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Section I – Kinesiology

Sagittal Plane Landing Mechanics Differences between Limbs during Single- and Double-Leg Jump Landings: Implications for ACL Injury Risk in Females

by

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Given the disparity in anterior cruciate ligament (ACL) injury rates between dominant and non-dominant legs during single-leg and double-leg jump landings (SLJLs and DLJLs, respectively) in females, landing mechanics may differ across limbs and tasks. Furthermore, ACL injury risk stemming from peak kinetics is influenced by the knee flexion (KF) angle during force exertion. This study aimed to examine differences in sagittal plane landing mechanics between limbs during the SLJL and the DLJL in healthy females, with a specific focus on KF at peak kinetic values. Nineteen recreationally active females (age = 21.11 ± 3.28 yr; body height = 167.26 ± 7.26 cm, body mass = 67.28 ± 9.25 kg) were included. Sagittal plane biomechanics were recorded using a motion capture system interfaced with two force plates during the SLJL and the DLJL. Paired-sample t-tests were conducted to assess biomechanical differences between legs for both tasks. Limb asymmetries in KF at peak kinetic values were found between legs in healthy females. The non-dominant leg exhibited smaller KF at peak anterior tibial shear force (ATSF) than the dominant leg during the SLJL ($p = 0.04$, $d = 0.49$). During the DLJL, the dominant leg displayed smaller KF at peak knee extension moment (KEM) than the non-dominant leg ($p = 0.03$, $d = 0.53$). The findings indicate that females may have a higher ACL injury risk in the non-dominant leg during the SLJL, and in the dominant leg during the DLJL. These task-specific differences highlight that screenings should consider both limbs and landing types to enhance injury risk detection. Considering KF at peak kinetics when evaluating landing mechanics may enhance sensitivity and accuracy in identifying the leg at high risk for ACL injury.

Keywords: *injury prevention; knee; limb symmetry; neuromuscular control*

Introduction

Anterior cruciate ligament (ACL) injury is one of the most common orthopedic injuries in sports, with 120,000 cases reported annually in the United States—a number that continues to rise, particularly among female athletes (Kaeding et al., 2017). One possible reason is that females are more likely than males to adopt landing mechanics associated with a higher risk of ACL injuries (Krosshaug et al., 2007; Orishimo et al., 2014). Besides, disparities in ACL injury risks were observed between the dominant and non-dominant legs in females (Brophy et al., 2010; Krosshaug et al., 2007), with differences noted across various landing tasks (Krosshaug et al., 2007). Previous studies have reported that 63% of female recreational skiers and 68% of female soccer players suffered ACL injury in their non-dominant leg (Brophy et al., 2010; Ruedl et al., 2012). In addition, in female basketball players, video analysis has revealed that non-contact ACL injuries occur during both single-leg and double-leg landings, accounting for 18% and 41% of cases,

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respectively (Krosshaug et al., 2007). This discrepancy in ACL injury rates between legs and tasks may be attributed to variations in landing mechanics across limbs during various movements (Morishige et al., 2019; Wang and Fu, 2019). Despite extensive research, a widely accepted biomechanical explanation for the differences in ACL injury rates between limbs and landing tasks remains unclear.

Numerous studies have compared side-to-side differences in landing biomechanics among females; however, the findings remain inconsistent across various landing tasks (Ford et al., 2003; Mokhtarzadeh et al., 2017; Morishige et al., 2019; Wang and Fu, 2019). For instance, Wang and Fu (2019) investigated symmetry in landing biomechanics between the dominant and non-dominant legs during a single-leg landing in female athletes. Those authors found a significantly smaller knee and hip flexion excursion in the non-dominant leg compared with the dominant leg (Wang and Fu, 2019). This ''stiff'' landing pattern has been frequently linked to a higher risk of ACL injuries (Laughlin et al., 2011). In contrast, Morishige and colleagues (2019) explored leg symmetry during a double-leg drop vertical jump and reported no significant differences in sagittal plane kinematic and kinetic variables between legs in healthy females. The diverse results of the aforementioned studies may be partially attributed to methodological differences, particularly the use of different landing tasks—one examining single-leg landings and the other focusing on double-leg landings. This highlights the need to assess both landing tasks within the same study. More importantly, most prior research determining which leg presents a higher ACL injury risk has relied on individual kinematic or kinetic variables, often overlooking the importance of considering the knee joint angle under loading (Mokhtarzadeh et al., 2017; Morishige et al., 2019; Wang and Fu, 2019).

