Effect of Foot-Planting Strategy on Anterior Cruciate Ligament Loading in Women During a Direction Diversion Maneuver

A Musculoskeletal Modeling Approach

Jusung Lee,* MSc, Prabhat Pathak,† MSc, Siddhartha Bikram Panday,‡‡ MSc, and Jeheon Moon,§ PhD

Investigation performed at the Department of Sport Science, Korea Institute of Sport Science, Seoul, Republic of Korea

Background: Although there is a higher prevalence of noncontact anterior cruciate ligament (ACL) injuries during a direction diversion maneuver (DDM), no previous studies have reported how foot-planting strategies affect ACL loading.

Purpose: To investigate the effect of foot-planting strategies on ACL loading in women during a DDM task using a musculoskeletal modeling approach.

Study Design: Descriptive laboratory study.

Methods: A total of 13 female participants performed a DDM task, which involved running at 4.5 ± 0.2 m/s and turning left at 35° to 55° under a foot-planting strategy in 3 directions: neutral, toe-in, and toe-out. Kinematic and kinetic data were measured with the use of a 3-dimensional motion capture system and force platform to calculate variables such as joint angle, shear force, and moment. Anterior ACL and posterior ACL forces were extracted using musculoskeletal modeling.

Results: The peak anterior ACL force was significantly larger for the toe-out condition (31.29 ± 4.02 N/body weight [BW]) compared with the toe-in condition (25.43 ± 5.68 N/BW) (P = .047), with no significant difference in the neutral condition. The toe-out condition had a higher knee valgus angle (2.98° ± 4.20°; P = .041), knee shear force (10.20 ± 1.69 N/BW; P = .009), and knee internal rotation moment (–0.18 ± 0.16 N m/BW x height; P = .012) than the toe-in and neutral conditions.

Conclusion: Through musculoskeletal modeling, we were able to conclude that the toe-out condition during the DDM might result in a higher risk of ACL injuries. Athletes and sports practitioners should avoid the toe-out foot-planting strategy when participating in a sporting activity.

Clinical Relevance: Based on these findings, medical professionals and athletic coaches can gain knowledge on how foot-planting strategy affects ACL loading. Understanding the actual cause of an ACL injury can be useful for designing preventive training programs or strategies to decrease the risk of such injuries.

Keywords: anterior cruciate ligament; foot-planting strategy; musculoskeletal modeling; kinematics; kinetics

The anterior cruciate ligament (ACL) is one of the most frequent injury sites, accounting for more than 20.5% of all knee injuries with an injury rate of 6.5 per 100,000 athlete-exposures during different sporting activities. Although the most common mechanism of injury is player-to-player contact, Joseph et al reported that noncontact injuries account for 37.9% of ACL injuries, at an injury rate of 8.9 per 100,000 athletes, and the prevalence of such injuries is higher among female athletes.

Numerous studies have acknowledged the severity of ACL injuries among athletes: Around 76% of ACL injuries require surgery, which ultimately results in a decrease in the quality of life through increasing the risk of secondary injuries or disabilities, leading to financial burden. Therefore, an understanding of the mechanisms involved in the increased risk of ACL injuries is necessary.

The threshold for an ACL injury has been reported to be at 2000 N. From a biomechanical perspective, the risk of ACL injuries increases primarily when there is a smaller knee flexion angle, large ground-reaction force, large knee valgus angle and moment, and greater knee...
shear force than typically encountered. These biomechanical risk factors can increase during sporting situations when an athlete has to change direction to bypass his or her opponents suddenly, also known as a direction diversion maneuver (DDM). In this situation, the toe is likely to be directed in a neutral, toe-in, or toe-out position depending on the intended movement direction. Various biomechanical risk factors for ACL injury are associated with these 3 movement strategies. The toe-in position is postulated to increase the hip adduction angle as well as the knee internal rotation angle and moment during planting movements. In the case of the toe-out position, studies have reported that it results in an increase in the valgus angle and varus moment of the knee joint during landing and side-cutting movements. However, these studies only postulated that there would be an increase in ACL loading, without directly estimating the ACL load.

The purpose of our study was to investigate the effect of a foot-planting strategy on ACL force in a sporting movement using a musculoskeletal modeling approach. This technique has been implemented to directly estimate the ACL force and its relation to an ACL injury during running and stop-jump, and sidestep cutting. Previous studies have postulated the effect of a foot-planting strategy on ACL force and the risk of ACL injuries using a causal relationship with lower limb joint kinematics and kinetics. In our study, we used musculoskeletal modeling to estimate the ACL force directly in addition to the lower limb joint kinematics and kinetics. Hence, our approach allowed us to determine the foot-planting strategy that results in the largest ACL force and thereby the highest risk of ACL injuries. Further, we aimed to determine how the increase in ACL force changes the lower limb kinematics and kinetics to confirm the risk factors of an ACL injury according to the foot-planting strategy mentioned in previous studies. Because Teng et al reported knee valgus angle and varus moment as significant risk factors that increase the ACL load compared with other risk factors, we hypothesized that ACL loading during the toe-out condition would be significantly higher than that for the neutral and toe-in conditions.

