

Knee Joint Kinematics and Kinetics During a Lateral False-Step Maneuver

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Context: Cutting maneuvers have been implicated as a mechanism of noncontact anterior cruciate ligament (ACL) injuries in collegiate female basketball players.

Objective: To investigate knee kinematics and kinetics during running when the width of a single step, relative to the path of travel, was manipulated, a lateral false-step maneuver.

Design: Crossover design.

Setting: University biomechanics laboratory.

Patients or Other Participants: Thirteen female collegiate basketball athletes (age = 19.7 ± 1.1 years, height = 172.3 ± 8.3 cm, mass = 71.8 ± 8.7 kg).

Intervention(s): Three conditions: normal straight-ahead running, lateral false step of width 20% of body height, and lateral false step of width 35% of body height.

Main Outcome Measure(s): Peak angles and internal moments for knee flexion, extension, abduction, adduction, internal rotation, and external rotation.

Results: Differences were noted among conditions in peak knee angles (flexion [$P < .01$], extension [$P = .02$], abduction

[$P < .01$], and internal rotation [$P < .01$]) and peak internal knee moments (abduction [$P < .01$], adduction [$P < .01$], and internal rotation [$P = .03$]). The lateral false step of width 35% of body height was associated with larger peak flexion, abduction, and internal rotation angles and larger peak abduction, adduction, and internal rotation moments than normal running. Peak flexion and internal rotation angles were also larger for the lateral false step of width 20% of body height than for normal running, whereas peak extension angle was smaller. Peak internal rotation angle increased progressively with increasing step width.

Conclusions: Performing a lateral false-step maneuver resulted in changes in knee kinematics and kinetics compared with normal running. The differences observed for lateral false steps were consistent with proposed mechanisms of ACL loading, suggesting that lateral false steps represent a hitherto neglected mechanism of noncontact ACL injury.

Key Words: noncontact injury mechanisms, anterior cruciate ligament, sidestep cutting

Key Points

- In collegiate female basketball players, lateral false-step maneuvers were associated with greater peak flexion, abduction, and internal rotation angles; greater peak abduction, adduction, and internal rotation moments; and greater between-trials variability in angles and moments at the knee.
- Peak knee abduction angles, internal rotation angles, and adduction moments were greater for a wider lateral false step.
- Similar to sidestep cutting, a lateral false step during running may also place the anterior cruciate ligament at risk of injury; the risk may rise as step width increases.

The incidence of anterior cruciate ligament (ACL) injuries is considerable in comparison with other knee injuries.^{1,2} The possible need for surgical reconstruction to restore knee stability, substantial time lost from athletic participation, and long-term consequences of osteoarthritis after ACL injury substantiate the importance of determining which factors contribute to this injury.^{3,4} Of particular interest are the well-documented greater ACL injury rates in women than in men in matched sporting activities, such as soccer and basketball.^{1,2,5}

Noncontact injury mechanisms account for 70% to 78% of all ACL ruptures in the athletic population.^{6,7} The primary mechanism of noncontact injury is a rapid deceleration, typically associated with a double-leg or single-leg landing from a jump,^{7,8} a sudden stop while running,^{7,8} or an evasive running maneuver in which the athlete quickly redirects the path of travel through a plant and cut.^{6,9,10} One type of evasive maneuver, sidestep

cutting, has been extensively investigated over the past decade.^{10–18} The typical research paradigm for sidestep cutting involves straight-ahead running abruptly redirected at a prescribed angle between 30° and 90° from the original direction of travel. Sidestep cutting results in different knee kinematics than normal running.^{11–13,19} Additionally, the resulting knee kinematics and kinetics of sidestep cutting are consistent with proposed mechanisms of increased loading of the ACL.^{19,20}

The typical sidestep cutting maneuver is not the only agility movement used to evade or “fake” an opponent in athletics. The lateral false step (LFS) is often used as an alternative to sidestep cutting to give the appearance of a commitment to change direction. The LFS is performed by extending the first step of a running stride in a lateral direction and, although the athlete appears to be committed to a new running direction, he or she immediately returns to the original path of travel. Therefore, the most

functional difference between the LFS and sidestep cut is that the former does not change the athlete's direction of travel as does the latter. The LFS involves altering foot placement, resulting in increased mediolateral velocities, thus likely affecting knee kinematics and kinetics and potentially increasing the risk for knee injury. Changes in peak values or in between-trials variability would be of interest, as either could affect peak ACL loading. Despite how frequently the LFS is used in athletics, little is known about its effects on the knee.

Our purpose was to compare knee joint kinematics and kinetic patterns during the stance phase of normal running to those for LFSs of 20% and 35% of body height (BH). These LFS widths were selected as being both realistic and acceptably safe, based on preliminary investigation. Peak joint angles, peak joint moments, and average between-trials variability in the joint angles and moments were considered in 3 dimensions. The aim of this investigation of the LFS maneuver was to better understand the net loading of the knee and the potential risks for ACL disruption during this unstudied change-of-direction task.