Sagittal plane mechanics play a significant role in force attenuation and ACL loading during landings (Li et al., 2005; Shimokochi and Shultz, 2008), and ACL loading is collectively affected by both the knee flexion (KF) angle and the magnitude of applied force (Englander et al., 2019; Imai et al., 2024; Li et al., 2005; Shimokochi and Shultz, 2008). Greater internal and external forces, such as peak knee extension moment (KEM), peak vertical ground reaction force (vGRF) and peak anterior tibial shear force (ATSF), have been identified as risk factors for ACL injury and directly influence the magnitude of ACL loading during landing (Leppänen et al., 2017; Lin et al., 2012; Wang et al., 2025). Additionally, under a given force, a smaller KF angle corresponds to a higher patellar tendon-tibial shaft angle (the angle between the plane of the patellar tendon and the long axis of the tibia) and the ACL elevation angle (the angle between the plane of the ACL and the tibial plateau) (Englander et al., 2019; Li et al., 2005), leading to a greater proportion of shear loading on the ACL. Since ligaments are less resistant to shear forces compared to tensile forces, a smaller KF angle increases the risk of ACL injury as the proportion of anterior tibial shear force rises (Huang et al., 2024). Therefore, apart from examining individual sagittal plane kinematic and kinetic parameters, incorporating the KF angle at the point of peak kinetics could potentially offer a more precise evaluation of ACL injury risk. However, to the best of our knowledge, no study has considered these variables when assessing ACL injury risk during landing in healthy females.

Therefore, the purpose of this study was to exploratorily investigate the differences in sagittal plane landing mechanics, with particular focus on KF at peak kinetic variables, between limbs during single- and double-leg jump landings (SLJLs and DLJLs, respectively) in healthy females. Our hypotheses were as follows: (1) during the SLJL, the non-dominant leg would exhibit a smaller KF at peak kinetics compared to the dominant leg; (2) during the DLJL, no significant differences in KF at peak kinetics would be found between the legs.

Methods

Participants

This retrospective analysis was a secondary data analysis on a subset of healthy female participants from a larger project investigating the influence of explosive quadriceps strength on landing mechanics during functional landing tasks in females with or without ACL reconstruction (ACLR) (Huang et al., 2021). Some information, such as participants' characteristics, has been published previously (Huang et al., 2021). Sample size estimation was performed using G*Power software (version 3.1.9, Heinrich-Heine University, Düsseldorf, Germany) (Faul et al., 2007) based on a previous study by Mokhtarzadeh et al. (2017), which examined the effects of leg dominance and landing height on ACL loading in female athletes. That study found that landing from a greater height exhibited landing biomechanics associated with a higher ACL injury risk, while ACL loading did not differ between the legs (Mokhtarzadeh et al., 2017). The ATSF values (dominant leg: 2.96 ± 0.78 N/body weight, non-dominant leg: 2.90 ± 0.98 N/body weight) reported in that study were used to calculate the effect size (Mokhtarzadeh et al., 2017), as these values directly contribute to ACL loading (Sakane et al., 1999). The calculated effect size was 0.68. With a significance level of 0.05 and a statistical power of 0.8, the required sample size was estimated to be 19. Accordingly, 19 participants were included in our study. All participants were recreationally active, defined by self-reported engagement in moderate to vigorous physical activity (64%–95% of the maximum heart rate) for a minimum of 150 min per week (Garber et al., 2011). The exclusion criteria of this study ruled out participants with back or lower extremity injuries restricting physical activity within the six months before study participation, as well as those with neurologic or cardiorespiratory disorders diagnosed by a physician. The Tegner Activity Scale was employed to evaluate the activity level of each participant (Briggs et al., 2009). The Institutional Review Board of the Oregon State University, Corvallis, USA, approved the study protocol (protocol code: 8800; approval date: 04 February 2019). All subjects were provided with comprehensive information about the procedures, potential risks, and the study's objectives. They then signed an informed consent form before participating in the study.

