is suggestive of increased risk for ACL injury including 1) decreased hip and knee flexion angles, 2) increased knee abduction angles and knee adductor moments, and 3) an increased knee–hip extensor moment ratio. We also hypothesized that compared with men, women would demonstrate a higher knee–hip extensor strength ratio. Finally, we hypothesized that the knee–hip extensor strength ratio would be positively correlated with knee–hip extensor moment ratio during the deceleration phase of the drop-jump task.

METHODS

Forty subjects between the ages of 18 and 25 yr participated in this study (20 men and 20 women; Table 1). All subjects were healthy and met the following inclusion criteria: 1) recreationally active (i.e., participation in some form of physical activity for at least 30 min two times a week) and 2) a history of at least 2 yr of recreational and/or competitive sports participation at some point in their lifetime. All subjects were matched for their activity level based on the Global Physical Activity Questionnaire and minutes of activity per week. Subjects were excluded from participation in the study if they reported any of the following: 1) a history or diagnosis of hip or knee pathology or trauma, 2) current hip or knee pain during any sports activities or activities of daily living, or 3) inability to perform the landing task due to any neurologic or medical condition.

Kinematic data were collected at 250 Hz using a ten-camera motion analysis system (Qualisys Motion Capture Systems, Gothenburg, Sweden). Reflective markers (14-mm spheres) were placed at specific anatomical landmarks, which were used to determine the three-dimensional motion of the pelvis and dominant lower extremity. Ground reaction forces were collected at a rate of 1500 Hz using two floor-embedded AMTI force plates (AMTI, Newton, MA).

Muscle performance testing was conducted using a uniaxial force transducer (Model LCCA-1K; Omega Engineering, Inc., Stamford, CT) connected to a nonstretchable fabric strap. The subject's skin was protected by a fabric pad lined with a dense foam pad fixed to the distal end of the strap. The strap and the connectors had a maximum capacity of 1779 N, and the tensile capacity of the force transducer was 4454 N. The signal from the force transducer was sampled digitally at 1500 Hz. Real-time feedback of the force generation was displayed on a computer monitor using a custom computer program in Labview (Labview® version 8.0.1; National Instruments Corp., Austin, TX).

All data were collected at the Jacquelin Perry Musculoskeletal Biomechanics Research Laboratory at the University of Southern California. Subjects first underwent muscle performance testing followed by biomechanical analysis of a drop-jump task. Before participation, the purposes of the study, procedures, and risks were explained to each subject. Written informed consent was obtained from each participant as approved by the institutional review board for the University of Southern California Health Science Campus. The height, the weight, and the dominant limb of each subject were recorded. For all data collection procedures outlined in the following paragraphs, data were only obtained from the dominant limb. Leg dominance was determined as the leg with which subjects preferred to perform a maximal single leg forward jump.

For muscle performance testing of the knee extensors, subjects were positioned in a seated position with their hip and knee joints in 90° and 60° flexion, respectively (Fig. 1A). A testing strap was placed around the distal tibia just superior to the lateral malleoli. The strap was then connected to the force transducer, which was connected in series with a second strap affixed to a crossbar under the testing platform. A sliding plate attached to the crossbar allowed for positioning of the force transducer and fixation of the strap so that the line of pull of the transducer was perpendicular to the tibia. The pelvis was secured to the testing chair with a second strap to ensure that the pelvis did not move during testing. The distance from the lateral knee joint line (representative of the knee joint axis of rotation) to the lateral midpoint of the testing strap was measured for the calculation of knee extensor torque.

For muscle performance testing of the hip extensors, subjects were positioned prone on the testing table with their nondominant leg on the ground (Fig. 1B) (26). The dominant limb was positioned with the hip and knee joints in 60° and 90° flexion, respectively. The testing strap was placed around the thigh of the dominant leg 10 cm superior to the lateral knee joint line. The distance from the greater trochanter (representative of the hip joint axis of rotation) to the lateral midpoint of the testing strap was measured for the calculation

TABLE 1. Subject demographics.

	Male ($n = 20$)	Female ($n = 20$)	P
Age (yr)	23.2 ± 1.3	23.7 ± 1.2	0.30
Height (m)	1.8 ± 0.1	1.7 ± 0.1	<0.001
Weight (kg)	81.5 ± 16.7	63.3 ± 11.1	<0.001
Global Physical Activity Questionnaire score	2122.1 ± 917.9	1915.0 ± 520.2	0.19
Minutes of exercise per week	249.5 ± 104.6	223.7 ± 90.6	0.31

Values are presented as mean ± SD.

