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original research

Changing Sagittal-Plane Landing Styles to Modulate Impact and Tibiofemoral Force Magnitude and Directions Relative to the Tibia

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Context: Ground reaction force (GRF) and tibiofemoral force magnitudes and directions have been shown to affect anterior cruciate ligament loading during landing. However, the kinematic and kinetic factors modifying these 2 forces during landing are unknown.

Objective: To clarify the intersegmental kinematic and kinetic links underlying the alteration of the GRF and tibiofemoral force vectors secondary to changes in the sagittal-plane body position during single-legged landing.

Design: Crossover study.

Setting: Laboratory.

Patients or Other Participants: Twenty recreationally active participants (age = 23.4 ± 3.6 years, height = 171.0 ± 9.4 cm, mass = 73.3 ± 12.7 kg).

Intervention(s): Participants performed single-legged landings using 3 landing styles: self-selected landing (SSL), body leaning forward and landing on the toes (LFL), and body upright with flat-footed landing (URL). Three-dimensional kinetics and kinematics were recorded.

Main Outcome Measure(s): Sagittal-plane tibial inclination and knee-flexion angles, GRF magnitude and inclination angles relative to the tibia, and proximal tibial forces at peak tibial axial forces.

Results: The URL resulted in less time to peak tibial axial forces, smaller knee-flexion angles, and greater magnitude and a more anteriorly inclined GRF vector relative to the tibia than did the SSL. These changes led to the greatest peak tibial axial and anterior shear forces in the URL among the 3 landing styles. Conversely, the LFL resulted in longer time to peak tibial axial forces, greater knee-flexion angles, and reduced magnitude and a more posteriorly inclined GRF vector relative to the tibia than the SSL. These changes in LFL resulted in the lowest peak tibial axial and largest posterior shear forces among the 3 landing styles.

Conclusions: Sagittal-plane intersegmental kinematic and kinetic links strongly affected the magnitude and direction of GRF and tibiofemoral forces during the impact phase of single-legged landing. Therefore, improving sagittal-plane landing mechanics is important in reducing harmful magnitudes and directions of impact forces on the anterior cruciate ligament.

Key Words: anterior cruciate ligament, tibial posterior slope, lower extremity biomechanics, injury prevention, landing strategy

Key Points

- At the impact phase of single-legged landing, sagittal-plane kinetics and kinematics of body segments were strongly related, affecting both the magnitude and direction of the ground reaction force (GRF) and tibiofemoral forces relative to the tibia and anterior cruciate ligament injury risk.
- A longer time to peak GRF in single-legged landing on the toes with the body leaning forward allowed greater knee flexion, anterior tibial inclination, and shock attenuation, leading to a reduced magnitude of GRF that was more posteriorly inclined relative to the tibia and smaller tibial axial forces and posteriorly directed tibial shear forces.
- A shorter time to peak GRF during flat-footed landing with the body upright led to less knee flexion, anterior tibial inclination, and shock attenuation, resulting in a greater magnitude of GRF that was more anteriorly inclined relative to the tibia and greater tibial axial and anterior shear forces.

Researchers[1] have estimated that more than 200 000 anterior cruciate ligament (ACL) injuries occur annually during sport activities in the United States alone. Most of these injuries are noncontact and are sustained during sharp decelerating motions, such as landing or changing direction while running.[2–4] Anterior cruciate ligament injuries are devastating for athletes because of the long periods missed from sport participation[5]; high medical costs[6]; and increased risk of chronic degenerative conditions, such as osteoarthritis.[7] Therefore, clarifying biomechanical strategies to protect the ACL during sharp decelerating motions is necessary, particularly for female athletes, who have much higher rates of ACL injury than male athletes.[4,8]

Proximal tibial anterior shear forces are thought to be the primary cause of ACL injury because they directly load the ACL, especially at shallow knee-flexion angles.[3,9] Investigators[10–12] have shown that the ground reaction forces

(GRFs) that athletes incur right after foot contact in sharp decelerating motions greatly affect the magnitude of proximal tibial anterior shear forces and ACL loading. Authors[13,14] of in vivo studies who measured the amount of ACL strain during landing tasks demonstrated that the timing of peak ACL strain coincides with the peak GRF that occurs immediately after the foot contact of landing. Taylor et al[13] also demonstrated that knee-flexion angles were inversely related to ACL strain during landing. Furthermore, the direction of peak GRF relative to the tibia in the sagittal plane has been reported to substantially affect the amount of net proximal tibial anterior shear forces immediately after the foot contact of landing.[11,12] In their video analysis, Krosshaug et al[15] estimated that noncontact ACL injuries occurred approximately 40 milliseconds after foot contact during landing or cutting, which closely corresponds to the time to peak GRF during single-legged landing.[16] The results of these studies collectively have indicated that sagittal-plane mechanics, including GRF direction and magnitude and sagittal-plane knee kinetics and kinematics, may greatly affect the direction and magnitude of peak tibiofemoral forces, influencing ACL loading immediately after foot contact during sharp decelerating motions.