METHODS

Participants

A total of 13 women (mean age, 24.23 ± 3.54 years; mean height, 162.77 ± 5.96 cm; mean weight, 55.31 ± 6.17 kg) participated in the study. Only female participants were selected because of their higher risk for ACL injury compared with men. The participants were right foot–dominant, had no known neurological or muscle diseases, and regularly participated in physical activity (at least twice a week). All participants signed informed consent forms regarding the procedure and the risks involved in the study, which were approved by our institutional review board.

Experimental Procedure

The participants were provided with athletic clothing and shoes before the experiment, and a period of stretching and a warm-up session were granted to minimize the risk of injuries. A total of 51 reflective markers were placed on the joints and segments of the body (see Appendix Table A1 for locations of markers). The position of the reflective markers was recorded using 18 infrared cameras combined with motion capture software (Oqus 7+ Qualisys) at a sampling frequency of 200 Hz. The participants were then asked to perform a DDM task. The task involved running toward a force platform (Type 9287BA; Kistler) from a distance of 5 m at a speed of 4.5 ± 0.2 m/s and turning left at 35° to 55° from the direction of travel once the right foot landed on the force platform. The instant of planting was defined as the point when the right foot landed on top of the force platform. The 3-directional force and moment obtained from the force platform were recorded at a sampling frequency of 1000 Hz.

The DDM task was performed for 3 conditions: when the toe position was neutral, toe-in, and toe-out from the direction of travel at the landing zone. A foot-planting position template (nonslippery tape) was pasted on top of the force platform to offer feedback regarding the direction of the toe. The neutral position was defined as 0° of rotation from the anterior line along the sagittal plane of the body, while toe-in and toe-out were defined as 30° of internal and external rotation, respectively. There were 3 trials performed for each condition, and a trial was considered successful when (1) the entire foot landed on top of the force platform and (2) the participant correctly planted her foot on the position template (Figure 1). The running speed was controlled using 2 timing gates (Micogate).
Data Analysis

The kinematic and kinetic data obtained from the motion capture system and the force platform after labeling were exported as C3D files for processing in Visual3D (V 5.01; C-Motion). Thereafter, a fourth-order, zero-lag low-pass Butterworth filter with a cutoff frequency of 15 Hz was applied to the data.12

Calculation of Kinematic and Kinetic Variables. The kinematic model was constructed based on the standing trial using Visual3D software. The midpoint of each joint was estimated, and we specified the segments and their orientation from the proximal and distal endpoints. The Cardan sequence (x-y-z) was used to define the rotation of the segments for the local coordinate system. Peak and mean values for different parameters were calculated using inverse kinematics and dynamics algorithms at the instant of planting from each trial, which was identified with a force profile in the vertical direction (Fz). For inter-participant comparability, all the calculated parameters were also normalized to body weight (BW) and height (HT). The parameters computed for analysis were as follows:

1. Ground-reaction force (N, N/BW): peak force at the instant of foot planting.
2. Center of mass (COM): based on the location of the COM of each segment expressed in meters (m), COM velocity is the first derivative of the COM expressed in meters per second (m/s).
3. COM–center of pressure (COP) displacement (m/HT): resultant vector obtained by subtracting displacement of the COP from displacement of the COM and normalized to HT.

Figure 1. (A-C) The 3 foot-planting conditions performed in the study. The images were extracted at the peak anterior ACL (aACL) load during each condition. (D) The 2-bundle ACL model used in musculoskeletal modeling. The model extracted both aACL and posterior ACL (pACL) forces. (E) The foot-planting position template for the 3 different conditions. Participants were instructed to plant at the neutral, toe-out, or toe-in position. ACL, anterior cruciate ligament.
4. Joint angle (deg): ankle (dorsiflexion/abduction/inversion), knee (valgus/flexion/external rotation), and hip (flexion/adduction/internal rotation).
5. Joint moment (N·m, N·m/BW-HT): ankle (plantarflexion/adduction/inversion), knee (extension/varus/internal rotation), and hip (extension/abduction/external rotation), with all moments computed in the study defined as internal rotation moments and normalized to HT and BW.
6. Joint force (N/BW): peak shear force and BW-normalized shear force.
7. ACL force (N, N/BW): anterior ACL (aACL) and posterior ACL (pACL) forces and BW-normalized force.

ACL Model. The musculoskeletal model to calculate the ACL load was developed using OpenSim software (Version 3.3; Stanford University). The skeleton of the model had a total of 14 segments, 23 degrees of freedom, and 92 muscles, with 2 bundles of the ACL (anterior and posterior bundles) to calculate the ACL load.13 The anterior and posterior bundles are 2 main functional bundles of the ACL that exhibit differing length-change patterns during flexion-extension of the knee (Figure 1). The anterior bundle is the primary restraint against anterior tibial translation, while the posterior bundle is the stabilizer during full extension of the knee and works to prevent rotatory loads.14,15 A detailed description of the model's properties is provided in Appendix 2.