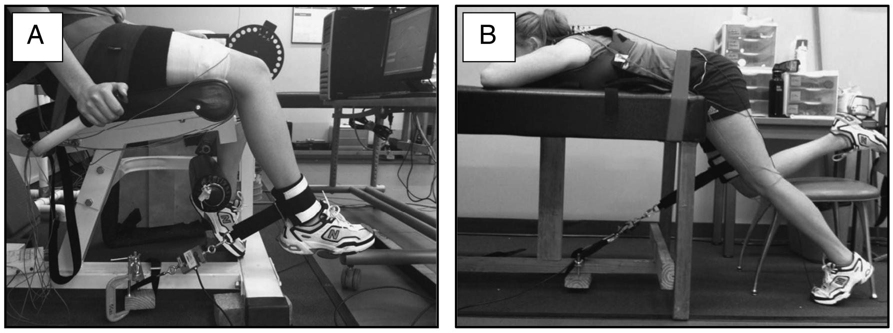

FIGURE 1—Muscle performance testing setup for the knee extensors (A) and the hip extensors (B). Testing of the knee extensors was performed with the knee in 60° flexion and the hip in 90° flexion. Testing of the hip extensors was performed in a prone position with the hip in 60° flexion and the knee in 90° flexion.

of hip extensor torque. Positioning and fixation of the straps and force transducer were identical with that described above for testing of the knee extensors.

For each position described previously, subjects performed a 5-s maximum isometric contraction. Before each trial, subjects were asked to apply a preload of 7 lb (31 N) to take slack out of the strap system. Visual confirmation of the force applied before the contraction was confirmed by the investigator through the visual display. For each trial, subjects were encouraged to push against the fixed strap "as hard as possible." Subjects were verbally encouraged to produce their maximum force throughout the duration of the trial. A total of two isometric trials were collected for each muscle group of interest. A 2-min rest was provided between trials.

To control for the potential influence of varying footwear, each subject was fitted with same style of cross-training shoe (New Balance Inc., Boston, MA). Reflective markers (14-mm spheres) were attached to the following bony landmarks: distal first toe, first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral epicondyles of the femur, greater trochanters, anterior superior iliac spines, iliac crests, and L5–S1 junction. Noncollinear tracking cluster markers were attached to neoprene bands secured around the thigh and shank of the dominant lower extremity. In addition, noncollinear tracking cluster markers were secured to the heel counter of the shoe on the dominant extremity. Once all markers were attached, a standing calibration trial was captured. After the collection of the calibration trial, all markers were removed, except for the tracking clusters and the markers on the iliac crest and L5–S1 junction.

Subjects performed a double-leg drop-jump task from a 36-cm platform. During the drop-jump task, subjects were instructed to grab a ball on the descent phase of the drop-jump and then jump up with the ball overhead (Fig. 2). This task required subjects to focus on the action of grabbing a ball as opposed to the performance of the landing maneuver. The ball was placed on an adjustable stand located just in front of the force plates upon which the subjects were landing. To position the ball at the appropriate height, subjects were asked to stand on the force platform in the approximate location on which they were going to land and hold the ball with their arms outstretched in front of them in 45° shoulder flexion. The stand was then adjusted to the appropriate height to place the ball in this position for the drop-jump task.

Subjects were instructed to hop off the support platform, grab the ball on the descent phase of landing, land with one foot on each of the two force plates, and then jump as high as

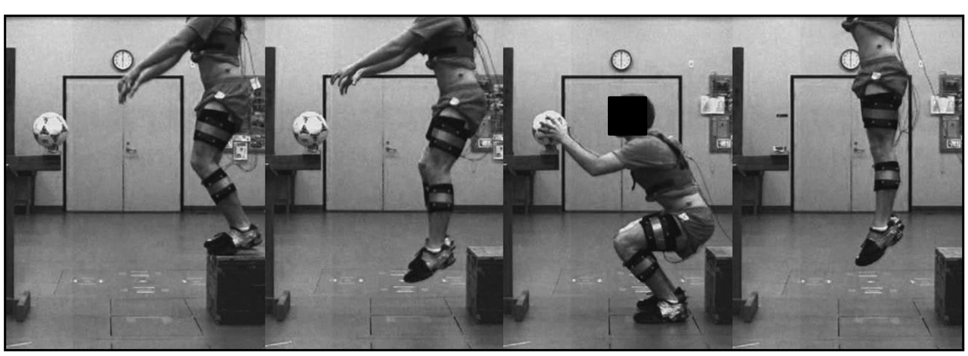

FIGURE 2—The double-leg drop-jump task used in the current study. Subjects were instructed to drop off the platform, land with one foot on each force plate while simultaneously grabbing a ball, then jump up as high as possible with the ball overhead.

possible with the ball overhead. Subjects were not provided with any instruction regarding their jump mechanics and were allowed two to three practice trials to familiarize themselves with the task. After practice, four trials were collected.