Design and Procedures

A within-subjects design was selected to compare differences in landing mechanics between legs. We proceeded to measure the participants' body height, body mass, and leg dominance after verifying eligibility. We determined the dominant leg by identifying the limb used to complete at least two of the following three tasks: 1) the leg used to first step up onto a step, 2) the leg used to kick the ball the farthest distance, and 3) the leg used to step forward when recovering from a small perturbation from behind (Hoffman et al., 1998). Before data collection, participants completed a 5-min warm-up on a stationary bike or a treadmill at submaximal intensity, followed by SLJL and DLJL tasks. For both the SLJL and DLJL tasks, participants were required to jump from a 30-cm box, positioned at 25% of their body height away from the force plates for the SLJL and 50% of their body height for the DLJL (Goerger et al., 2015). The instructions for both tasks were as follows: (1) jump using both feet, focusing on distance rather than height; (2) ensure that the tested leg's foot lands fully on a force plate for the SLJL, or each foot lands fully on a separate force plate for the DLJL; (3) perform an immediate maximal vertical jump following the landing. A trial was deemed successful if these instructions were adhered to and the movement was executed smoothly (Huang et al., 2021). For each task, a minimum of three practice trials were performed, followed by the collection of three successful trials, with at least 60 s of rest between individual trials.

The marker cluster placement was previously described in the larger study (Huang et al., 2021). The position data of the clusters were captured during two landing tasks utilizing an 8-camera motion-capture system (OptiTrack, Natural Point, Corvallis, OR, USA). These data were then streamed in real-time through a plug-in to the MotionMonitor software (Innovative Sports Training, Chicago, IL, USA). The kinematic data were also synchronized in terms of timing with kinetics data obtained from two force plates (Bertec Corp, Columbus, OH, USA), which were directly sampled using the MotionMonitor software. Following the attachment of all cluster sets, we proceeded to digitize anatomical landmarks to construct a biomechanical model using the MotionMonitor software. To create the biomechanical model, the positional data of segments collected through the clusters were integrated with virtual markers representing joint centers, established using a standardized digitization procedure. The three-dimensional coordinates of the hip joint center were determined by digitizing the anterior-superior iliac spine, utilizing the approach outlined by Bell et al. (1990). Meanwhile, the positioning of the ankle-joint and knee-joint centers occurred precisely at the halfway point between the digitized medial and lateral malleoli, and between the digitized medial and lateral femoral epicondyles, correspondingly. Local coordinate systems for the foot, the shank, the thigh, and the pelvis were established, wherein the positive x-axis indicated the anterior direction, the positive y-axis indicated the leftward direction, and the positive z-axis indicated the upward direction.

Measures

Data Processing

Kinematic and kinetic data were gathered at a sampling frequency of 150 Hz and 1500 Hz, respectively. Subsequently, a fourth-order low-pass Butterworth filter with a cut-off frequency of 12 Hz was employed to filter both kinetic and kinematic data (Bisseling and Hof, 2006; Chang et al., 2020; Pollard et al., 2020). The 12-Hz cut-off frequency, determined from residual analysis in previous studies from the same laboratory (Chang et al., 2020; Pollard et al., 2020) minimizes noise while preserving true kinematic data characteristics. Following Bisseling and Hof's (2006) recommendations, this frequency was selected for kinetic data to reduce impact inaccuracies and enhance knee moment accuracy. Kinematic data were resampled to a frequency of 1500 Hz. Kinematic data was then time-synchronized with the kinetic data using cubic spline interpolation to find missing data (Huang et al., 2021). The angles of the knee joint were established using Euler angles, derived from the shank reference frame in relation to the thigh reference frame, employing a rotation sequence of flexion-extension (y-axis), valgus-varus (x-axis), and internal-external rotation (z-axis). Net internal knee-joint moments were calculated using the MotionMonitor software, which employed filtered kinematic, kinetic, and anthropometric data (including segmental lengths, mass, center of mass locations, moments of inertia, and joint center locations), following the inverse dynamics methodology outlined by Gagnon and Gagnon (Gagnon & Gagnon, 1992).