METHODS

Participants

Fourteen healthy women from local university and community college varsity basketball programs volunteered to participate in this study. Data from 1 person were discarded because of a data collection error, resulting in 13 participants being studied (age = 19.7 ± 1.1 years, height = 172.3 ± 8.3 cm, mass = 71.8 ± 8.7 kg). Volunteers completed a demographic and health history questionnaire, which was used for inclusion decisions. Exclusionary criteria included a past history of any reconstructive surgical procedure for treatment of an injury in the past year; current receipt of medical care for an injury to the upper or lower extremity that prevented participation in regular conditioning and practice sessions; and an injury to the upper or lower extremity resulting in time loss from participation exceeding 3 weeks during the previous 6 months. Players were of moderately equivalent sport maturation experience, having been involved in at least 3 years of high school athletics. Three athletes had undergone ACL reconstruction in the past 1.5 to 2 years (to the nondominant limb in 2 patients) and had since returned to full athletic participation without the use of external bracing. This study received approval by the university's institutional review board, and all participants provided written informed consent before data collection.

Testing Procedures

Volunteers performed repeated trials of running in a laboratory setting under 3 conditions: normal running (NRUN), lateral false-step width of 20% of measured BH (LFS20), and lateral false-step width of 35% of measured BH (LFS35). The LFSs were performed by the nondominant limb. Limb dominance was determined by the results of the ball-kick, step-up, and balance-recovery tests.²¹ The preferred limb used for kicking or stepping in at least 2 of the 3 tasks was designated as the dominant limb.

The LFSs were performed onto a pair of force plates (model 4060-08; Bertec Corp, Columbus, OH) located near

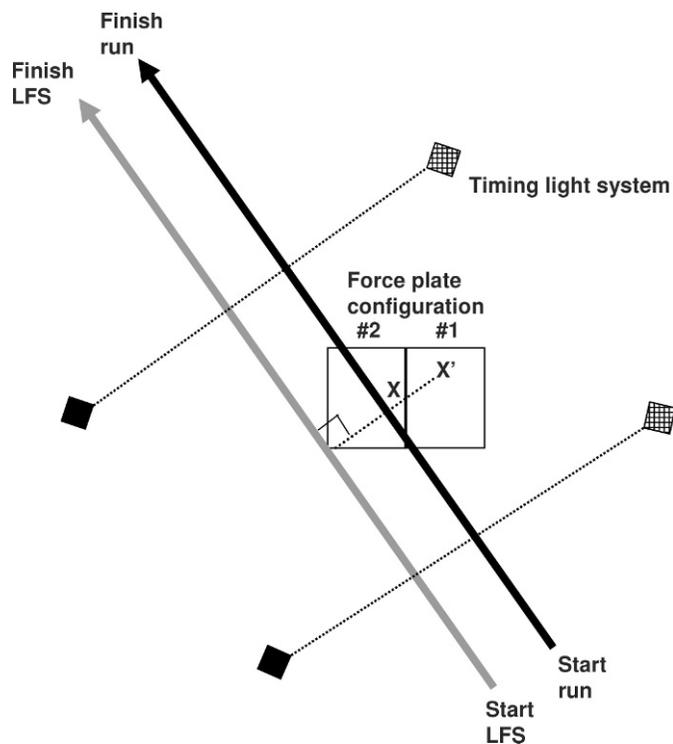


Figure 1. Setup for data acquisition for running and lateral false-step (LFS) trials for a left-foot-dominant participant (ie, LFS performed with the right limb). Target stepping foot placements for LFS of 20% and 35% body height are indicated by X and X', respectively.

the midpoint of the 10-m running path. The specified running path and target foot locations for the LFS at each width were marked with adhesive tape (Figure 1). Timing lights placed 1.83 m on either side of the force plates were used to monitor running speed. In addition, an 8-camera optical motion capture system (model 612 with M2 cameras; Vicon Motion Systems Inc, Lake Forest, CA) was used to record the 3-dimensional kinematics of the participants at 120 Hz, in synchrony with the acquisition of kinetic data from the force plates at 1080 Hz. For motion capture, volunteers wore form-fitting clothing, and 16 external reflective markers (9-mm diameter) were affixed bilaterally at the fifth metatarsal, lateral ankle, heel, leg, knee, thigh, and anterior and posterior superior iliac spines.²² During the testing, athletes wore the shoes they were currently using for their organized basketball activities.

Before data collection, participants warmed up with 5 minutes of self-paced treadmill running. Each player performed 3 or 4 practice trials for each of the 3 test conditions. This aided in determining the starting location of the run and facilitated the volunteer's hitting the force plate consistently and correctly. In final preparation, a static calibration trial of segmental orientation for the kinematic model was collected with the athlete standing in an anatomic neutral position with the arms at 90° of abduction and with the feet parallel and hip-width apart.