ACL Load Calculation. The aACL and pACL forces were calculated using the method developed by Delp et al.9 The process includes body scaling, inverse kinematics, inverse dynamics, the residual reduction algorithm (RRA), computed muscle control (CMC), and forward dynamics (Figure 2). Body scaling applies the participant's body information to the musculoskeletal model modified in this study. Inverse kinematics is the process of calculating the joint angle based on coordinate data. Inverse dynamics calculates the force and moment of each joint based on the joint angle calculated through inverse kinematics. The RRA dynamically optimizes the force and moment of each joint using a residual elimination algorithm that minimizes the errors included in the measurement. CMC calculates the excitation of individual muscles by taking into account the maximum muscle force and actual measured motion and force to perform static dynamics. The information calculated through CMC goes through a static dynamics process and becomes a final model, which includes individual muscle and tendon length, active force, and passive force.9,43,44 The Hill model was used for calculating the ACL force, and the formula is as follows:25,26

\[ f_m = a_m f_{flv}(l_m, i_m) + f_{pass}(l_m) \cos(a_m), \]

where \( a_m \) refers to muscle activation, \( f_{flv} \) is the dynamic force effected by the force-length-velocity curve of the Hill model, \( l_m \) is the muscle length, \( i_m \) is the tendon velocity acting in the muscle direction, \( f_{pass} \) is the passive force, and \( a_p \) is the muscle pennation angle.34 Here, because the anterior and posterior bundles of the ACL cannot exert active force, only passive force was used for analysis.25,26 The model was validated using a standard protocol and is presented in Appendix 3 and Appendix Table A2.

Figure 2. The process of calculating the anterior cruciate ligament load using a musculoskeletal modeling approach. The \( \frac{d}{dt} \) denotes differentiation of the position and velocity to calculate velocity and acceleration, respectively.
The Orthopaedic Journal of Sports Medicine

Effect of Foot-Planting Strategy on ACL Loading

TABLE 1
Kinematic and Kinetic Variables at the Instant of Foot Planting

| Foot-Planting Strategy | Toe-In | Neutral | Toe-Out | P Value | \(\eta^2\)/Power | Post Hoc Test |
|------------------------|--------|---------|---------|---------|-----------------|--------------|
| DDM task validation    |        |         |         |         |                 |              |
| COM\(_{velocity}\), m/s | 4.54 ± 0.24 | 4.47 ± 0.36 | 4.42 ± 0.38 | .405 | 0.152/0.179 |              |
| Kinematics, deg        |        |         |         |         |                 |              |
| Ankle dorsiflexion     | 68.12 ± 8.72\(a\) | 61.60 ± 5.12\(b\) | 62.39 ± 8.83\(b\) | .001 | 0.464/0.976 | Neutral, toe-out < toe-in |
| Ankle abduction        | −1.36 ± 5.98 | −1.43 ± 5.79\(b\) | −2.72 ± 5.25\(b\) | .057 | 0.407/0.562 | Neutral < toe-out |
| Ankle inversion        | −19.13 ± 6.18\(b\) | −16.28 ± 6.65\(b\) | −12.80 ± 7.75\(b\) | .007 | 0.595/0.886 | Neutral, toe-out < toe-in |
| Knee flexion           | −32.64 ± 6.63 | −29.80 ± 4.88 | −30.81 ± 5.95 | .055 | 0.330/0.425 |              |
| Knee valgus            | 0.16 ± 3.56\(b\) | 2.28 ± 3.77\(b\) | 2.98 ± 4.20\(b\) | .004 | 0.375/0.900 | Toe-in < neutral < toe-out |
| Knee external rotation | −5.56 ± 7.34\(b\) | −11.52 ± 9.39\(b\) | −12.13 ± 9.94\(b\) | < .001 | 0.506/0.990 | Toe-in < neutral, toe-out |
| Hip flexion            | 45.98 ± 11.06 | 48.24 ± 10.46 | 49.86 ± 8.73 | .021 | 0.274/0.717 |              |
| Hip adduction          | −1.86 ± 5.10\(b\) | −7.45 ± 4.03\(b\) | −11.28 ± 7.25\(b\) | < .001 | 0.588/0.999 | Toe-in < neutral < toe-out |
| Hip internal rotation  | 13.02 ± 6.23\(b\) | 9.30 ± 7.03\(b\) | 4.88 ± 8.56\(b\) | .006 | 0.609/0.904 | Toe-out < neutral < toe-in |
| Peak shear force, N/BW |        |         |         |         |                 |              |
| Knee                  | 9.73 ± 1.65 | 9.22 ± 1.47\(b\) | 10.20 ± 1.69\(b\) | .006 | 0.611/0.907 | Neutral < toe-out |
| Joint moment, N-m/(BW×HT) |        |         |         |         |                 |              |
| Ankle plantarflexion  | −0.36 ± 0.39 | −0.20 ± 0.20 | −0.20 ± 0.23 | .146 | 0.295/0.369 |              |
| Ankle adduction       | 0.14 ± 0.14 | 0.13 ± 0.10 | 0.13 ± 0.14 | .911 | 0.008/0.063 |              |
| Ankle inversion       | −0.01 ± 0.07 | −0.02 ± 0.05 | −0.05 ± 0.04 | .093 | 0.179/0.472 |              |
| Knee extension        | 0.79 ± 0.34 | 0.66 ± 0.37 | 0.67 ± 0.40 | .183 | 0.132/0.342 |              |
| Knee varus            | 0.15 ± 0.33 | 0.15 ± 0.26 | 0.17 ± 0.29 | .909 | 0.008/0.063 |              |
| Knee internal rotation| −0.03 ± 0.16\(b\) | −0.11 ± 0.12 | −0.18 ± 0.16\(b\) | .045 | 0.430/0.605 | Toe-in < toe-out |
| Hip extension         | −1.50 ± 0.87 | −1.76 ± 0.35 | −1.90 ± 0.44 | .130 | 0.310/0.393 |              |
| Hip adduction         | −0.27 ± 0.39\(b\) | 0.19 ± 0.34\(b\) | 0.53 ± 0.53\(b\) | < .001 | 0.829/0.100 | Toe-in < neutral < toe-out |
| Hip external rotation | −0.36 ± 0.17 | −0.25 ± 0.31 | −0.24 ± 0.31 | .077 | 0.192/0.507 |              |
| ACL force, N/BW       |        |         |         |         |                 |              |
| aACL force            | 25.43 ± 5.68\(b\) | 29.61 ± 4.94 | 31.29 ± 4.02\(b\) | .005 | 0.360/0.879 | Toe-in < toe-out |
| pACL force            | 14.44 ± 2.95 | 16.81 ± 3.78 | 15.20 ± 3.20 | .136 | 0.153/0.400 |              |