Whereas noncontact ACL injuries are thought to occur frequently when GRFs increase sharply, researchers[17] have presumed that tibial axial forces play an important role in ACL rupture. This presumption was based on cadaveric studies[18,19] in which investigators demonstrated that proximal tibial axial forces of sufficient magnitude were transformed into tibial anterior shear forces, especially with posterior slope on the tibial plateau, and eventually rupture the ACL even if no muscle forces are present. Given that GRFs are the largest external forces on the body, particularly immediately after foot contact, researchers[18] have implied that incurring a large GRF that is parallel to the longitudinal axis of the tibia increases the risk of ACL injuries. However, during normal double-legged landing, authors[11,20] of simulation studies reported that GRFs were directed posteriorly to the tibia, and these deceleration forces pushed the tibia posteriorly and reduced the ACL strain. Overall, ACL injury is most likely to occur when one incurs excessive GRFs that are directed parallel or more anteriorly inclined relative to the tibia immediately after the foot contact of landing, whereas small GRFs that are posteriorly inclined relative to the tibia should protect against ACL injury.

To our knowledge, no one has investigated what conditions modulate the magnitude and direction of GRFs and tibiofemoral forces relative to the tibia in the sagittal plane during landing. Understanding these biomechanical links will (1) provide new insights into how sagittal-plane kinetics and kinematics affect noncontact ACL injury mechanics, (2) offer clinicians a better basis for teaching athletes about safe movement patterns for protecting the ACL, and (3) enable practitioners to design training and rehabilitation programs that promote safe movement patterns during sharp decelerating motions. Therefore, the purpose of our study was to examine 2 hypothetical sagittal-plane kinematic and kinetic intersegmental links that would modulate both the magnitude and direction of GRFs and tibiofemoral forces during single-legged landings. Shimokochi et al[16] demonstrated that landing flat-footed in an upright position results in reduced shock-attenuation capacity and increased peak GRF, as well as decreased time to peak GRF during single-legged landing. Contrarily, landing with the body leaning forward and landing on the toes increased shock-attenuation capacity, decreased peak GRF, and increased time to peak GRF. Therefore, we tested 2 hypotheses: (1) We hypothesized that landing flat-footed in an upright position would lead to a greater and earlier occurrence of peak tibial axial forces after foot contact during landing due to the greater and earlier occurrence of peak GRF. This would be associated with smaller knee-flexion and tibial anterior-inclination angles relative to the world vertical axis at peak tibial axial forces owing to less time for knee flexion after foot contact. Given this small tibial anterior-inclination angle from vertical, the GRF vector would be aligned parallel or inclined more anteriorly relative to the tibia in the sagittal plane. Therefore, the tibia should receive little or no posterior shear force (or greater anterior shear force). (2) In contrast, we hypothesized that landing on the toes with the body leaning forward would result in a smaller and delayed occurrence of peak tibial axial forces after foot contact during landing owing to a small and delayed occurrence of peak GRF. This would be associated with greater knee-flexion and tibial anterior-inclination angles relative to the world vertical axis at peak tibial axial forces owing to more time for knee flexion after foot contact. Secondary to this greater tibial anterior inclination from vertical, the GRF vector would be inclined more posteriorly relative to the tibia in the sagittal plane. Therefore, the tibia should receive a posterior shear force (or little anterior shear force).

METHODS

Participants

We analyzed kinematic and kinetic data from 20 recreationally active adults, including 10 men (age = 25.4 ± 3.8 years, height = 178.7 ± 5.9 cm, mass = 81.4 ± 7.2 kg) and 10 women (age = 21.4 ± 1.8 years, height = 163.2 ± 4.2 cm, mass = 62.0 ± 8.7 kg), who participated in an earlier study.[16] We defined *recreationally active* as being involved in physical activities for at least 30 minutes per day, 3 times per week. All participants provided written informed consent, and the study was approved by the Institutional Review Board for the Protection of Human Participants at the University of North Carolina at Greensboro. Volunteers with any history of knee ligament injury or lower extremity pain at the time of participation were excluded.