Ten trials were obtained for each of the 3 conditions. All participants completed NRUN trials first, followed by LFS20 or LFS35 conditions in counterbalanced order. Players were required to contact the corresponding mark on the force plate with the nondominant foot without a

break in stride and while maintaining the desired speed of $3.5 \text{ m}\cdot\text{s}^{-1} \pm 10\%$. In addition, during the steps immediately preceding and following the step onto the force plate, the dominant foot had to contact the floor to its side of the tape marking the running path. This served to ensure that the LFS width would closely approximate the desired distance. Trials not meeting these criteria were considered unacceptable and were repeated. For their recovery time, the athletes walked back to the starting position between trials.

At the end of data collection, knee and ankle widths were measured with a joint anthropometer (model 01291; Lafayette Instrument Co, Lafayette, IN) and shoe length was measured with a tape measure. Body weight (BW) was also measured. Measurements were obtained and recorded by the same researcher for all volunteers. Data were collected in a single 1.5-hour testing session. To minimize fatigue effects, all athletes participated in the testing before any conditioning or practice sessions on that same day.

Data Analysis

Three-dimensional kinematic and kinetic analyses of the nondominant limb were performed using BodyBuilder software (Vicon Motion Systems Inc), with the hip, knee, and ankle joints each assigned 3 degrees of freedom. The paths of the markers were reconstructed in 3 dimensions from the raw kinematic data and were filtered with a fourth-order, zero-lag Butterworth low-pass filter with a cut-off frequency of 16 Hz, as determined through a residual analysis.²³ Marker positions from the static trial, combined with anthropometric measures, were used to determine a set of transformations for identifying joint center locations and body segment orientations. We used these transformations to compute the corresponding joint locations and segment orientations during the running trials. Knee joint angles were determined as a sequence of flexion-extension, abduction-adduction, and then internal-external rotation of the leg about its anatomical axes relative to the thigh.²⁴ Internal-external rotation during the static trial was defined as zero. From the kinematic and force data, the internal moments acting at the knee during the step on the force plates were computed using inverse dynamics. Segmental inertial parameters were determined as outlined by de Leva²⁵ for females. Knee flexion-extension, abduction-adduction, and internal-external rotation moments were expressed relative to the anatomical axes of the leg. All moments were normalized to BH and BW.²⁶ The period of stance was identified from the force plate data.

Peak joint angles and moments during stance phase of the LFS, or of the corresponding step on the force plates for NRUN, were determined for the ipsilateral (ie, nondominant) knee. The percentage of stance at which each peak value occurred was also determined. Peak values were averaged across all trials of a given condition for each participant. In addition, the average between-trials variability in the knee angles and moments for the stance phase of interest was determined across all trials of a given condition. This procedure, adapted from Winter,²⁷ entailed first normalizing each angle and moment time course to 21 samples, at intervals of 5% of stance, using cubic spline

Table 1. Running Trial Characteristics

Condition	Average Increase in Step Width, cm	Average Speed, $\text{m}\cdot\text{s}^{-1}$
Normal running	NA	3.48 ± 0.11
Lateral false step		
20% of body height	34.5 ± 1.7	3.60 ± 0.17
35% of body height	60.3 ± 2.9	3.54 ± 0.15
All conditions	NA	3.54 ± 0.03

Abbreviation: NA, not applicable.

interpolation. Average variability across stance was then expressed for each player as the root mean square value of the between-trials SD at each sample.

Statistical Analysis

We used separate, 1-way, repeated-measures analyses of variance (ANOVAs) to analyze the effect of the different running conditions on knee kinematics and kinetics during the stance phase (version 13.0; SPSS Inc, Chicago, IL). Running condition (NRUN, LFS20, or LFS35) was the within-subjects factor in each ANOVA. Dependent variables in the ANOVA were the peak knee angles and peak knee moments in each direction about each anatomical axis (flexion-extension, abduction-adduction, internal-external rotation) and the average variability in the knee angles and moments about each anatomical axis. The level of significance was set to $\alpha = .05$. When appropriate, post hoc comparisons among the 3 conditions were conducted using the Tukey honestly significant difference test with a level of significance of $\alpha = .05$.

RESULTS

Across all participants, mean running speeds were not different among the NRUN, LFS20, and LFS35 trials ($F_{2,24} = 3.30, P = .05$; Table 1). Thus, comparisons across the 3 step-width conditions were appropriate.

