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Anticipatory Effects on Lower Extremity Neuromechanics During a Cutting Task

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Context: Continued research into the mechanism of noncontact anterior cruciate ligament injury helps to improve clinical interventions and injury-prevention strategies. A better understanding of the effects of anticipation on landing neuromechanics may benefit training interventions.

Objective: To determine the effects of anticipation on lower extremity neuromechanics during a single-legged land-and-cut task.

Design: Controlled laboratory study.

Setting: University biomechanics laboratory.

Participants: Eighteen female National Collegiate Athletic Association Division I collegiate soccer players (age = 19.7 ± 0.8 years, height = 167.3 ± 6.0 cm, mass = 66.1 ± 2.1 kg).

Intervention(s): Participants performed a single-legged land-and-cut task under anticipated and unanticipated conditions.

Main Outcome Measure(s): Three-dimensional initial contact angles, peak joint angles, and peak internal joint moments and peak vertical ground reaction forces and sagittal-plane energy absorption of the 3 lower extremity joints; muscle activation of selected hip- and knee-joint muscles.

Results: Unanticipated cuts resulted in less knee flexion at initial contact and greater ankle toe-in displacement. Unanticipated cuts were also characterized by greater internal hip-abductor and external-rotator moments and smaller internal knee-extensor and external-rotator moments. Muscle-activation profiles during unanticipated cuts were associated with greater activation of the gluteus maximus during the precontact and landing phases.

Conclusions: Performing a cutting task under unanticipated conditions changed lower extremity neuromechanics compared with anticipated conditions. Most of the observed changes in lower extremity neuromechanics indicated the adoption of a hip-focused strategy during the unanticipated condition.

Key Words: anticipation, anterior cruciate ligament, biomechanics

Key Points

- Participants demonstrated that the hip joint played a substantially greater role as part of the neuromechanical landing strategy during the unanticipated condition.
- The unanticipated condition was characterized by only a few changes in landing mechanics consistent with greater anterior cruciate ligament loading.

Noncontact anterior cruciate ligament (ACL) injuries are a common occurrence in sport.[1] When compared with their male counterparts, females are at a 3.5 to 4 times higher risk of sustaining an injury to the ACL.[2–4] Among ACL injuries in females, the most frequent mechanism of injury occurs in the absence of contact. On average, 70% to 80% of all noncontact ACL tears involve rapid deceleration during a landing or cutting maneuver.[2,5] To develop clinical interventions that aim to prevent injuries, research efforts have been directed at determining how biomechanical and neuromuscular risk factors manifest within the ACL injury mechanism.[6–16] Most authors[9,14,17,18] who investigated these risk factors and their role in the ACL injury mechanism focused on kinetic and kinematic variables measured during landing and cutting tasks. These results suggest that deleterious knee kinetics are characterized by greater external-flexion, abduction, and internal-rotation moments.[9,14,17,18] In addition, other kinematic factors, such as smaller hip and knee sagittal-plane angles along with greater frontal-plane angles and ranges of motion, have been identified as components of an at-risk movement pattern.[7,11,14,17,18] This movement pattern has also been evident during direct observations of ACL injuries using video analysis of in-game footage and as part of a prospective investigation of ACL injury risk.[6,19,20]

Many of the biomechanical studies[8,11,13,21,22] that investigated landing mechanics and movement patterns used experimental models in which participants completed preplanned movement tasks (eg, cutting or landing or both) with an anticipated or known direction of movement. However, executing movement tasks under such conditions may not accurately represent the dynamic and evolving environment in which athletic activities occur. Researchers, therefore, try to mimic the uncertainty of these conditions by having participants perform tasks without knowing which direction to move in before they actually initiate the movement.[7,9,12,14,15,17,18,23,24] It is interesting, however, that direct comparisons between lower extremity mechanics performed under anticipated and unanticipated conditions have not been conducted as often as comparisons between groups that performed only unanticipated conditions. Yet, analyzing biomechanical and neuromuscular variables under either condition alone may not provide sufficient information about the neuromechanical strategies that athletes adopt when faced with situations that more closely resemble the dynamic athletic environment. That may ultimately limit the insight available to develop appropriate training programs to prevent injury.

The few researchers[12,14,15] who directly compared anticipated and unanticipated conditions generally indicated that lower extremity mechanics were exacerbated when movement tasks were performed under unanticipated conditions. These results, however, primarily described differences in knee-joint biomechanics (ie, kinematics and kinetics) between movement tasks performed under anticipated and unanticipated conditions. Much less is known about the effects of anticipation on the biomechanics of more proximal joints (ie, the hip) or the underlying neuromuscular-control strategies that athletes adopt to cope with the demands presented by the unanticipated conditions.[8,25] Collecting electromyographic (EMG) data during dynamic movement tasks generally enhances the interpretation of traditional kinematic and kinetic data and, in the case of unanticipated movement conditions, would provide better global insight into the neuromechanical strategies that athletes adopt under suboptimal conditions for planning and executing task-appropriate movement patterns. Given the importance of hip-joint function in controlling upper body momentum and influencing lower body mechanics during landing,[10,21,26,27] the lack of knowledge about proximal biomechanics and muscle-activation patterns (eg, the gluteus maximus and medius) during unanticipated landing and cutting tasks could possibly limit insight into how neuromechanical strategies affect the ACL injury mechanism.