Selection and Definition of Variables

The kinematic variables collected between the initial contact (IC) and peak KF included KF at IC, peak KF, and KF excursion, which was defined as the change in the knee flexion angle from IC to peak KF. The kinetic variables included peak vGRF, peak ATSF, peak KEM, KEM at IC, and the loading rate. Combined kinematic-kinetic variables included KF at peak vGRF, KF at peak KEM, and KF at peak ATSF. Kinetic and combined variables were identified and measured during the initial 100 ms after the IC, aiming to correspond to the critical injury time frame for ACL injuries (Cerulli et al., 2003; Koga et al., 2010). Custom computer software (LabVIEW, National Instruments, Austin, TX, USA) was used to identify the IC, when the instant vGRF was greater than 10 N. The loading rate was computed by dividing the peak vGRF by the time from the IC to the peak vGRF. The ATSF was determined by resolving the forces along the tibial plateau, following established biomechanical models. The vGRF, ATSF and the loading rate were normalized by body mass ($x \cdot kg^{-1}$), while the net internal knee-joint moments were normalized by body mass·height ($N \cdot m \ x \ [kg \cdot m]^{-1}$). The statistical analysis utilized the mean value obtained from the three trials.

Statistical Analysis

Descriptive statistics were utilized to summarize the characteristics of the participants (age, body height, body mass and Tegner Activity Scale score). The Shapiro-Wilk test and Q-Q plots were used to assess normality of data distribution. Separate paired-sample *t*-tests were performed to explore potential differences in the landing mechanics variables of interest between the non-dominant and dominant legs during SLJL and DLJL tasks. Effect sizes were classified as small ($d = 0.2$), medium ($d = 0.5$), and large ($d \geq 0.8$) (Sullivan and Feinn, 2012). All statistical analyses were conducted using SPSS (version 25.0; IBM Corp), and significant levels were set at $\alpha \leq 0.05$.

Results

Participants' demographics are presented in Table 1. The Tegner score of 6.53 reflected a moderate level of physical activity typical of recreational athletes (Tegner and Lysholm, 1985). The data from the SLJL and DLJL tasks are outlined in Tables 2 and 3, respectively. During the SLJL task, the non-dominant leg exhibited smaller KF at peak ATSF when compared to the dominant leg ($p = 0.04$, $d = 0.49$). Additionally, a similar trend was observed for KF at peak KEM, with the non-dominant leg exhibiting less KF at peak KEM compared to the dominant leg ($p = 0.05$, $d = 0.47$). Contrastingly, during the DLJL task, a significant difference was noted in KF at peak KEM between legs, revealing that the dominant leg exhibited smaller KF at peak KEM compared to the non-dominant leg (p = 0.03, d = 0.53). Across both tasks, no individual kinematic and kinetic variable differences were observed between dominant and non-dominant legs.

Table 1. Participants' demographics and anthropometrics.

	Mean (SD)
Age, years	21.11 (3.28)
Body height, cm	167.26 (7.26)
Body mass, kg	67.28 (9.25)
Tegner Activity Scale score	6.53 (1.17)

SD, standard deviation

Table 2. Landing mechanics differences between legs during single-leg jump landings.