Knee Angles

The LFS maneuvers produced differences in peak knee flexion ($F_{2,24} = 15.92, P < .001$), extension ($F_{2,24} = 4.39, P = .02$), abduction ($F_{2,24} = 38.37, P < .001$), and internal rotation ($F_{2,24} = 23.85, P < .001$) during stance (Figure 2; Table 2). The Tukey post hoc procedure revealed that peak knee flexion was greater for LFS20 and LFS35 than for NRUN ($P < .05$; Table 2), whereas peak extension was less for LFS20 than for NRUN ($P < .05$; Table 2). Peak knee abduction was greater for LFS35 than for NRUN and LFS20 ($P < .05$; Table 2). Peak knee internal rotation was different among all 3 running conditions ($P < .05$), increasing progressively from NRUN to LFS20 to LFS35 (Table 2). Peak knee adduction and external rotation during stance did not differ across the 3 step-width conditions (adduction: $F_{2,24} = 0.43, P = 0.66$; external rotation: $F_{2,24} = 0.18, P = .84$; Table 2).

Average between-trials variability in knee angles across stance phase was greater for LFS maneuvers than for NRUN (Table 3). The Tukey post hoc test revealed that knee flexion-extension during stance was more variable for

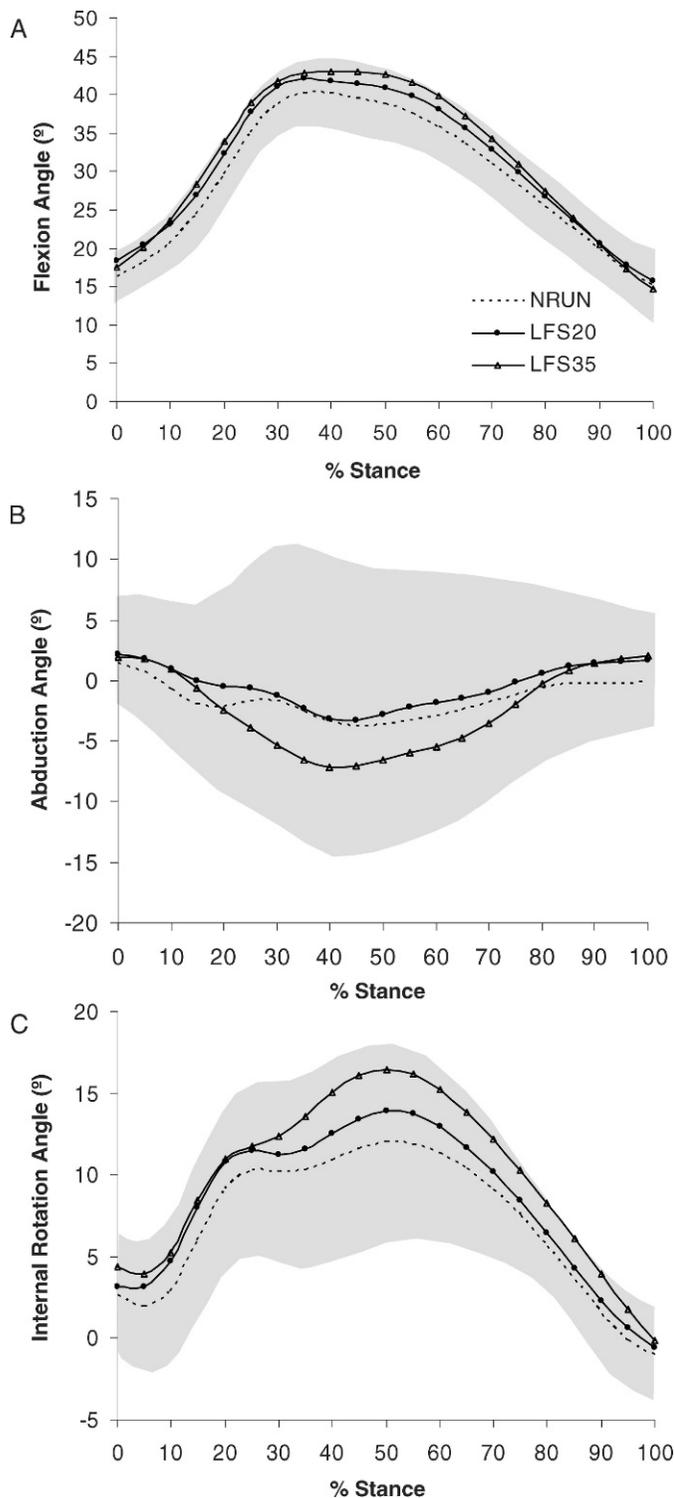


Figure 2. Ensemble average, across participants, of the mean joint angles at the knee for A, flexion-extension; B, abduction-adduction; and C, internal-external rotation during stance phase for normal running (NRUN) and for lateral false steps of 20% of body height (LFS20) and 35% of body height (LFS35). Shaded areas represent ± 1 SD for NRUN. Positive values indicate flexion, abduction, and internal rotation.

LFS35 than for LFS20 and NRUN ($P < .05$; Table 3), whereas knee abduction-adduction and internal-external rotation were more variable for both LFS20 and LFS35 than for NRUN ($P < .05$; Table 3).