The purpose of our study was to analyze the effects of anticipation on lower extremity neuromechanics during a single-legged land-and-cut task. Specific emphasis was placed on muscle-activation patterns, in addition to kinematic and kinetic analyses, to obtain a more complete understanding of the neuromechanical strategies during landing in relation to the ACL injury mechanism. We hypothesized that landing under unanticipated conditions would be characterized by more deleterious joint kinematics, kinetics, and muscle-activation patterns in regard to the risk of noncontact ACL injury.

METHODS

Participants

Eighteen female collegiate (National Collegiate Athletic Association Division I) soccer players between 18 and 21 years of age were recruited during a preseason injury screening. The age, height, and mass of the participants were 19.7 ± 0.8 years, 167.3 ± 6.0 cm, and 66.1 ± 2.1 kg, respectively. All volunteers were healthy without musculoskeletal, neurologic, or cardiovascular conditions that would prevent them from fully participating in the testing protocol. A health history questionnaire was given to all athletes to determine whether injury occurred within the last year. If the athlete experienced a knee or ankle injury within the last year, she was excluded from the study. Before the study began, we explained all the testing procedures and obtained informed consent, as approved by the university's institutional review board.

Participant Preparation

The EMG electrodes were placed over the gluteus maximus, gluteus medius, biceps femoris, rectus femoris, vastus lateralis, and vastus medialis. Electrode placements were in accordance with SENIAM procedures (Roessingh Research and Development, the Netherlands) and were confirmed by manual muscle testing.[28] Using double-sided tape, we attached 23 single retroreflective markers for a static trial to the anatomical landmarks of the C7 spinous process, T10 spinous processes, and sternum and bilaterally over the posterior-superior iliac spine (PSIS), anterior-superior iliac spine, iliac crests, greater trochanters, medial and lateral knee-joint lines, medial and lateral malleoli, and first and fifth metatarsal heads. Individual markers that remained during testing were reinforced using tape to prevent movement or marker loss. Clusters of 4 markers attached to a plastic plate fixed to a hook-and-loop strap were used for the bilateral thighs and shanks.[29] These clusters were also reinforced using tape to prevent movement or slippage. A cluster of 3 markers attached to a plastic plate was fixed directly to the bilateral heels using standard athletic tape.[29] All noncluster markers were removed after the static trial except for the C7, T10, sternum, and bilateral PSIS and bilateral anterior-superior iliac spine markers.

Maximal Voluntary Isometric Contractions

Participants performed maximal voluntary isometric contractions (MVICs) for hip extension, hip abduction, knee flexion, and knee extension. For hip extension and hip abduction, participants completed the MVIC in a standing position with the test leg fully extended and hands placed on a support surface for stability. For knee flexion and extension, the participants were seated with the knees and hips at 90°. They were given verbal encouragement while producing a maximal force against the resistance of the investigator. Three 5-second trials were performed with a 1-minute rest between trials to minimize the effects of fatigue. The MVIC value was calculated as the average muscle activity over 100 milliseconds surrounding the peak EMG during three 5-second contractions and was identified using a custom MATLAB (MathWorks, Natick, MA) program.

Cutting Tasks

Each cutting trial began with the participant standing on a box that was normalized to her maximal vertical-jump height and placed at a distance from the force plate that was normalized to her maximum horizontal-jump distance.[29] The box height was normalized by having each participant first perform 5 maximal-effort vertical jumps. We used Vicon Nexus software (version 1.8.2; Vicon Motion Systems Ltd, Denver, CO) to generate a plot based on the position of the right PSIS marker during the third trial of the maximum vertical jump. The difference between the standing-height marker position and peak jump-height marker position was used to determine maximal vertical jump height. This difference was used to set the box height for the cutting trials. From this position, each participant performed a single-leg forward stride to land on her *dominant leg*, which was defined as the limb she would use to kick a soccer ball. A Tapeswitch signal mat (CVP, Farmingdale, NY) was placed on top of the box and interfaced with a custom LabView program (National Instruments Corporation, Austin, TX). The program displayed 3 lights to the participant on a monitor in front. When the participant's foot fully left the mat as she leaped forward, 1 of 3 visual signals was presented in randomized order on the video screen, cueing her to perform 1 of 3 movement tasks. The 3 tasks performed during the landing phase were continue running forward (RUN), stop and balance on the landing leg (STOP), and cut to the side away from the landing leg (CUT). The STOP activity served as a catch-trial condition to force the participant to exert more control over her center of gravity and to prevent her from assuming an adoptive hybrid strategy between the RUN and CUT tasks. For the final analysis, however, only the CUT task was evaluated. For the anticipated condition, the participants were given oral and visual cues to perform 5 RUN, 5 STOP, and 5 CUT trials, with approximately 30 seconds' rest between trials to minimize fatigue. For the unanticipated condition, the LabView program randomized each of the movement tasks. Participants performed the 3 tasks first under anticipated conditions, followed by unanticipated conditions. We chose this order to familiarize the participants with the 3 tasks and to ensure safe completion of the task under unanticipated conditions. Each participant performed the unanticipated condition until she completed either 20 total or 5 successful trials of each task. Successful trials required participants to land with the dominant foot completely on the force plate, without the nondominant leg contacting the force plate, and to correctly complete the designated task.