Analog data obtained from the muscle performance testing were analyzed using a custom Matlab program (The MathWorks, Inc., Natick, MA). Data from each trial were filtered using a second-order, zero-lag Butterworth filter with a cutoff frequency of 10 Hz. Raw data were converted to newtons and multiplied by the respective moment arm to calculate the torque generated during testing. The peak torque produced during each trial was calculated, and the peak value produced during either trial was used for statistical analysis. Torque data were normalized to body mass.

To examine the interlimb strength ratio between the knee and the hip extensors, the knee–hip extensor strength ratio was calculated for each subject. This was accomplished by dividing the maximum isometric strength of the knee extensors by that of the hip extensors. A muscle performance ratio >1 indicated that the strength of the knee extensors was greater in magnitude than that of the hip extensor, whereas a ratio <1 indicated that the strength of the hip extensors was greater in magnitude than that of the knee extensors.

Three-dimensional marker coordinates recorded during the drop-jump task were obtained using the QTM software (Qualisys Track Manager, version 2.5; Qualisys Motion Capture Systems, Gothenburg, Sweden). Visual3D software (C-motion, Rockville, MD) was used to process the raw coordinate data and compute the segmental kinematics and kinetics. Trajectory and analog data were filtered using a fourth-order zero-lag Butterworth 10-Hz low-pass filter. The pelvis was modeled as a cylinder, and the lower extremity segments were modeled as a frusta of cones. The local coordinate systems of the pelvis, thigh, shank, and foot were derived from the standing calibration trial. Joint kinematics were calculated using Euler angles with the following rotation order: flexion/extension, abduction/ adduction, and internal/ external rotation. The knee joint angle was defined as the orientation of the shank segment with respect to the thigh segment, and the hip joint angle was defined as the orientation of the thigh segment relative to the pelvis. Sagittal and frontal plane net joint moments (internal) were calculated using inverse dynamics equations and normalized to body mass.

For purposes of this study, kinematic and kinetic variables from the subject's dominant limb were analyzed over the deceleration phase of the drop-jump task (i.e., initial contact to the time of peak knee flexion). The biomechanical variables of interest included the peak knee and hip flexion angles, the peak knee abduction angle, the average knee extensor and adductor moment, and the average hip extensor moment. As previously described by Sigward et al. (23), the knee and hip moments were averaged across the deceleration phase to examine the relative contributions of the hip and knee to deceleration of the center of mass during landing. The knee–hip extensor moment ratio was calculated by dividing the average knee extensor moment by the average hip extensor moment. A ratio >1 indicated that the knee extensor moment was greater in magnitude than the hip extensor moment, whereas a ratio <1 indicated that the hip extensor moment was greater in magnitude than the knee extensor moment.

A one-way between-group multivariate analysis of variance was performed to investigate sex differences in the kinematic, kinetic, and strength variables of interest. Ten dependent variables were used: peak knee flexion angle, peak hip flexion angle, peak knee abduction angle, average knee extensor moment, average hip extensor moment, average knee adductor moment, hip extensor maximum strength, knee extensor maximum strength, knee–hip extensor moment ratio, and knee–hip extensor strength ratio. The independent variable was gender.

Pearson correlations were used to assess the relationship between the knee–hip extensor moment ratio and the strength of the knee extensors, hip extensors, and knee–hip extensor strength ratio. If a statistically significant sex difference was found for the muscle performance variables and

TABLE 2. Sex differences in lower extremity kinematics, kinetics, and muscle performance variables.