Table 2. Peak Knee Angles ($^{\circ}$) as a Function of Lateral Step Width During Running (Mean \pm SD, N = 13)

Direction	Condition		
	Normal Running	Lateral False Step of 20% of Body Height	Lateral False Step of 35% of Body Height
Flexion	40.7 \pm 4.3	42.5 \pm 4.4 ^a	44.1 \pm 4.2 ^a
Extension	13.2 \pm 2.9 ^b	14.3 \pm 2.9 ^{a,b}	13.6 \pm 2.5 ^b
Abduction	0.7 \pm 3.7	1.7 \pm 3.4	3.2 \pm 4.0 ^{a,c}
Adduction	8.2 \pm 5.9	7.8 \pm 6.9	8.2 \pm 7.0
Internal rotation	13.1 \pm 5.4	15.2 \pm 5.9 ^a	17.6 \pm 6.8 ^{a,c}
External rotation	1.9 \pm 2.0	1.6 \pm 3.4	1.6 \pm 4.1

^a Tukey honestly significant difference test: $P < .05$ versus normal running.

^b Positive values correspond to flexion from the anatomical position.

^c Tukey honestly significant difference test: $P < .05$ versus lateral false step of 20% of body height.

Knee Moments

The LFS maneuvers produced differences in peak frontal-plane and transverse-plane knee abduction ($F_{2,24} = 12.39$, $P = .001$), adduction ($F_{2,24} = 33.04$, $P < .001$), and internal rotation ($F_{2,24} = 4.13$, $P = 0.03$) moments during stance (Figure 3; Table 4). Tukey post hoc testing revealed that peak knee abduction, adduction, and internal rotation moments were greater for LFS35 than for NRUN ($P < .05$; Table 4). The peak knee adduction moment was also greater for LFS35 than for LFS20 ($P < .05$; Table 4). Peak knee flexion, extension, and external rotation moments did not differ across the 3 conditions (flexion: $F_{2,24} = 0.26$, $P = .78$; extension: $F_{2,24} = 0.57$, $P = .58$; external rotation: $F_{2,24} = 3.43$, $P = .05$; Table 4).

Tukey post hoc analysis revealed that average between-trials variability in knee flexion-extension and abduction-adduction moments was greater for both LFS conditions than for NRUN ($P < .05$; Table 5). The average variability in internal-external rotation moments was greater only for LFS35 than for NRUN ($P < .05$; Table 5).

Table 3. Average Between-Trials Variability in Knee Angles ($^{\circ}$) Over Stance Phase as a Function of Lateral Step Width During Running (Mean \pm SD, N = 13)^a

Direction	Condition		
	Normal Running	Lateral False Step of 20% of Body Height	Lateral False Step of 35% of Body Height
Flexion-extension	1.8 \pm 0.5	2.1 \pm 0.7	2.5 \pm 0.7 ^{b,c}
Abduction-adduction	0.8 \pm 0.2	1.2 \pm 0.5 ^b	1.3 \pm 0.5 ^b
Internal-external rotation	1.3 \pm 0.3	1.9 \pm 1.1 ^b	2.0 \pm 0.8 ^b

^a Values represent the root mean square SD between trials.

^b Tukey honestly significant difference test: $P < .05$ versus normal running.

^c Tukey honestly significant difference test: $P < .05$ versus lateral false step of 20% of body height.