Data Acquisition

The EMG data were collected with a wireless Trigno system at 960 Hz (DelSys Inc, Boston, MA). Three-dimensional coordinate data from the reflective markers were collected with 14 cameras at 120 Hz (Vicon Motion Systems Ltd). Kinetic data were collected using 2 force plates at 960 Hz (AMTI Corporation, Watertown, MA). All data were synchronously collected with Vicon Nexus software (version 1.8.2; Vicon Motion Systems Ltd) and exported to Visual3D software (version 4; C Motion Inc, Rockville, MD) for analysis.

Data Processing

All EMG data were bandpass filtered at 10 to 500 Hz, rectified, and smoothed with a 4th-order low-pass Butterworth filter at 10 Hz. The EMG data collected during the dynamic tasks were then amplitude normalized to EMG data from MVICs to calculate percentage of activation during the anticipated and unanticipated landing tasks. A 4th-order, low-pass Butterworth filter with a cutoff frequency of 12 Hz was used to filter the exported kinematic and kinetic data.[29] A static standing trial was used to individualize marker position for each participant to allow accurate calculation of joint centers and segment positions during testing. A local coordinate system was used to define the model segments for the pelvis, thigh, shank, and foot segments based on marker position over the anatomical landmarks of the proximal and distal aspects of the segment. A joint-coordinate system determined the angles of the hip, knee, and ankle.[30,31] The *hip-joint center* was estimated to be at 25% of the horizontal distance between the greater trochanter markers.[29] The *knee-joint center* was defined as the point equidistant between the medial and lateral knee-joint markers, and the *ankle-joint center* was defined as the point equidistant between the medial and lateral malleoli markers. Inverse-dynamics procedures and anthropometric data were used to calculate joint kinetic data from the ground reaction force and kinematic data.[32] Joint moments are expressed in the distal segment relative to the proximal segment[33,34] and are reported as internal joint moments. All data were analyzed from *initial contact*, which was defined as the point at which the vertical ground reaction force was greater than 30 N, through the stance phase. Dependent variables extracted for analysis were joint angles at initial contact, joint displacement, peak joint moments, ground reaction forces, joint energies, and peak percentage of activation for precontact EMG and landing-phase EMG. *Range of motion* was defined as the angle at initial contact minus the angle at the end of the *landing phase*, which was defined as maximum knee flexion. For this study, *toe-in* and *toe-out* were defined as the positions between the shank and foot segments. The angle of the foot in reference to the shank in the frontal plane was used during the precontact phase and the shank in reference to the foot segment during landing was used to indicate toe-in and toe-out angles. Joint power was calculated as the dot product of the joint moment and joint angular velocity. Negative power phases in the sagittal plane were analyzed and energy absorption during landing was subsequently calculated through numerical integration. Precontact EMG activation included data collected 100 milliseconds before initial contact, with the timing choice based on previously conducted research.[8,35–37]

Statistical Analysis

Differences between dependent kinematic and kinetic variables were analyzed with general linear models. Specifically, several 2 $\times$ 3 (condition $\times$ joint) repeated-measures analyses of variance (ANOVAs) were used to assess the effects of anticipation within each plane on initial-contact angles, peak angles, peak torques, and energies. Similar 2 $\times$ 2 (condition $\times$ phase) repeated-measures ANOVAs were used to evaluate the effects of anticipation on activation of the 6 muscles of interest during the 2 conditions and the 2 movement phases (ie, precontact and postcontact). In all cases, if the overall ANOVA was significant, post hoc testing with paired *t* tests was performed. Ground reaction forces during the anticipated and unanticipated conditions were compared using paired *t* tests. All *t* tests were 2 tailed. Statistical significance was set a priori at $\alpha = .05$ and adjusted for multiple comparisons where necessary. All analyses were performed using SPSS statistical software (version 20.0; IBM Corp, Armonk, NY).