	Dominant leg	Non-dominant leg	Mean difference	*p*	Effect size
Kinematic					
KF at IC, °	9.53 (4.27)	8.83 (5.32)	0.7	0.48	0.16
Peak KF, °	57.12 (8.11)	55.15 (8.38)	1.97	0.10	0.40
KF excursion, °	44.10 (12.49)	45.99 (13.53)	−1.89	0.30	0.26
Kinetic					
Peak vGRF, x·kg^{-1}	3.02 (0.27)	3.00 (0.26)	0.02	0.69	0.11
Peak ATSF, x·kg^{-1}	1.11 (0.14)	1.09 (0.17)	0.02	0.39	0.20
Peak KEM, N·m x [kg·m]$^{-1}$	−0.18 (0.03)	−0.16 (0.03)	−0.02	0.08	0.48
KEM at IC, N·m x [kg·m]$^{-1}$	0.02 (0.01)	0.02 (0.01)	0	0.12	<0.001
Loading rate, N·[kg·s]$^{-1}$	46.31 (6.17)	44.66 (7.46)	1.65	0.19	0.31
Combined					
KF at peak vGRF, °	37.29 (5.56)	36.49 (6.53)	0.8	0.40	0.20
KF at peak ATSF, °	48.58 (4.56)	46.66 (5.45)	1.92	0.04*	0.49
KF at peak KEM, °	48.21 (4.79)	46.17 (5.75)	2.04	0.05	0.47

* *Significant difference between legs, p < 0.05; KF, knee flexion; IC, initial contact; vGRF, vertical ground reaction force; KEM, knee extension moment; ATSF, anterior tibial shear force*

Table 3. Landing mechanics differences between legs during double-leg jump landings.

	Dominant	Non-dominant	Mean difference	*p*	Effect size
Kinematic					
KF at IC, °	16.99 (5.52)	16.84 (7.36)	3.15	0.89	0.03
Peak KF, °	82.43 (11.51)	81.99 (11.59)	0.44	0.67	0.10
KF excursion, °	65.44 (11.28)	65.44 (12.00)	0	0.81	<0.001
Kinetic					
Peak vGRF, x kg^{-1}	1.79 (0.30)	1.70 (0.27)	0.09	0.24	0.28
Peak ATSF, x kg^{-1}	0.82 (0.17)	0.79 (0.16)	0.03	0.27	0.29
Peak KEM, N·m x [kg·m]$^{-1}$	−0.154 (0.03)	−0.145 (0.02)	−0.01	0.10	0.39
KEM at IC, N·m x [kg·m]$^{-1}$	0.02 (0.01)	0.02 (0.01)	0	0.98	<0.001
Loading rate, N·[kg·s]$^{-1}$	28.15 (8.26)	26.49 (6.04)	1.66	0.35	0.22
Combined					
KF at peak vGRF, °	53.97 (7.93)	53.36 (7.75)	0.61	0.61	0.12
KF at peak ATSF, °	67.17 (5.81)	66.17 (7.23)	1	0.31	0.27
KF at peak KEM, °	65.19 (4.92)	67.05 (6.18)	−1.86	0.03*	0.53

* *Significant difference between legs, p < 0.05; KF, knee flexion; IC, initial contact; vGRF, vertical ground reaction force; KEM, knee extension moment; ATSF, anterior tibial shear force*

Discussion

The primary aim of this study was to explore differences in sagittal plane landing mechanics between limbs during the SLJL and the DLJL in healthy females, with a particular focus on knee flexion at peak kinetic values. Our study revealed that asymmetry in KF at peak kinetic values between legs existed in healthy females. Compared to the contralateral leg, the non-dominant leg exhibited smaller KF at peak ATSF during the SLJL, while the dominant leg demonstrated smaller KF at peak KEM during the DLJL. These findings support our first hypothesis but differ from the second one, indicating that the non-dominant leg exhibited landing mechanics associated with ACL injury risk during the SLJL task, while the dominant leg demonstrated such mechanics during the DLJL task.