\(a\)Data are reported as mean ± SD unless otherwise indicated. aACL, anterior ACL; ACL, anterior cruciate ligament; BW, body weight; COM, center of mass; DDM, direction diversion maneuver; HT, height; pACL, posterior ACL.

\(b\)Significant differences revealed after post hoc analysis.

Statistical Analysis

One-way repeated-measures analysis of variance (ANOVA) was used to evaluate the differences in the dependent variables (validation variables [COM\(_{velocity}\]), joint kinematic and kinetic variables, ACL forces) based on the foot-planting strategy (neutral, toe-in, toe-out). A post hoc test was performed using Bonferroni correction, and the statistical significance level was set at \(z = .05\). All statistical analyses were performed using SPSS 23.0 (IBM).

RESULTS

Validation of the DDM Maneuver

The mean COM\(_{velocity}\) at the instant of planting for the neutral, toe-in, and toe-out conditions was 4.47 ± 0.36, 4.54 ± 0.24, and 4.42 ± 0.38 m/s, respectively (Table 1). The mean COM\(_{velocity}\) for any of the foot-planting conditions was not significantly different from the controlled running speed: 4.5 ± 0.2 m/s \((F_{2,11} = 3.52; P < .405)\). Hence, the running velocity was effectively controlled.

Joint Kinematics, Moments, and Shear Force

Table 1 presents the joint kinematics and kinetics of the lower extremities during the DDM at the instant of planting. One-way repeated-measures ANOVA concluded that there was a significant main effect of foot-planting strategy on ankle dorsiflexion \((F_{2,11} = 10.39; P = .001)\) and inversion \((F_{2,11} = 8.086; P = .007)\), knee valgus \((F_{2,11} = 4.325; P = .004)\) and external rotation \((F_{2,11} = 8.721; P < .001)\), and hip adduction \((F_{2,11} = 10.07; P < .001)\) and internal rotation \((F_{2,11} = 8.571; P = .006)\).

The post hoc analysis showed that ankle dorsiflexion for the toe-in condition was significantly higher than that for the neutral and toe-out conditions. The ankle abduction angle for the toe-out condition was significantly higher than that for the neutral condition, and the ankle inversion angle for the toe-in condition was significantly higher than that of the neutral and toe-out conditions. In the case of the knee joint, the valgus angle was significantly higher \((1)\) for the toe-out condition versus the toe-in and neutral conditions and \((2)\) for the neutral condition versus the toe-in condition. The knee external rotation angle was significantly higher \((1)\) for the toe-out condition versus the neutral and toe-in conditions and \((2)\)
for the neutral condition versus the toe-in condition. Regarding the hip joint, the adduction angle was significantly higher (1) for the toe-out condition versus the neutral and toe-in conditions and (2) for the neutral condition versus the toe-in condition. The hip internal rotation angle was significantly higher (1) for the toe-in condition versus the neutral and toe-out conditions and (2) for the neutral condition versus the toe-out condition (Table 1).

The ANOVA results for peak knee shear force indicated that there was a significant main effect of foot-planting strategy on BW-normalized peak shear force \((F_{2,11} = 8.649; P = .006)\). The post hoc analysis revealed that the BW-normalized peak knee shear force was significantly higher for the toe-out condition compared with the neutral condition, whereas no significant difference was observed with the peak knee shear force of the toe-in condition.

The ANOVA results for lower limb joint moments indicated that there was a significant main effect of foot-planting strategy on joint moment only for knee internal rotation \((F_{2,11} = 5.234; P = .045)\) and hip abduction \((F_{2,11} = 26.63; P < .001)\). The post hoc analysis revealed that the knee internal rotation moment for the toe-out condition was significantly higher than that for the toe-in condition, whereas no significant difference was observed with the neutral condition. Further, the hip abduction angle was significantly higher (1) for the toe-out versus the toe-in and neutral conditions and (2) for the neutral versus the toe-in condition (Table 1).

**Peak ACL Force**

The ANOVA results for ACL forces (Table 1) indicated that there was a significant main effect of foot-planting strategy only on normalized peak aACL force \((F_{2,11} = 4.133; P = .005)\). The post hoc analysis revealed that the normalized peak aACL force was significantly higher for the toe-out condition compared with the toe-in condition.