RESULTS

Significant interactions between condition and joint were observed for sagittal ($F_{2,32} = 11.0$, $P = .001$)- and frontal

Table 1. Angles (°) of the Lower Extremity Joints at Initial Contact During the Anticipated and Unanticipated Conditions

Dependent Variable	Condition		P Value
	Anticipated	Unanticipated	
Sagittal plane[a]			
Hip flexion	35.6 ± 12.0	36.7 ± 10.9	.144
Knee flexion[b]	19.6 ± 4.1	17.4 ± 4.1	.001
Ankle plantar flexion	10.1 ± 13.0	12.9 ± 12.6	.030
Frontal plane[a]			
Hip abduction	22.4 ± 5.2	23.3 ± 6.6	.207
Knee adduction	4.4 ± 2.4	4.8 ± 2.5	.152
Ankle inversion	8.3 ± 5.9	6.6 ± 6.6	.028
Transverse plane			
Hip internal rotation	0.8 ± 7.4	2.6 ± 8.6	.034
Knee external rotation	6.2 ± 6.1	7.5 ± 4.6	.069
Ankle toe-in rotation	7.9 ± 4.0	5.3 ± 3.8	.001

[a] Indicates significant interaction between joint and condition ($P < .05$).
[b] Indicates significant post hoc *t* test between conditions ($P < .017$).

Table 2. Ranges of Motion (°) of the Lower Extremity Joints During the Landing Phase of the Anticipated and Unanticipated Conditions

Dependent Variable	Condition		P Value
	Anticipated	Unanticipated	
Sagittal plane			
Hip flexion	25.6 ± 8.8	31.2 ± 6.1	.005
Knee flexion	47.9 ± 7.3	48.3 ± 7.1	.821
Ankle dorsiflexion	28.7 ± 15.8	31.9 ± 15.6	.029
Frontal plane			
Hip abduction	9.0 ± 4.2	11.6 ± 7.7	.082
Knee abduction	6.1 ± 3.1	5.7 ± 2.6	.307
Ankle inversion	7.2 ± 3.7	7.7 ± 4.1	.309
Transverse plane[a]			
Hip internal rotation	1.0 ± 1.8	2.1 ± 2.7	.079
Knee internal rotation	0.4 ± 1.1	0.1 ± 0.4	.238
Ankle toe-in rotation[b]	3.7 ± 3.5	7.6 ± 3.8	.001

[a] Indicates significant interaction between joint and condition ($P < .05$).
[b] Indicates significant post hoc *t* test between conditions ($P < .017$).

($F_{2,32} = 4.7$, $P = .016$)-plane initial-contact angles. Post hoc testing indicated that at initial contact, participants exhibited less knee flexion during the unanticipated condition (Table 1).

A significant interaction between condition and joint was evident for transverse-plane ($F_{2,32} = 14.6$, $P = .001$) range of motion. Post hoc testing showed that participants exhibited greater ankle toe-in rotation range of motion during the landing phase under the unanticipated versus the anticipated condition (Table 2).

Significant interactions between condition and joint were observed for sagittal ($F_{2,32} = 5.0$, $P = .02$)-, frontal ($F_{2,32} = 19.7$, $P = .001$)-, and transverse ($F_{2,32} = 36.6$, $P = .001$)-plane joint moments (Figure 1). Post hoc testing showed that in the sagittal plane, only knee-extensor moments were smaller during the unanticipated condition (Table 3). In the frontal plane, hip-abductor and ankle-invertor moments were both greater during the unanticipated condition. In the transverse plane, hip external-rotator moments were greater during the unanticipated condition, whereas knee external-rotator and ankle toe-out rotation moments were both smaller during the unanticipated condition.

A significant interaction occurred between condition and joint ($F_{2,32} = 13.4$, $P = .001$) for sagittal-plane energy absorption. Post hoc testing showed that the hip joint absorbed significantly more energy during the unanticipated versus the anticipated condition ($P = .001$), whereas the ankle joint absorbed less energy during the unanticipated condition than during the anticipated condition (Table 4).

Peak vertical ground reaction force (vGRF) differed between conditions. During the unanticipated condition, vGRFs were higher when compared with the anticipated condition ($P = .028$).

No significant interactions were noted between condition and phase for any of the muscles, but significant main effects were seen for the gluteus maximus ($F_{1,10} = 16.3$, $P = .002$) and vastus lateralis ($F_{1,8} = 5.3$, $P = .049$) muscles (Table 5). Post hoc testing demonstrated that the gluteus maximus muscles exhibited greater activity during both phases of the unanticipated condition compared with the anticipated condition (Figure 2).