Our study revealed that during the SLJL task females tended to land with a more extended knee when the peak load was exerted. Specifically, we observed smaller KF at peak ATSF in the non-dominant leg compared to the dominant leg. Additionally, a similar trend was noted in KF at peak KEM. While this difference did not reach statistical significance, a medium effect size was nearly attained (p = 0.05, d = 0.47). These findings indicate that the non-dominant leg demonstrates landing mechanics linked to a higher ACL injury risk during the SLJL, as a smaller KF angle corresponds to higher ACL loading under a given force (Li et al., 2005; Shimokochi and Shultz, 2008). As previously discussed, this is due to the increase in both the patellar tendon-tibial shaft angle and the ACL elevation angle with a decreasing KF angle (Englander et al., 2019; Li et al., 2005). As the knee extension force is transmitted through the patellar tendon to the tibia anteriorly, higher patellar tendon-tibial shaft and ACL elevation angles result in an increased proportion of shear loading on the ACL. Consequently, a smaller KF angle heightens ACL injury risk by increasing the proportion of ATSF, as ligaments are more susceptible to shear than tensile forces. Previous cadaveric research has investigated the effect of the KF angle on ACL loading and demonstrated that ACL loading increased as the KF angle decreased under constant quadriceps force (Li et al., 1999). With a quadriceps load of 200 N, ACL tension peaked at approximately 45 N at 15° of KF and decreased to approximately 10 N at 90° of KF (Li et al., 1999). Considering that the knee joint angle increased by a total of 75°, resulting in a 78% decrease in ACL loading (Li et al., 1999), it can be implied that for each additional degree of flexion, ACL loading decreases by approximately 1%. In our study, we observed that the non-dominant leg had approximately a 2° reduction in KF compared to the dominant leg at peak ATSF and KEM during the SLJL, while both legs exhibited similar peak ATSF and KEM values. Therefore, the ACL loading in the non-dominant leg during the SLJL is estimated to be approximately 2% higher compared to the dominant leg. However, whether a 2% difference in ACL loading is clinically meaningful requires careful consideration and interpretation.

In contrast to previous literature that revealed decreased sagittal plane knee and hip joint excursion in the non-dominant compared to the dominant leg during the SLJL (Wang and Fu, 2019), no significant differences between the legs were found in both individual kinetic and kinematic variables in the present study during the SLJL. Higher landing height and different participants' characteristic might be reasons for the inconsistency. In the Wang and Fu's (2019) study, subjects performed the SLJL from a height of 40 cm, whereas our participants performed the SLJL from a height of 30 cm. It is possible that a greater landing height increases the demands placed on the landing leg while reducing the influence of the ceiling effect. This may result in observable kinematic differences in the sagittal plane between legs (Wang and Fu, 2019). Additionally, differences in participants' characteristics may also contribute to the inconsistent results. Wang and Fu (2019) recruited healthy female collegiate soccer players with an average of 9 years of soccer experience (Wang and Fu, 2019), while our participants were recreationally active females without specialized training. In soccer, the dominant leg is predominantly utilized for kicking, whereas the non-dominant leg serves mainly as a supporting limb (Paillard and Noé, 2020). Disparities in training approaches between legs could intensify asymmetry in the lower limbs.

Lastly, differences in the methods used to define the dominant leg may also contribute to variations in results. Wang and Fu (2019) determined the dominant leg by asking participants which leg they would use to kick a ball. Our study employed a more comprehensive approach, incorporating functional tasks such as stepping up first onto a step, kicking a ball the farthest, and stepping forward to recover from a backward perturbation. This multi-task approach reduces variability and recall bias, leading to a more accurate determination of limb dominance. Taken together, different landing heights, participants' characteristics and leg dominance definitions are probable factors contributing to the observed inconsistencies in results. Further research is needed to investigate the impact of landing height and specialized sports training on lower limb symmetry in female athletes.