**DISCUSSION**

The purpose of the study was to investigate the effect of foot-planting strategy (neutral, toe-in, and toe-out) on ACL loading in women during a DDM task using musculoskeletal modeling. We hypothesized that because of kinematic and kinetic differences during the toe-out condition, the ACL load would be larger compared with that in the other foot-planting conditions. Our hypothesis was confirmed, as the peak aACL force was significantly larger for the toe-out condition compared with the toe-in condition, with no significant differences with the neutral condition.

Sporting movements involve phases of rapid deceleration, such as the instant of planting during a DDM and jump. The sudden deceleration causes changes in kinematics and kinetics, particularly at the knee joint; these include increases in flexion and valgus angles, shear force, and varus and internal rotation moments, which increase the risk of ACL injuries.\(^{41,48}\) The risk further increases when there is a change in the foot direction during planting.\(^{45}\) Hence, a quantitative understanding of these risk factors can provide guidelines for preventive mechanisms for ACL injury. Previous studies have provided a causal relationship between joint kinematics and kinetics and the risk of ACL injuries according to the foot-planting strategy. Tran et al\(^{45}\) and Teng et al\(^{42}\) have provided contradicting views regarding the reason behind the higher risk of ACL injuries during toe-in and toe-out foot-planting strategies, respectively. The musculoskeletal modeling technique used in the current study was novel in that it allowed us to estimate the ACL forces. The significantly higher ACL force and consequent increase in knee shear force and knee internal rotation moment that we found regarding the toe-out condition indicate that this particular foot-planting strategy during a DDM has the highest risk of ACL injuries.

The kinetic chain at the planting phase initiates from the foot and the ankle, which then transfers the forces to the knee joint.\(^{1}\) Dorsiflexion of the ankle joint has been reported to be positively correlated with knee flexion displacement and restriction in ankle range of motion in the transverse plane; that is, a larger dorsiflexion angle increases the valgus angle of the knee joint.\(^{15,28}\) Hence, the larger dorsiflexion observed during the toe-out foot-planting strategy compared to the toe-in condition had a kinematic chain effect on the proximal joints, specifically increasing the valgus angle and ACL force (Table 1).

A study conducted by Donnelly et al\(^{11}\) reported that the dynamics of force distribution varies depending on foot-planting strategy (forefoot vs rearfoot). Those authors reported that forefoot planting requires a large dynamic force on the ankle joint, while rearfoot planting has a large dynamic force on the knee joint, which increases the risk of ACL injuries. Particularly, rearfoot planting during a DDM has been reported to increase knee joint extension and valgus angle, thereby increasing the ACL load.\(^{6}\) The DDM performed in this study was intended to observe these mechanical changes to the knee joint load by performing rearfoot planting.

The position of the upper body in the sagittal plane, that is, forward leaning, decreases the possibility of ACL injury by increasing the shock absorption volume and flexion angle of the knee joint as well as reducing the activity of the quadriceps.\(^{37}\) Although we did not measure the trunk angle that corresponds to a forward-leaning posture, we found no significant differences in hip flexion angle between the 3 foot-planting strategies. Nevertheless, hip adduction angle, hip internal rotation angle, and hip abduction moment were statistically different among all 3 conditions. From a similar perspective, a study on sidestep cutting reported that a change in whole-body COM to the medial and forward direction could significantly reduce the valgus moment of the knee joint and decrease the risk of ACL injury.\(^{15}\) Chaudhari et al\(^{19}\) reported that the movement of the arm and upper body position during planting could also affect knee varus moment. Therefore, if we assume that there is a
shift in the COM that coincides with rotation of the torso, the braking force changes the direction of the COM and contributes to maintaining a constant speed, preventing the body from excessive rotation about the vertical axis when performing a DDM. This may explain the consistency of the DDM. The result of COM-COP displacement in the planting phase confirms the different foot-planting conditions.

Opposing views have been formulated regarding the biomechanical risk factors for ACL injury due to foot direction during planting. The risk factors for toe-in foot planting have been reported as increased hip adduction and knee internal rotation angles and moments and decreased hip and knee flexion angles. In the case of the toe-out foot planting, the biomechanical risk factors are increased knee valgus angles and varus moments. Fukuda et al and Hewett et al have shown through cadaveric and clinical studies that larger valgus angles increase anterior tibial translation and consequently increase the ACL force. In our study, we directly estimated the ACL forces and also found that increased valgus angle during a DDM with toe-out foot planting resulted in significantly higher ACL forces. Further, we found that there was an increase in knee shear forces during toe-out foot planting, which is known to result in greater strain on the anteromedial bundle of the ACL. Hence, there was a higher risk of ACL injury during the DDM with the toe-out-foot-planting strategy because of significantly larger ACL forces, knee valgus angles, and shear forces.

Avoiding toe-out planting during sporting situations can be challenging, as an ACL injury occurs within 50 milliseconds after ground contact and is faster than the time (120 ~ 140 milliseconds) required by our nervous system to generate the reflex mechanisms to protect the knee. Therefore, ACL injury prevention programs must include not only targeted exercises but also automatic movements that resemble complex unanticipated events observed on the field specific to the type and level of sport, that is, elements of anticipation, perturbation, attention, and visual-motor control.