DISCUSSION

The purpose of this study was to determine the effects of anticipation on lower extremity neuromechanics during a single-legged land-and-cut task. Our primary hypothesis was that the landing phase of the unanticipated condition would be characterized by joint kinematics, kinetics, and muscle-activation patterns that are purported to increase the risk of noncontact ACL injury. Contrary to our hypothesis, however, most of the observed changes in lower extremity neuromechanics during the unanticipated condition were consistent with landing strategies thought to decrease the risk of injury. A major feature of these positive changes was a shift toward a hip-dominant strategy, which may have important clinical implications for the design of injury-prevention protocols.

Our most noteworthy findings relate to the substantially greater role of the hip joint in the neuromechanical landing strategy during the unanticipated condition. When compared with the anticipated condition, participants exhibited greater sagittal-plane hip-energy absorption, internal hip external-rotator and hip-abductor moments, and gluteus maximus muscle activation during the landing phase of the cutting task under unanticipated conditions. These changes may have important clinical implications with respect to the mechanisms typically purported to cause ACL injury. For example, a lack of change in transverse-plane hip kinematics may indicate that the greater hip external-rotation moment could help protect the femur against internal-rotation motion, which has been suggested to be a deleterious posture that moves the lower extremity into a position of "no return" with regard to ACL injury.[38] Similarly, given a lack of change in frontal-plane hip kinematics, greater abduction moments at the hip could serve to limit femoral adduction, which relates to the potentially injurious dynamic valgus posture.[6,39] Brown et al[17] drew similar conclusions after they reported greater hip abduction during unanticipated cutting, which they hypothesized was a plausible protective mechanism to decrease potentially hazardous knee-abduction motion. The greater energy absorption at the hip joint, greater gluteus maximus muscle activation, and smaller knee-extensor moment