During the DLJL task, healthy females exhibited smaller KF at the peak KEM in the dominant compared to the non-dominant leg, and no more significant differences in other biomechanical variables between legs were found in the current study. Based on the observed reduction in angles, it was noted that the dominant leg exhibited a roughly 2° smaller KF compared to the non-dominant leg at peak KEM during the DLJL. As discussed previously, it is estimated that ACL loading in the dominant leg during the DLJL is approximately 2% higher than in the contralateral side (Li et al., 1999). These findings are partially supported by prior literature where no significant difference was found in sagittal plane kinematic and kinetic variables between the legs during the DLJL in healthy females (Morishige et al., 2019). However, despite the absence of differences in individual kinematic and kinetic variables between the legs during the DLJL in the current study, a difference was found in the KF at peak KEM. This suggests that the KF at peak KEM may be more sensitive than particular kinematic or kinetic variables in detecting asymmetries in landing biomechanics between legs. Moreover, less KF at peak KEM could be a crucial factor associated with an increased risk of ACL injury on the dominant leg in females during the DLJL. Future prospective epidemiological research should further investigate whether KF at peak kinetic values is a proper indicator to assess ACL injury risk.

The ACL injury-related landing mechanics observed in the non-dominant leg during the SLJL and in the dominant leg during the DLJL in females may be attributed to dominant-side development and compensatory mechanisms. The dominant leg indicates the preferred utilization of one leg in voluntary motor activities, and it is linked to the functional dissimilarity between the two hemispheres of the human brain (Sadeghi et al., 2000). The development of the dominant side may result from functional asymmetry due to the preferential use of a leg, or it could be attributed to structural asymmetry caused by differences in the bone and joint morphology (Stephens 2nd et al., 2005). Athletes likely utilize the preferred leg more and earlier than the non-preferred leg to absorb the landing impact, as the preferred leg is commonly stronger and may instill greater confidence (Britto et al., 2015). Thus, compared to the contralateral side, the non-dominant leg may pose a greater ACL risk during the SLJL, while dominant leg compensatory mechanisms during the DLJL may result in adverse landing mechanics and increase the risk of ACL injury. However, these differences are discernible only when the variables are related to both KF angle and peak kinetics.

From a clinical perspective, this study not only identified differences in ACL injury risk between legs during different tasks, but also highlighted the need for targeted injury prevention strategies. Previous research has shown that injury prevention programs designed to improve physical performance and landing technique can effectively reduce the ACL injury incidence (Huang et al., 2020). However, these programs often lack task-specific training tailored to optimize landing mechanics for each leg and reduce ACL injury risk in both legs. Integrating limb- and task-specific interventions may further enhance the effectiveness of injury prevention efforts.

Limitations of our study should be acknowledged. First, our study only recruited recreationally active females, thus the findings may not be extrapolated to males and competitive athletes. Nonetheless, considering that ACL injuries can also occur in individuals not engaged in high-risk activities, our findings with recreational active females are highly valuable. Secondly, our study was conducted in a controlled laboratory environment, which cannot fully replicate the real-world ecological conditions where ACL injuries frequently occur. However, our experimental design aimed to simulate as closely as possible the movements in real sports settings that could lead to ACL injury. Lastly, the dependent variables reported in our study lack error rate corrections (e.g., Bonferroni correction), which might increase the risk of Type I errors. However, this does not compromise our study's validity, as error rate adjustments are more pertinent to interventional than observational studies, and some recommendations favor an event-to-variable ratio over strict error rate corrections (Burcal et al., 2019).

Conclusions

Our results indicate that asymmetry exists in KF at peak kinetic values between the dominant and non-dominant legs in healthy females. The landing mechanics related to higher ACL injury risk varied depending on the task. A decreased KF at the peak ATSF was noted in the non-dominant leg during the SLJL, while a reduced KF at the peak KEM was observed in the dominant leg during the DLJL. Opposite ACL injury-related landing mechanics between legs in different landing tasks may stem from dominant-side development and compensatory mechanisms. The findings suggest that healthy females may have a higher risk of ACL injury in the non-dominant leg during the SLJL and in the dominant leg during the DLJL. Given the exploratory nature of our study and the absence of error rate corrections, these findings should be interpreted with caution. Future studies should give considerable attention to KF at peak kinetics values when assessing the risk of ACL injury. Injury prevention programs should integrate limb- and task-specific training to refine landing mechanics and reduce ACL injury risk in each limb.

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