Limitations

There are several limitations to our study. First, our sample size was quite small (N = 13), and our study was limited to a single sporting movement (the DDM). A future study with a larger sample size and the inclusion of various sporting movements (stop-jump, drop landing, sidestep cutting, etc.) might enhance the generalizability of our results. Second, in our study, we focused on noncontact movements because of the difficulty of simulating dynamic contact movements in a controlled experimental setting. Finally, the participants in our study had no known history of ACL injury, so the kinematic and kinetic factors that increase the risk of ACL injury might be different than for those with a history of ACL injuries. Hence, our results should be used to develop ACL injury prevention protocols and training programs for athletes with no history of ACL injury.

CONCLUSION

This is the first study to use musculoskeletal modeling to estimate the ACL forces for 3 foot-planting strategies (toe-in, neutral, and toe-out) during a DDM. Compared with previous studies, the direct estimation of the ACL forces allowed us to define the relationship between foot-planting strategies and ACL injury; higher ACL forces resulted in a higher risk of injury. Further, the results of our study showed that for the toe-out condition there was a significant increase in knee valgus angle, which is a significant risk factor for ACL injury. No significant differences were observed for knee varus moment, however there were also increases in knee shear force, knee internal rotation moment, and ACL force, which allowed us to conclude that the toe-out foot direction during a DDM might lead to a higher risk of ACL injury. These results can be used by athletes, coaches, and medical professionals to create a preventive training program or strategy to decrease the risk of initial ACL injury.

REFERENCES

1. Ali N, Rouhi G. Barriers to predicting the mechanisms and risk factors of non-contact anterior cruciate ligament injury. Open Biomed Eng J. 2010;4:178-189.
2. Benjaminse A, Gokeler A, Dowling AV, et al. Optimization of the anterior cruciate ligament injury prevention paradigm: novel feedback techniques to enhance motor learning and reduce injury risk. J Orthop Sports Phys Ther. 2015;45(3):170-182.
3. Boden BP, Dean GS, Feagin JA Jr, Garrett WE Jr. Mechanisms of anterior cruciate ligament injury. Orthopedics. 2000;23(6):573-578.
4. Chappell JD, Creighton RA, Giuliani C, Yu B, Garrett WE. Kinematics and electromyography of landing preparation in vertical stop-jump: risks for noncontact anterior cruciate ligament injury. Am J Sports Med. 2007;35(2):235-241.
5. Chaudhari AM, Hearn BK, Andriacchi TP. Sport-dependent variations in arm position during single-limb landing influence knee loading: implications for anterior cruciate ligament injury. Am J Sports Med. 2005;33(6):824-830.
6. Cortes N, Onate J, Abrantes J, et al. Effects of gender and foot-landing techniques on lower extremity kinematics during drop-jump landings. J Appl Biomech. 2007;23(4):289-299.
7. Dai B, Mao D, Garrett WE, Yu B. Anterior cruciate ligament injuries in soccer: loading mechanisms, risk factors, and prevention programs. Journal of Sport and Health Science. 2014;3(4):299-306.
8. Dai B, Mao M, Garrett WE, Yu B. Biomechanical characteristics of an anterior cruciate ligament injury in javelin throwing. Journal of Sport and Health Science. 2015;4(4):333-340.
9. Delp SL, Anderson FC, Arnold AS, et al. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans Biomed Eng. 2007;54(11):1940-1950.
10. Dempsey AR, Lloyd DG, Elliott BC, et al. The effect of technique change on knee loads during sidestep cutting. Med Sci Sports Exerc. 2007;39(10):1765-1773.
11. Donnelly CJ, Chinnasee C, Weir G, Sasimontonkul S, Alderson J. Joint dynamics of rear- and fore-foot unplanned sidestepping. J Sci Med Sport. 2017;20(1):32-37.
12. Donnelly CJ, Lloyd DG, Elliott BC, Reinbolt JA. Optimizing whole-body kinematics to minimize valgus knee loading during sidestepping: implications for ACL injury risk. J Biomech. 2012;45(8):1491-1497.
13. Dorn TW, Lin Y-C, Pandy MG. Estimates of muscle function in human gait depend on how foot-ground contact is modelled. Comput Methods Biomech Biomed Engin. 2012;15(6):657-668.

14. Edwards A, Bull AM, Amis AA. The attachments of the anteromedial and posterolateral fibre bundles of the anterior cruciate ligament. Knee Surg Sports Traumatol Arthrosc. 2007;15(12):1414-1421.

15. Fong CM, Blackburn JT, Norcross MF, McGrath M, Padua DA. Ankle-dorsiflexion range of motion and landing biomechanics. J Athl Train. 2011;46(1):5-10.

16. Fukuda Y, Woo SLY, Loh JC, et al. A quantitative analysis of valgus torque on the ACL: a human cadaveric study. J Orthop Res. 2003; 21(6):1107-1112.

17. Gokeler A, Seil R, Kerkhoffs G, Verhagen E. A novel approach to enhance ACL injury prevention programs. J Exp Orthop. 2018;5(1):22.

18. Granan LP, Bahr R, Steindal K, Furnes O, Engebretsen L. Development of a national cruciate ligament surgery registry: the Norwegian National Knee Ligament Registry. Am J Sports Med. 2008;36(2):308-315.

19. Grooms DR, Onate JA. Neuroscience application to noncontact anterior cruciate ligament injury prevention. Sports Health. 2016;8(2):149-152.

20. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sports Med. 2005;33(4):492-501.

21. Hopkins JT, Brown TN, Christensen L, Palmieri-Smith RM. Deficits in peroneal ankle instability. J Orthop Res. 2009;27(12):1541-1546.

22. Jindrich DL, Besier TF, Lloyd DG. A hypothesis for the function of braking forces during running turns. J Biomech. 2006;39(9):1611-1620.

23. Joseph AM, Collins CL, Henke NM, et al. A multisport epidemiologic comparison of anterior cruciate ligament injuries in high school athletes. J Athl Train. 2013;48(6):810-817.

24. Kaeding CC, Pedroza AD, Reinke EK, et al. Risk factors and predictors of subsequent ACL injury in either knee after ACL reconstruction: prospective analysis of 2488 primary ACL reconstructions from the MOON cohort. Am J Sports Med. 2015;43(7):1583-1590.

25. Kar J, Quesada PM. A musculoskeletal modeling approach for estimating anterior cruciate ligament strains and knee anterior-posterior shear forces in stop-jumps performed by young recreational female athletes. Ann Biomed Eng. 2013;41(2):338-348.

26. Kar J, Quesada PM. A numerical simulation approach to studying anterior cruciate ligament strains and internal forces among young recreational women performing valgus inducing stop-jump activities. Ann Biomed Eng. 2012;40(8):1679-1691.

27. Koga H, Nakamae A, Shima Y, et al. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. Am J Sports Med. 2010;38(11):2218-2225.

28. Krosshaug T, Nakamae A, Boden BP, et al. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. Am J Sports Med. 2007;35(3):359-367.

29. Lin C-F, Liu H, Gros MT, et al. Biomechanical risk factors of non-contact ACL injuries: a stochastic biomechanical modeling study. Journal of Sport and Health Science. 2012;1(1):36-42.

30. Luc B, Gribble PA, Pietrosimone BG. Osteoarthritis prevalence following anterior cruciate ligament reconstruction: a systematic review and numbers-needed-to-treat analysis. J Athl Train. 2014;49(6):806-819.

31. Mall NA, Chalmers PN, Moric M, et al. Incidence and trends of anterior cruciate ligament reconstruction in the United States. Am J Sports Med. 2014;42(10):2363-2370.

32. McLean SG, Huang X, Su A, Van Den Bogert AJ. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. Clin Biomech (Bristol, Avon). 2004;19(8):828-838.

33. McLean SG, Huang X, van den Bogert AJ. Investigating isolated neuromuscular control contributions to non-contact anterior cruciate ligament injury risk via computer simulation methods. Clin Biomech (Bristol, Avon). 2008;23(7):926-936.

34. Moon J, Kim H, Lee J, Panday SB. Effect of wearing a knee brace or sleeve on the knee joint and anterior cruciate ligament force during drop jumps: a clinical intervention study. Knee. 2018;25(6):1009-1015.

35. Petersen W, Zanto T. Anatomy of the anterior cruciate ligament with regard to its two bundles. Clin Orthop Relat Res. 2007;454:35-47.

36. Samaan MA, Ringleb SI, Bawab SY, Greska EK, Weinhandl JT. Anterior cruciate ligament (ACL) loading in a collegiate athlete during sidestep cutting after ACL reconstruction: a case study. Knee. 2016;23(4):74-752.

37. Shimokochi Y,AMBeggaonkar JP, Meyer EG, Lee SY, Shultz SJ. Changing sagittal plane body position during single-leg landings influences the risk of non-contact anterior cruciate ligament injury. Knee Surg Sports Traumatol Arthrosc. 2013;21(4):888-897.

38. Shin CS, Chaudhari AM, Andriacchi TP. The influence of deceleration forces on ACL strain during single-leg landing: a simulation study. J Biomech. 2007;40(5):1145-1152.

39. Sigward SM, Ota S, Powers CM. Predictors of frontal plane knee excursion during a drop land in young female soccer players. J Orthop Sports Phys Ther. 2008;38(11):661-667.

40. Tashman S, Kolowich P, Collon D, Anderson K, Anderst W. Dynamic function of the ACL-reconstructed knee during running. Clin Orthop Relat Res. 2007;454:56-73.

41. Taylor KA, Terry ME, Uutturkar GM, et al. Measurement of in vivo anterior cruciate ligament strain during dynamic jump landing. J Biomech. 2011;44(3):365-371.

42. Teng P, Kong P, Leong K. Effects of foot rotation positions on knee valgus during single-leg drop landing: implications for ACL injury risk reduction. Knee. 2017;24(3):547-554.

43. Thelen DG, Anderson FC. Using computed muscle control to generate forward dynamic simulations of human walking from experimental data. J Biomech. 2006;39(6):1107-1115.

44. Thelen DG, Anderson FC, Delp SL. Generating dynamic simulations of movement using computed muscle control. J Biomech. 2003;36(3):321-328.

45. Tran AA, Gatewood C, Harris AH, Thompson JA, Dragoo JL. The effect of foot landing position on biomechanical risk factors associated with anterior cruciate ligament injury. J Exp Orthop. 2016;3(1):13.

46. Weinhandl JT, Earl-Boehm JE, Ebersole KT, et al. Anticipatory effects on anterior cruciate ligament loading during sidestep cutting. Clin Biomech (Bristol, Avon). 2013;28(8):655-663.