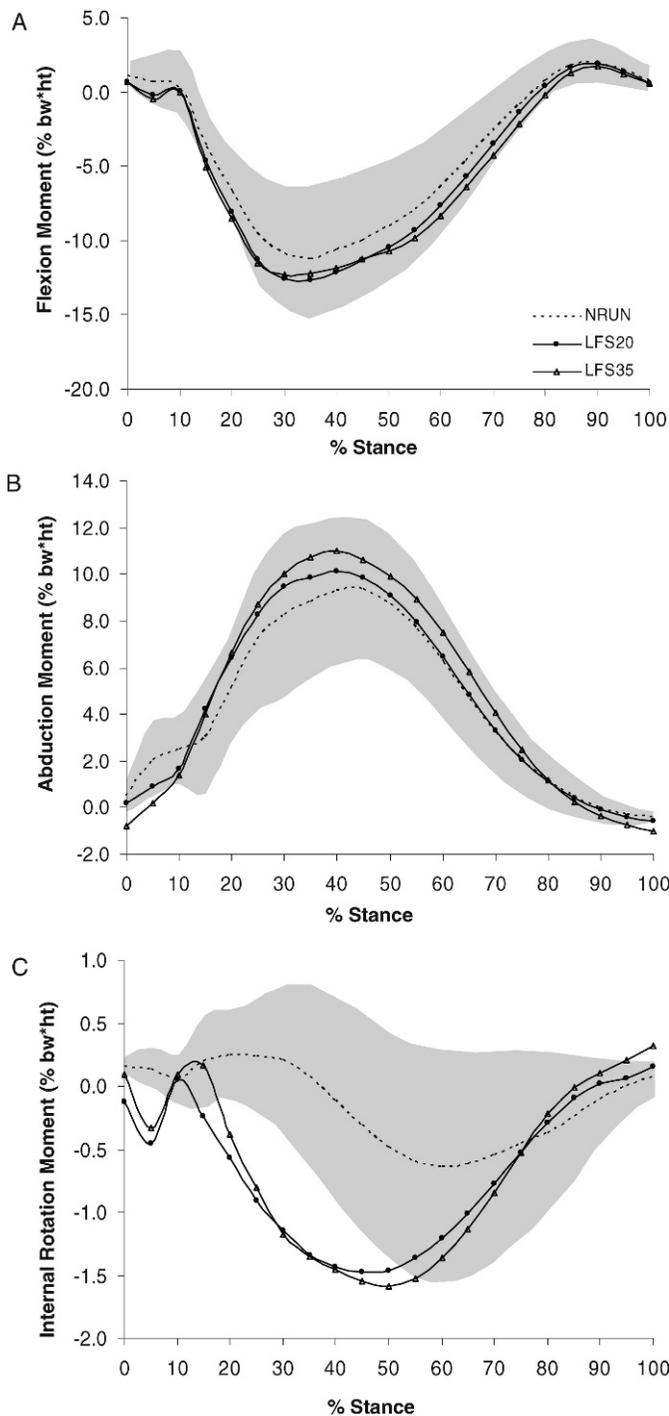


Figure 3. Ensemble average, across participants, of the mean internal moments acting at the knee for A, flexion-extension; B, abduction-adduction; and C, internal-external rotation during stance phase for normal running (NRUN) and lateral false steps of 20% of body height (LFS20) and 35% of body height (LFS35). Shaded areas represent ± 1 SD for NRUN. Moments are normalized to percentage of body weight \times body height (%bw*ht). Positive values indicate flexion, abduction, and internal rotation.

DISCUSSION

Our primary purpose was to quantify, in female basketball players, the peak knee angles and moments during the stance phase of normal running compared with those for LFS widths of 20% and 35% of BH. An LFS is defined as a widening of a single step within a running

Table 4. Peak Internal Knee Moments (Percentage of Body Weight \times Body Height) as a Function of Lateral Step Width During Running (Mean \pm SD, N = 13)

Direction	Condition		
	Normal Running	Lateral False Step of 20% of Body Height	Lateral False Step of 35% of Body Height
Flexion	3.37 \pm 1.55	3.23 \pm 1.13	3.53 \pm 1.55
Extension	12.23 \pm 2.37	12.73 \pm 2.60	12.50 \pm 3.11
Abduction	10.19 \pm 3.62	11.03 \pm 4.26	11.76 \pm 4.26 ^a
Adduction	0.53 \pm 0.35	1.02 \pm 0.83	2.01 \pm 0.83 ^{a,b}
Internal rotation	0.58 \pm 0.30	0.73 \pm 0.27	0.77 \pm 0.16 ^a
External rotation	1.02 \pm 0.76	1.08 \pm 0.89	1.37 \pm 1.08

^a Tukey honestly significant difference test: $P < .05$ versus normal running.

^b Tukey honestly significant difference test: $P < .05$ versus lateral false step of 20% of body height.

stride, followed by an immediate return to the original path of travel. The secondary purpose of this study was to investigate the average between-trials variability of these dependent measures across the 3 conditions.

Our aim in quantifying the kinematic and kinetic patterns associated with an LFS maneuver was to better understand the net loading of the knee and the potential risks for ACL disruption during this task. Redirection during running can be performed with a variety of movement behaviors. Although sidestep cutting and crossover cutting have been investigated as methods of redirection with regard to ACL injury mechanisms,^{12,18,19,28} other variations exist. Using an LFS to fake a change of direction in order to set up and avoid an opponent is arguably common in sports such as football, basketball, and soccer. The LFS may serve as an evasive maneuver in itself, as in the present study, or may be performed in combination with a sidestep-cutting maneuver. The latter is evident when athletes step laterally with the cutting limb just before executing a sidestep cut in the contralateral direction. Despite the common occurrence of the LFS maneuver, its kinematics and kinetics had not been studied to date.

Table 5. Average Between-Trials Variability in Knee Moments (% Body Weight \times Body Height) Over Stance Phase as a Function of Lateral Step Width During Running (Mean \pm SD; N = 13)^a

Direction	Condition		
	Normal Running	Lateral False Step of 20% of Body Height	Lateral False Step of 35% of Body Height
Flexion-extension	0.99 \pm 0.26	1.27 \pm 0.48 ^b	1.47 \pm 0.47 ^b
Abduction-adduction	0.71 \pm 0.17	0.89 \pm 0.26 ^b	1.04 \pm 0.47 ^b
Internal-external rotation	0.19 \pm 0.06	0.25 \pm 0.12	0.32 \pm 0.13 ^b

^a Values represent the root mean square SD between trials.