	Male	Female	*P*
Kinematics			
Peak knee flexion angle (°)	109.6 ± 7.6 (105.7–113.5)	93.0 ± 9.5 (89.2–96.9)	<0.001
Peak hip flexion angle (°)	92.5 ± 5.6 (89.6–95.4)	83.1 ± 7.0 (80.3–86.0)	<0.001
Peak knee abduction angle (°)	2.0 ± 2.7 (0.6–3.5)	6.5 ± 3.5 (5.0–7.9)	<0.001
Kinetics			
Average knee extensor moment ($N{\cdot}m{\cdot}kg^{-1}$)	1.21 ± 0.2 (1.29–1.13)	1.12 ± 0.2 (1.20–1.04)	<0.001
Average hip extensor moment ($N{\cdot}m{\cdot}kg^{-1}$)	1.11 ± 0.2 (1.21–1.00)	0.88 ± 0.3 (0.99–0.78)	<0.001
Knee–hip extensor moment ratio	1.12 ± 0.3 (0.90–1.35)	1.44 ± 0.6 (1.20–1.65)	<0.001
Average knee adductor moment ($N{\cdot}m{\cdot}kg^{-1}$)	−0.11 ± 0.1 (0.16–0.05)	0.07 ± 0.1 (0.01–0.12)	<0.001
Maximum isometric strength			
Knee extensors ($N{\cdot}m{\cdot}kg^{-1}$)	3.66 ± 0.6 (3.44–3.89)	2.82 ± 0.4 (2.60–3.05)	<0.001
Hip Extensors($N{\cdot}m{\cdot}kg^{-1}$)	4.27 ± 1.0 (3.90–4.64))	2.87 ± 0.8 (2.50–3.24)	<0.001
Knee–hip extensor strength ratio	0.89 ± 0.2 (0.81–0.96)	1.01 ± 0.2 (0.94–1.09)	<0.001

Values are presented as mean ± SD (95% confidence interval).

the knee–hip extensor moment ratio, the Pearson partial correlation was calculated controlling for sex. All statistical analyses were performed using PASW statistical software (Chicago, IL) with a significance value of $P \leq 0.05$.

RESULTS

There was a statistically significant difference between men and women on the combined dependent variables ($F_{20,60}$ = 16.352, $P < 0.001$, Pillai trace = 1.690, partial eta squared = 0.845). When the results for the dependent variables were considered separately, the genders differed significantly on all dependent variables ($P < 0.001$) (Table 2). When compared with men, women demonstrated significantly lower peak knee flexion angles ($P < 0.001$) and hip flexion angles ($P < 0.001$). With respect to frontal plane angles and moments, women demonstrated significantly higher peak knee abduction angles ($P < 0.001$) and average knee adductor moments ($P < 0.001$) when compared with men. Although women exhibited significantly lower average knee extensor moments ($P < 0.001$) and hip extensor moments ($P < 0.001$) compared with men, the knee–hip extensor moment ratio was significantly higher in women ($P < 0.001$).

When compared with men, women exhibited significantly lower maximum isometric strength for the hip extensors ($P < 0.001$) and knee extensors ($P < 0.001$) (Table 2). In addition, women had a significantly higher knee–hip extensor strength ratio ($P < 0.001$) compared with men. A significant positive correlation was found between the knee–hip extensor strength ratio and the knee–hip extensor moment ratio ($r = 0.41$, $P = 0.005$; Fig. 3). There was no significant correlation between the individual knee extensor or hip extensor strength variables and the knee–hip extensor moment ratio.

DISCUSSION

Consistent with previous literature (6,7,13,15–18,21,22,24), the women in our study demonstrate a kinematic and kinetic profile during landing that is suggestive of increased risk for ACL injury. Specifically, women exhibited decreased peak knee and hip flexion angles and increased knee adduction angles and knee abductor moments during the deceleration phase of the drop-jump task. Although the women in our study exhibited significantly lower knee and hip extensor moments, they had a greater knee–hip extensor moment ratio compared with men, indicating that the average knee extensor moment was greater than the average hip extensor moment during the deceleration phase of landing. In contrast, men used a more equal distribution of knee and hip extensor moments during landing as reflected by a knee–hip extensor moment ratio that was closer to 1. This finding is consistent with Sigward et al. (23), who reported that women had a significantly higher knee–hip moment ratio across different maturational groups when compared with men. In addition, Ford et al. (8) also reported a greater knee–hip extensor moment ratio in pubertal and postpubertal women compared with men.