47. Weinhandl JT, Earl-Boehm JE, Ebersole KT, et al. Reduced hamstring strength increases anterior cruciate ligament loading during anticipated sidestep cutting. Clin Biomech (Bristol, Avon). 2014;29(7):752-759.

48. Yu B, Garrett WE. Mechanisms of non-contact ACL injuries. Br J Sports Med. 2007;41(suppl 1):i47-i51.
APPENDIX 1

TABLE A1
Marker Locations During Static and Dynamic Trials

| Participant No. | Marker Location      | Participant No. | Marker Location      |
|-----------------|----------------------|-----------------|----------------------|
| 1               | Left acromion        | 28              | Right acromion       |
| 2               | Left ankle: lateral  | 29              | Right ankle: lateral |
| 3               | Left ankle: medial   | 30              | Right ankle: medial  |
| 4               | Left ASIS            | 31              | Right ASIS           |
| 5               | Left biceps          | 32              | Right biceps         |
| 6               | Left elbow           | 33              | Right elbow          |
| 7               | Left heel            | 34              | Right heel           |
| 8               | Left knee: lateral   | 35              | Right knee: lateral  |
| 9               | Left knee: medial    | 36              | Right knee: medial   |
| 10              | Left midfoot: lateral| 37              | Right midfoot: lateral|
| 11              | Left midfoot: superior| 38             | Right midfoot: superior|
| 12              | Left shank: front    | 39              | Right shank: front   |
| 13              | Left shank: rear     | 40              | Right shank: rear    |
| 14              | Left shank: upper    | 41              | Right shank: upper   |
| 15              | Left temple          | 42              | Right temple         |
| 16              | Left thigh: front    | 43              | Right thigh: front   |
| 17              | Left thigh: rear     | 44              | Right thigh: rear    |
| 18              | Left thigh: upper    | 45              | Right thigh: upper   |
| 19              | Left toe: lateral    | 46              | Right toe: lateral   |
| 20              | Left toe: medial     | 47              | Right toe: medial    |
| 21              | Left toe: tip        | 48              | Right toe: tip       |
| 22              | Left upper arm: rear | 49              | Right upper arm: rear|
| 23              | Left wrist: lateral  | 50              | Right wrist: lateral |
| 24              | Left wrist: medial   | 51              | Right wrist: medial  |
| 25              | Sternum              |                 |                      |
| 26              | Top of head          |                 |                      |
| 27              | V. sacrum            |                 |                      |

*a* ASIS, anterior superior iliac spine.

APPENDIX 2

ACL Model Properties

The model considered flexion/extension (range, $-120^\circ$ to $+10^\circ$), adduction/valgus (range, $-15^\circ$ to $+15^\circ$), and external/internal rotation (range, $-30^\circ$ to $+40^\circ$) of the knee.\textsuperscript{25,26,34} The x-, y-, and z-direction position of the anterior bundle of the ACL was $-0.718$, $-0.4004$, and $0.407$ cm, respectively, from the lower femur. With respect to the upper portion of the tibia, the x-, y-, and z-direction position of the anterior bundle of the ACL was $1.657$, $-3.009$, and $0.074$ cm, respectively. The stiffness, length, and strain rate were $1500$ N, $3.23$ cm, and $0.02$, respectively. The x-, y-, and z-direction position of the posterior bundle of the ACL was $-1.495$, $-4.098$, and $0.999$ cm, respectively, from the lower femur. With respect to the upper portion of the tibia, the x-, y-, and z-direction position of the anterior bundle of the ACL was $0.250$, $-3.250$, and $0.000$ cm, respectively. The stiffness, length, and strain rate were $1600$ N, $2.47$ cm, and $0.01$, respectively.

APPENDIX 3

Validation of the ACL Model

For verification and validation of the musculoskeletal models and simulations of movement, the residual force values of the residual reduction algorithm and computed muscle control were calculated. Our results were compared with threshold values presented in OpenSim software used for evaluating body simulations of walking and running, and the calculated values were graded as \textit{good}, \textit{okay}, or \textit{bad} based on the validation criteria. The threshold value for good, okay, and bad in the case of force are $0$-10 N, $10$-25 N, and more than $25$ N, respectively, whereas, in the case of moment are $0$-50 N.m, $50$-75 N.m, and more than $75$ N.m, respectively. Results are presented in Table A2.
| Threshold                              | Toe-In             | Neutral            | Toe-Out            |
|---------------------------------------|--------------------|--------------------|--------------------|
| Root mean square for RRA              |                    |                    |                    |
| Residual force, N                     | 9.87 ± 7.14 (okay) | 9.40 ± 5.68 (okay) | 9.30 ± 4.88 (okay) |
| Residual moment, N·m                  | 25.22 ± 16.13 (good)| 23.03 ± 9.12 (good)| 26.69 ± 15.22 (good)|
| Root mean square for CMC              |                    |                    |                    |
| Residual force, N                     | 12.06 ± 7.58 (okay)| 10.52 ± 7.44 (okay)| 10.37 ± 5.16 (okay)|
| Residual moment, N·m                  | 46.69 ± 58.56 (okay)| 35.49 ± 25.21 (okay)| 32.83 ± 19.48 (okay)|

*CMC, computed muscle control; RRA, residual reduction algorithm.