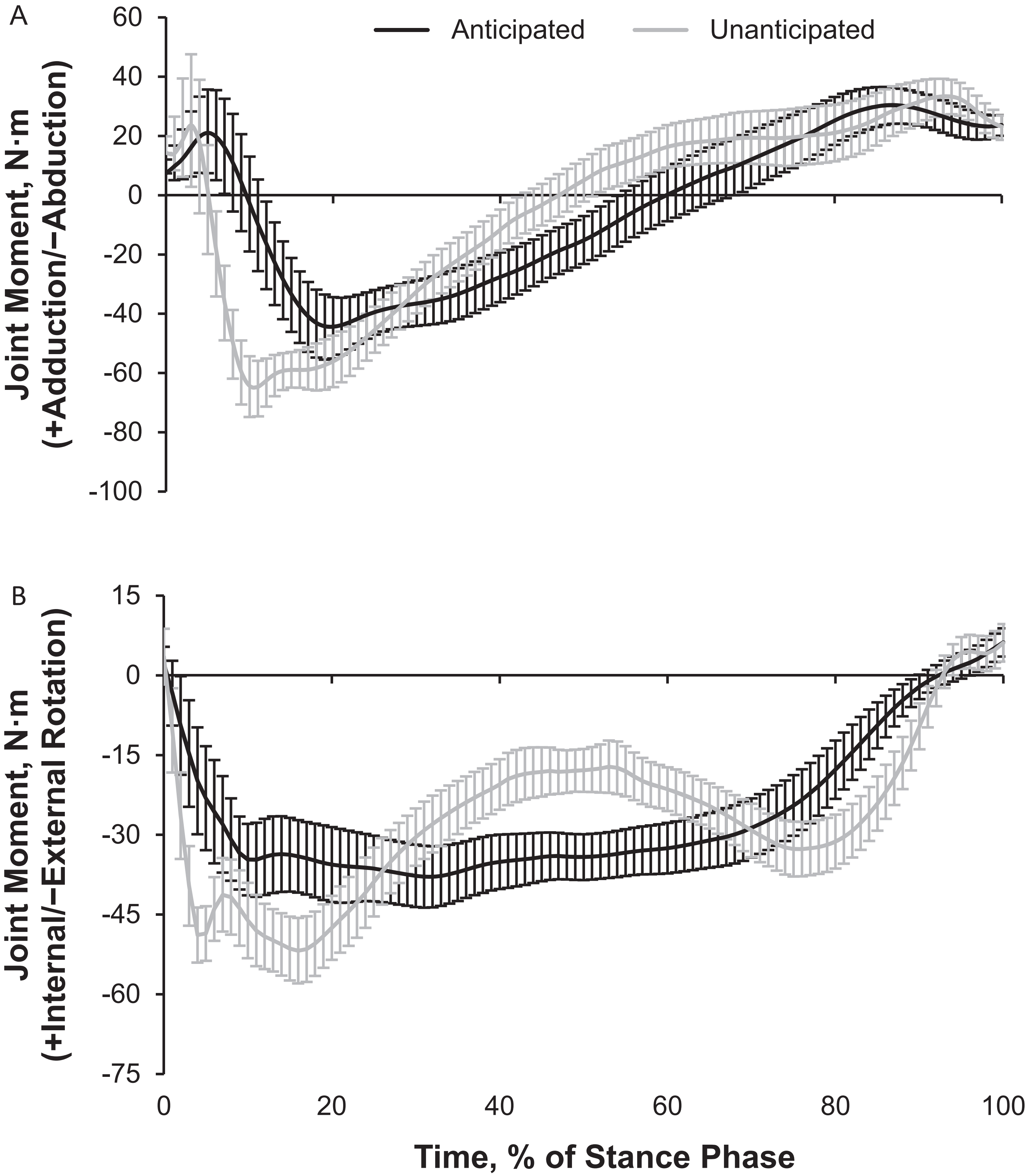

Figure 1. Hip kinetics (N·m) during the stance phase of the anticipated (black line) and unanticipated (gray line) single-legged cutting task. A, Hip frontal-plane moment. B, Hip transverse-plane moment.

during landing further support the proposition that the participants in the current study adopted a hip-dominant strategy during the unanticipated condition. Although most of the changes at the hip occurred during the landing phase, gluteus maximus activation was also greater during the precontact phase of the unanticipated condition. Several groups[8,13,18] have suggested that successful movements during landing and cutting tasks largely result from preplanned neuromechanical control strategies. Therefore, improved muscle activation under unanticipated conditions

Table 3. Peak (Internal) Moments (N·m) of the Lower Extremity Joints During the Landing Phase of the Anticipated and Unanticipated Conditions

Dependent Variable	Condition: Anticipated	Condition: Unanticipated	*P* Value
Sagittal plane[a]			
Hip extensor	168.0 ± 47.0	190.6 ± 66.1	.041
Knee extensor[b]	163.0 ± 25.0	155.0 ± 22.8	.008
Ankle plantar flexor	124.7 ± 19.1	119.6 ± 19.7	.278
Frontal plane[a]			
Hip abductor[b]	57.5 ± 17.7	72.1 ± 18.5	.001
Knee adductor	19.8 ± 8.1	21.7 ± 11.7	.520
Ankle invertor[b]	12.7 ± 6.0	19.0 ± 7.9	.002
Transverse plane[a]			
Hip external rotator[b]	49.0 ± 12.4	59.9 ± 11.7	.001
Knee external rotator[b]	9.8 ± 4.6	7.0 ± 3.9	.015
Ankle toe-out[b]	16.3 ± 5.5	12.9 ± 5.7	.013

[a] Indicates significant interaction between joint and condition ($P < .05$).
[b] Indicates significant post hoc *t* test between conditions ($P < .017$).

may indicate a shift toward an appropriate, centrally preprogrammed response that participants adopted to deal with the new task demands. Appropriate precontact muscle activations are also deemed more critical than muscle activations during landing because a muscle's response latency is generally considered too long to reactively stabilize a joint during landing.[40,41]

We originally hypothesized that executing movement tasks under unanticipated conditions would deleteriously alter lower extremity neuromechanics. This proposal was based on observations by other authors[9,14,15] of greater knee-joint and ACL loading during unanticipated conditions. For example, Besier et al[9] found that cutting tasks performed under unanticipated conditions resulted in greater deleterious knee varus-valgus and internal-external–rotation moments. Furthermore, Weinhandl et al[15] reported that unanticipated sidestep cutting was associated with greater modeled ACL strain compared with anticipated sidestep cutting. Again, in contrast to these findings, we noted that one aspect of knee-joint loading (ie, knee-extensor moment) was smaller during the landing phase of the unanticipated conditions. Although some of our results contradicted our original hypotheses and some of these earlier outcomes, not all of our results did. For example, we also found that participants landed with greater vGRFs and smaller knee-flexion angles at initial contact during the unanticipated condition. Participants in the current study

Table 4. Net Mechanical Energy (W·kg^{-1}) of the Lower Extremity Joints During the Landing Phase of the Anticipated and Unanticipated Conditions

Dependent Variable	Condition: Anticipated	Condition: Unanticipated	*P* Value
Sagittal plane[a]			
Hip[b]	−0.84 ± 0.39	−1.06 ± 0.49	.001
Knee	−1.47 ± 0.42	−1.26 ± 0.33	.017
Ankle[b]	−0.77 ± 0.31	−0.62 ± 0.27	.001

[a] Indicates significant interaction between joint and condition ($P < .05$).
[b] Indicates significant post hoc *t* test between conditions ($P < .017$).