^b Tukey honestly significant difference test: $P < .05$ versus normal running.

In an LFS, the contralateral foot initiates the maneuver by projecting the athlete to the side at toe-off, leading to more lateral contact of the stepping foot. The stepping limb enters stance phase and must respond to the atypical placement of the foot, generating the control needed to reverse the mediolateral motion of the body and returning it to the original path of travel. Finally, upon the subsequent step, the contralateral limb redirects the motion of the body back along the original path of travel. We hypothesized that the greatest changes in knee kinematics and kinetics relative to normal running would occur during the stance phase of the laterally placed stepping limb; hence, this phase was of primary interest in this study.

Not surprisingly, although peak knee angles in the sagittal plane during stance were affected by an LFS, the general patterns of knee flexion-extension angles and moments were similar to those for normal running (Figures 2A and 3A). For all 3 conditions, the knee averaged 15° to 20° of flexion at initial contact, and it rapidly flexed thereafter until a median of 38% of stance was reached. Also for all conditions, the knee moments over the early portion of stance tended to be small or in flexion (or both), with the peak extension moment occurring, on average, at 33% of stance as the knee approached peak flexion. Neither the peak flexion nor peak extension moment differed among conditions. The LFSs were, however, associated with an average of 2.6° greater peak knee flexion than NRUN, independent of the step width. This increased flexion could reflect an attempt to reduce knee loading while arresting the body's greater lateral momentum.^{12,13,16} Peak knee extension was also an average of 1.1° less for LFS20 than for NRUN. Why peak extension was less for LFS20 but not for LFS35 is unclear. Furthermore, the fact that peak extension occurred at initial contact in 37% of trials but near toe-off (within the last 5% of stance) in the remaining trials makes the implications of this difference uncertain.

Effects of an LFS on frontal-plane knee angles and moments increased with the width of the step. Again, the general patterns of knee abduction-adduction angles and moments for an LFS were similar to those for normal running (Figures 2B and 3B). The knee typically was abducted at initial contact and reached its peak abduction shortly thereafter, at a median of 6% of stance, accompanied by an opposing adduction moment. Notably, the peak adduction moment occurred within the first 15% of stance in 29% of LFS20 and 65% of LFS35 trials, versus only 11% of NRUN trials. The peak adduction moment otherwise typically occurred within the last 15% of stance. Across all 3 conditions, the peak abduction moment occurred at an average of 39% of stance as the knee approached peak adduction, which was reached at an average of 43% of stance. Greater peak angles and moments relative to NRUN were observed for LFS35. The greater peak abduction angle and adduction moment for LFS35 (+2.5° and +1.5% BW × BH versus NRUN, respectively) may reflect, in part, an initial buckling of the stepping limb during weight support with the foot lateral to the body center of mass, particularly in the plurality of LFS35 trials (37%) in which both peaks occurred near initial contact. The greater peak abduction moment (+1.6% BW × BH versus NRUN) that followed was then associated with reversing the greater lateral velocity of the center of mass.

Lateral false steps of increasing width also had large and increasing effects on the knee in the transverse plane. For all 3 conditions, the knee was internally rotated at initial contact and continued to internally rotate until a median of 50% of stance was reached (Figure 2C). However, peak knee internal rotation was 2.1° greater for LFS20 than for NRUN and another 2.4° greater for LFS35, with increased internal rotation appearing to be present across most of stance during the LFS maneuvers. These changes in kinematics were accompanied by a change in the pattern of knee internal-external rotation moments (Figure 3C). The peak internal rotation moment occurred at a median of 16% of stance across all 3 conditions. However, this peak moment was 0.2% BW × BH greater for LFS35 than for NRUN and was consistently preceded by an external rotation moment only for an LFS. The subsequent transition from internal to external rotation moments also appeared to be more rapid for LFS maneuvers, with the peak external rotation moment occurring at a median of 48% of stance, versus 55% for normal running. The early external rotation moments for an LFS are likely in resistance to the moment created by the ground reaction forces used to arrest the body's lateral motion during forefoot contact. The increased internal rotation during midstance then reflects the rotation of the thigh about the leg as the body progresses past the laterally placed foot.

Although the clinical significance of these results is yet to be determined, they suggest that executing an LFS may place the ACL at risk of injury and that this risk may increase with increasing step width. The ACL serves as the primary restraint to tibial anterior translation and as a secondary restraint to knee internal rotation, abduction, and adduction.^{29,30} As a result, knee external rotation moments combined with either adduction or extension moments are associated with nontrivial ACL loads when the knee is flexed to less than 30°. ^{29,30} That an LFS was characterized by knee abduction, internal rotation, and 15° to 20° of flexion at and shortly after initial contact, accompanied by adduction and external rotation moments, thus suggests that the ACL was being loaded. Furthermore, with increasing step width, the peak knee abduction angles (found most often near initial contact) and peak internal rotation angles increased, and the peak knee adduction moments both increased and occurred more often near initial contact. These characteristics are consistent with proposed mechanisms of increased ACL loading, thus suggesting that greater LFS widths may be associated with greater risk of injury.