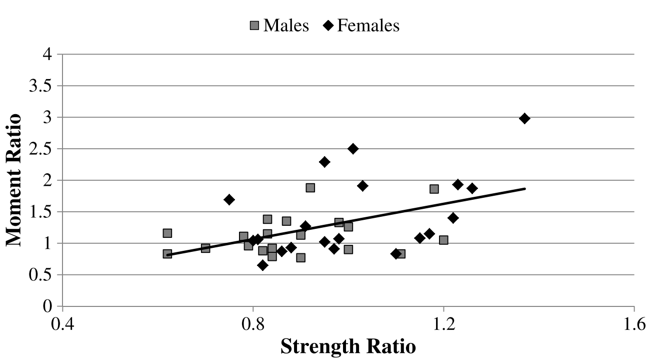

FIGURE 3—Scatterplot of the knee–hip extensor strength ratio and the knee–hip extensor moment ratio for all subjects.

With respect to strength variables, women exhibited decreased isometric strength of both knee and hip extensors when compared with men. This finding is consistent with that reported in previous studies (2,3,11,15,22). In addition, we found that sex differences in maximum isometric strength were more pronounced in the hip extensors compared with the knee extensors. On average, the hip extensors of the male group were 44% stronger than those of women. In comparison, the knee extensors of the men were only 28% stronger than those of women.

Although the women in our study demonstrated decreased torque producing capability of both the knee and hip extensors, women exhibited a significantly higher knee–hip extensor strength ratio compared with men. Specifically, women had a strength ratio close to 1, suggesting relatively equal strength of the knee and hip extensors. Conversely, men had a strength ratio <1, indicating that their hip extensors were relatively stronger than their knee extensors. More specifically, the hip extensors in the male group were 17% stronger than the knee extensors. In contrast, the hip extensors in the female group were only 2% stronger than the knee extensors.

Consistent with our proposed hypothesis, a significant association was found between the knee–hip extensor strength ratio and the knee–hip extensor moment ratio. Specifically, individuals with a strength ratio close to 1 demonstrated a biomechanical strategy characterized by greater reliance on the knee extensors relative to the hip extensors to decelerate the center of mass during landing. In turn, individuals with greater hip extensor strength relative to knee extensor strength exhibited a more equal distribution of the hip and knee extensor moments to decelerate the center of mass. However, it should be noted that the relationship between the knee–hip extensor strength ratio and the knee–hip extensor moment ratio was low, with only 17% of the variance in the knee–hip extensor moment ratio being explained by the knee–hip extensor strength ratio.

Although we observed a significant relationship between moment and strength ratios in our study, there were several limitations that could influence our findings. First, knee and hip extensor strength were measured during a maximum

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isometric concentric contraction. As the knee–hip extensor moment ratio was calculated during the deceleration phase of landing (a time when the muscles of interest are contracting eccentrically), a measurement that captured maximum eccentric strength (i.e., isokinetic testing) may have resulted in a higher correlation coefficient. Second, the knee–hip extensor moment ratio was calculated from the average of the knee and hip moments over the deceleration phase of landing, whereas the knee–hip extensor strength ratio was calculated when the knee and hip joints were in a specific position. The average strength generated over a specific range of motion, similar to the range of motion observed at the hip and knee joints during the deceleration phase of landing, may have additionally improved our model. Finally, as the landing task requires rapid torque generation upon ground contact, including an analysis of the rate of torque development of the knee and hip extensors may provide additional insight into the contribution of the knee and hip extensors to deceleration of the center of mass during landing.

On the basis of the findings of the current study, we propose that the ratio of knee extensor to hip extensor strength may, in part, underlie the biomechanical tendency of women to overuse the knee extensors during the deceleration phase of landing as compared with men. This has implications for ACL injury as the decreased use of the hip extensors during landing has been hypothesized to increase demand on the knee extensors. In turn, excessive quadriceps force may increase anterior shear loads at the knee joint and subsequent strain on the ACL (9,19). As such, intervention programs aimed at improving hip extensor strength (as opposed to knee extensor strength) may be important in reducing biomechanical risk factors associated with ACL injury in women. However, additional studies are needed to investigate whether such a training program would influence lower extremity biomechanics thought to be associated with ACL injury.

CONCLUSION

The women in the current study exhibited biomechanical tendencies thought to increase risk for ACL injury. More specifically, women exhibited higher knee extensor moments relative to hip extensor moments during the deceleration phase of landing when compared with men. The tendency of women to exhibit higher knee extensor moments relative to hip extensor moments may be explained, in part, by the relative strength of the hip and knee extensors. As such, training programs aimed at improving hip extensor strength relative to knee extensor strength should be considered as part of ACL injury prevention programs.

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