Table 5. Peak Muscle Activity (%EMG$_{MVIC}$) During the Precontact Phase (100 Milliseconds Before Initial Contact) and Landing Phase (First 20% After Initial Contact) of the Anticipated and Unanticipated Conditions

Dependent Variable	Condition: Anticipated	Condition: Unanticipated	*P* value
Gluteus maximus[a]			
Precontact[b]	0.34 ± 0.19	0.49 ± 0.27	.005
Landing[b]	1.0 ± 0.64	1.32 ± 0.60	.016
Gluteus medius			
Precontact	0.22 ± 0.11	0.32 ± 0.35	.033
Landing	0.68 ± 0.28	1.13 ± 0.65	.043
Biceps femoris			
Precontact	0.24 ± 0.15	0.24 ± 0.16	.967
Landing	1.10 ± 0.69	1.48 ± 1.11	.313
Rectus femoris			
Precontact	0.47 ± 0.39	0.46 ± 0.43	.759
Landing	0.87 ± 0.64	0.74 ± 0.65	.437
Vastus lateralis[a]			
Precontact	0.28 ± 0.14	0.33 ± 0.20	.983
Landing	1.48 ± 0.96	2.10 ± 0.85	.578
Vastus medialis			
Precontact	0.19 ± 0.15	0.16 ± 0.10	.978
Landing	0.98 ± 0.67	1.16 ± 0.73	.055

Abbreviations: EMG, electromyography; MVIC, maximal voluntary isometric contraction.
[a] Indicates main effect for condition ($P < .05$).
[b] Indicates significant post hoc *t* test between conditions ($P < .017$).

landed with smaller knee-flexion angles at initial contact, but their overall knee-flexion range of motion during the landing phase did not differ between the conditions, which would indicate that the overall knee-flexion angles during the landing phase of the unanticipated condition were smaller. In vivo studies[42,43] have demonstrated that ACL strain is greater when knee-flexion angles are limited. Furthermore, smaller knee-flexion angles and greater ground reaction forces have also been linked to ACL injuries in a prospective study.[6] A possible explanation as to why only a few of our findings showed changes in landing mechanics consistent with greater ACL loading may be the training level of the athletes we tested. Our population was composed entirely of high-level, National Collegiate Athletic Association Division I soccer players, all of whom participated in rigorous training throughout the year. In comparison, the authors of most other studies investigated the effects of decision making during unanticipated landing or cutting study models in recreationally trained participants. Indeed, a recent group[44] reported that Division I players were better able to complete tasks under unanticipated conditions than recreationally active individuals. It could be surmised that the greater experience and exposure of the Division I athletes better prepared them for more successful completion of the movement tasks under unanticipated conditions. Because the current findings suggest that high-level athletes might use more of a hip strategy to adapt to the unanticipated condition, untrained individuals in previous studies might not have been able to rely on an adaptation in hip-joint neuromechanics and therefore demonstrated increased knee-joint loading. Our results thus suggest that hip neuromechanics play an