Another potential contributor to greater risk of ACL injury during LFS maneuvers was the greater between-trials variability of the knee angles and moments during stance, in comparison with normal running. This increased variability was observed in each of the 3 anatomical planes for both step widths, with 2 exceptions: between-trials variability in flexion-extension angles and internal-external rotation moments did not differ between LFS20 and NRUN. Arguably, greater between-trials variability in movement and loading patterns may increase the likelihood that the failure load of the ACL will be exceeded by chance during the execution of an LFS, as has been proposed for sidestep-cutting maneuvers.¹² Again, the clinical significance of the observed differences remains to be determined.

To our knowledge, this experiment is the first to investigate the effects of executing an LFS on knee kinematics and kinetics during running. Previous research related to noncontact ACL disruption during evasive maneuvers has focused primarily on knee kinematics^{10–13,16,18,19} and kinetics^{14,16,19} during sidestep-cutting maneuvers. An LFS differs fundamentally from sidestep cutting in that the latter is associated with an overall change of direction, whereas the former is not. Nevertheless, these 2 maneuvers have many similar effects on knee kinematics and kinetics. Consistent with the present LFS maneuvers, sidestep cutting has been associated with increased knee flexion and abduction, as well as greater adduction moments, during the initial portion of stance.^{12,13,16} Between-trials variability in knee abduction-adduction has also been reported to be greater for sidestep cutting than for normal running.¹² The effects of an LFS and sidestep cutting on the knee, however, appear to be different. Sidestep cutting has been associated with a more externally rotated (rather than more internally rotated) knee across stance¹² and decreased (rather than increased) abduction moments during midstance.¹⁶ Each of these differences is consistent with a greater risk of ACL injury during an LFS than during sidestep cutting. As such, a direct comparison of these maneuvers is warranted.

Our decision to increase the step width to 20% and 35% of BH during an LFS while running at 3.5 m·s⁻¹ was based upon preliminary investigation and determined to be realistic while providing a safe environment for the participants. By normalizing step width to BH, the intent was to obtain similar body configurations at initial contact across volunteers of different heights (although normalization to lower limb length might have been preferable, BH and lower limb length are strongly correlated [$r = 0.66$]³¹). Yet no data quantifying the actual step widths used by athletes in executing an LFS exist. Athletes may adopt an increase in step width dependent upon environmental constraints, that is, whether handling an implement or not, the intended purpose of the maneuver, the confines of the available playing area, and the speed at which the athlete is running. The effect of running speed on the execution of an LFS is a potential factor in our results in that the differences in running speed among conditions, although less than 3.5%, approached statistical significance ($P = .054$). The limb used in the LFS maneuver might also affect execution because of interlimb differences in coordination. We assumed that basketball athletes most often perform an LFS by the nondominant limb, typically in the same direction as the hand used to maintain ball control. In addition, the population of study was limited in sex and experience level to collegiate female basketball players. The effects of these different factors on the kinematics and kinetics of the knee during an LFS and on the associated risk of ACL injury are presently unknown.

The peak angles and moments presently reported are for all of stance phase, and the time of occurrence of these peaks often differed greatly among trials. Most notably, peak values of extension, abduction, and external rotation angles and of flexion, adduction, and internal rotation moments variously occurred both near initial contact and near toe-off for all 3 running conditions. Hence, differences in the peak values are suggestive of, but not necessarily indicative of, differences in ACL injury risk.

A final potential limitation is that 3 of the 13 players had a previous history of ACL reconstruction. These individuals underwent reconstruction 1.5 to 2.0 years before involvement in this study and had returned to full athletic participation without the use of external bracing. It is possible these individuals still had decrements in knee proprioception compared with preinjury conditions and uninjured athletes. However, all data points for these 3 players fell within 2 SDs of the mean for the group, and on only 2 occasions did any of these fall into the furthest quartile. Thus, although it is unusual to include volunteers with ACL reconstructions in a study such as this, we are confident that including these athletes had no significant effect on the results.

In conclusion, executing an LFS maneuver was associated with changes in the knee kinematics and kinetics of female basketball players during stance phase relative to normal, straight-ahead running. The magnitude of these changes typically increased with increasing step width. In general, LFS maneuvers were associated with greater peak flexion, abduction, and internal rotation angles; greater peak abduction, adduction, and internal rotation moments; and greater between-trials variability in angles and moments at the knee. Much of the research to date on evasive maneuvers has focused on sidestep cutting, but these results suggest that an LFS during running also has the potential to produce ACL strain and possible disruption. As such, further investigation of the LFS is warranted, both as an evasive maneuver in itself and in conjunction with a sidestep-cutting maneuver. Such study may lead to a clearer understanding of the mechanisms of noncontact knee injuries among athletes.

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