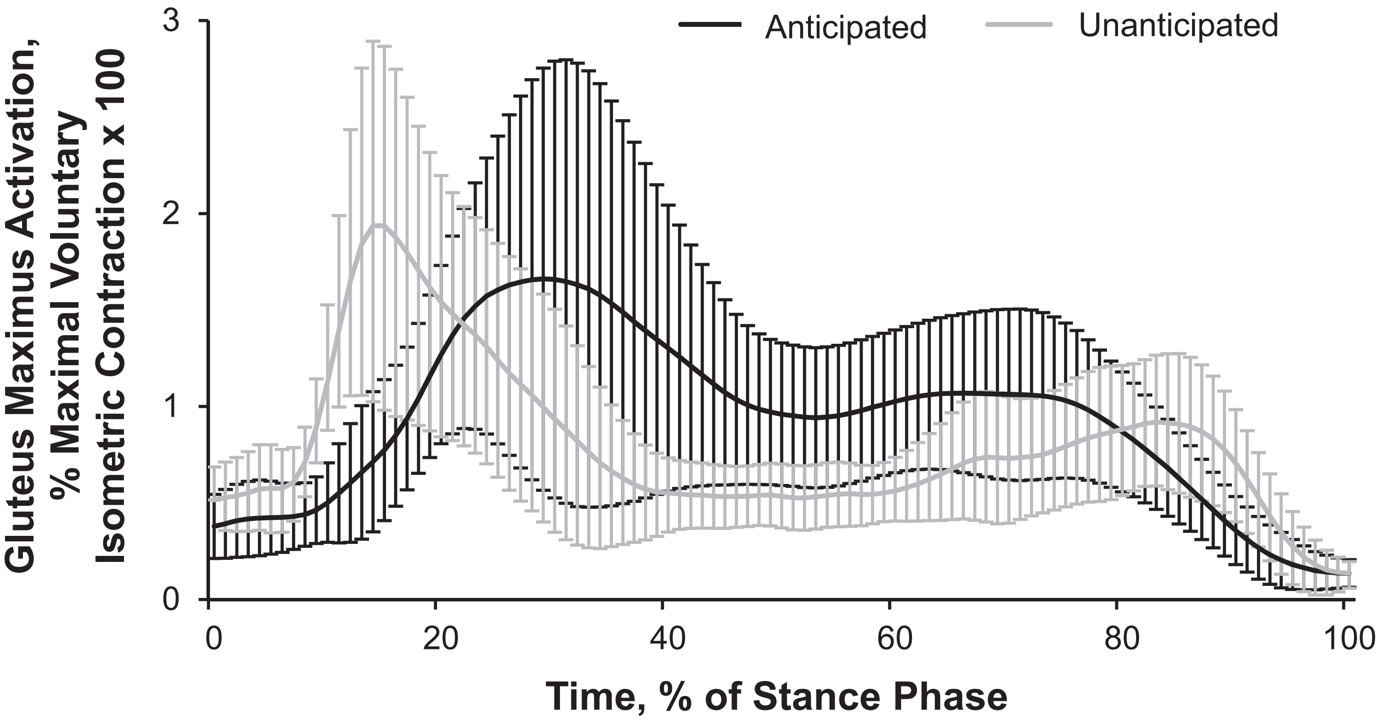

Figure 2. Gluteus maximus electromyograph (% maximal voluntary isometric contractions × 100) during the stance phase of the anticipated (black line) and unanticipated (gray line) single-legged cutting task.

important role in controlling or maintaining (or both) knee mechanics under unanticipated conditions. Another explanation for the discrepancy in outcomes could arise from the tasks that participants performed in these studies. For example, based on the nature of the task in our study, participants might have completed the task under the unanticipated condition with more caution and subsequently more proximal control to counteract the uncertainty associated with the condition. This may be reflected through increased neuromuscular preparation, as demonstrated by greater feed-forward muscular activation before initial contact[45] and the concomitant improvements in hip mechanics.

Given the aforementioned discrepancies among studies that have addressed the effects of anticipation on neuromechanical changes of the lower extremity,[9,14,17,18,23,24] it may also be prudent to consider general limitations to unanticipated models that are commonly used in the literature. Arguably, participants may simply assume less dangerous landing mechanics when completing tasks under unanticipated conditions compared with anticipated conditions. For example, Besier et al[9] showed that dynamic movements under unanticipated conditions are subject to different neuromotor-control strategies that favor joint stability. This raises questions about the efficacy and use of current unanticipated movement paradigms when researching noncontact ACL injury mechanisms and may indicate that adding further perturbations during unanticipated tasks would better simulate gamelike scenarios.[46] Future authors should thus focus on finding the appropriate stimulus, task, or paradigm to best recreate gamelike scenarios during landing.

We acknowledge several additional limitations of the present study. First, the timing with which each athlete received the visual stimulus is expected to vary slightly based on the experimental setup. Because box height was normalized to maximal vertical-jump height, participants with greater vertical-jump heights may have been allotted more time to receive and interpret the visual signal to successfully complete the task. However, varying the time with which visual stimuli are presented appears to have little effect on landing mechanics.[17] Second, the experimental setup was such that all participants completed the anticipated tasks before the unanticipated tasks. Although this order was used to ensure safety and familiarity on behalf of the participants, it may have inadvertently introduced an ordering or fatigue effect (or both). Whereas fatigue could also become a concern with an excessive number of trials, the total number of cuts in the current study (approximately 30) is well within recommended guidelines for plyometric training sessions, which are designed to minimize fatigue and ensure optimal exercise technique and performance, even for novice athletes.

Given the hip joint's role in controlling upper body momentum and influencing lower body mechanics during landing,[10,21,26,27] it may be sensible to consider this joint a potential target for interventions that aim to improve landing mechanics. Indeed, Stearns and Powers[47] demonstrated that a hip-focused strength-training program can improve landing mechanics. In addition to observing positive biomechanical changes at the hip, the same authors saw lower peak knee-abduction angles and average knee-adductor moments—both risk factors for ACL injury.[47] Although the extent to which hip strengthening would affect landing mechanics during unanticipated conditions or gamelike scenarios is not known, the current study's findings highlight the importance of a hip-dominant strategy when the time for planning and executing task-appropriate movement patterns is suboptimal. Therefore, strengthening programs may prove a viable part of ACL injury-prevention efforts in clinical settings and should be the focus of future research efforts.

CONCLUSIONS

When compared with anticipated conditions, the performance of single-legged land-and-cut maneuvers under unanticipated conditions led to significant neuromechanical changes that were characterized by a shift toward a hip-dominant strategy. We find it interesting that the combination of the observed changes in hip neuromechanics may reflect a less deleterious movement strategy, which may have clinical implications for injury-prevention efforts. Analysis of the proximal joints may help us to better understand neuromechanical landing strategies and their role within the complex nature of noncontact ACL injury mechanisms, especially under unanticipated movement conditions.

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