Data Collection

Participants performed all single-legged landings wearing their own running shoes. All kinematic and kinetic data were collected using a nonconductive force plate (type 4060; Bertec Corporation, Columbus, OH) and a 3-dimensional electromagnetic tracking system with Ascension Star hardware (Ascension Technology, Burlington, VT) and MotionMonitor software (Innovative Sports Training, Chicago, IL). The participants performed 3 different landing styles using the *dominant lower extremity*, which was defined as the limb preferred for the single-legged landing, from a 30-cm box (women) or a 45-cm box (men). The men landed from a greater box height because their jump capacities are often greater than those of women[21–24] and because we confirmed that recreationally active men and women could successfully perform the 3 landing styles from these box heights. After digitization procedures were performed as described previously,[16] the participants were familiarized with single-legged drop landings until they were comfortable with all styles. Next, they performed 5 single-legged landing trials onto a force plate using each of the 3 different landing styles: self-selected (SSL), body leaning forward with a more plantar-flexed position at foot contact (LFL), and body upright with a flat-footed position (URL).[16] The examiner (Y.S.) provided the same instructions to all participants as described in a previous study.[16] A trial was discarded and repeated for the reasons detailed in the earlier study.[16] The SSL was always performed first, followed by the LFL or URL in a counterbalanced order. Only data from successful trials were collected; those from unsuccessful trials in which participants could not land on the force plate and remain standing for 2 seconds after landing were discarded.

Data Processing and Reduction

Kinematic and kinetic data were collected at 1000 and 120 Hz, respectively, and kinematic data were interpolated linearly to align with kinetic data by resampling at 1000 Hz. Kinematic data were low-pass filtered using a second-order Butterworth filter at 8 Hz.[25] A laboratory coordinate system with the y, x, and z axes as vertical, anteroposterior, and mediolateral axes, respectively, during landing was embedded on the center of the force plate. Segment inertial and anthropometric properties were based on the data of Dempster.[26] *Knee-joint center* and *ankle-joint center* were defined as the midpoints of the lateral and medial femoral epicondyles and the lateral and medial malleoli, respectively. *Hip-joint center* was defined as described by Leardini et al.[27] The longitudinal axis for the local coordinate systems of the foot, shank, and thigh was aligned to the line between the tip of the second phalanx and the ankle-joint centers, between the ankle- and knee-joint centers, and between the knee- and hip-joint centers, respectively. Anteroposterior axes for each segment were aligned perpendicularly to the longitudinal axis of each segment and parallel to the x axis in the laboratory coordinate system when participants stood in an anatomically neutral position with their feet aligned parallel and shoulder-width apart and faced parallel to the x axis. Mediolateral axes for each segment were aligned perpendicularly to both the longitudinal and anteroposterior axes in the segmental local coordinate system. Relative joint angles were calculated using a Euler-Cardan angle method with the rotation order of flexion-extension, internal-external rotation, and abduction-adduction. Lower extremity joint-reaction forces and internal moments were calculated using a Newtonian inverse-dynamics approach described by Winter.[28] Proximal tibial anterior-posterior and superior-inferior joint-reaction forces and sagittal-plane knee moment were expressed using the local coordinate system embedded at the shank center of mass.

Proximal tibial anterior-posterior and *superior-inferior forces* were calculated using the inverse-dynamic approach and defined as the forces when magnitudes were equal but opposite in direction relative to the proximal tibial joint-reaction forces. Therefore, anteriorly and superiorly directed proximal tibial forces push the distal femur anteriorly and superiorly (and are, thus, assumed to strain the ACL), respectively.

To test our hypotheses, we extracted several targeted sagittal-plane kinematic and kinetic variables at the time of peak tibial axial forces after the foot contact of each single-legged landing: peak proximal tibial axial force, proximal tibial shear forces, vertical GRF (GRF_V), anteroposterior GRF (GRF_{AP}), sagittal-plane internal knee moment, GRF angles relative to vertical, tibial inclination angles relative to vertical, GRF angles relative to the tibia, and knee-flexion angles. Peak proximal tibial axial forces, proximal tibial shear forces, GRF_V, and GRF_{AP} were normalized by body weight (BW) in newtons. Sagittal-plane knee moments were normalized by the product of BW and height in meters. Thus, these kinetic variables have no unit.

Furthermore, to examine whether participants successfully modified their landing styles, center-of-gravity (COG) anterior-posterior positions relative to the ankle-joint center were calculated at the foot contact and peak tibial axial forces. Superiorly directed tibial axial forces, anteriorly directed proximal tibial shear forces and GRF, knee extension, posteriorly inclined tibia and GRF vector relative to vertical, posteriorly inclined GRF vector relative to the tibia, and anteriorly positioned COG relative to the ankle-joint center were assigned as positive (Figure 1).

Statistical Analyses

Separate univariate 2-way (sex $\times$ landing style) repeated-measures analyses of variance (RMANOVA), followed by Bonferroni pairwise comparisons as appropriate, were conducted for each variable. Degrees of freedom for RMANOVAs were corrected using the Greenhouse-Geisser procedure if Mauchly sphericity tests were different ($P <$ .05). Although sex differences were not our research question, we examined them to rule out possible sex-specific confounding factors because researchers[29,30] have frequently reported sex-specific neuromuscular differences in landing. When we observed interactions, we conducted separate 1-way RMANOVA followed by Bonferroni pairwise comparisons for each sex. To further confirm whether the hypothesized kinematic and kinetic links existed, we used Pearson product moment correlation coefficients to examine the relationships among targeted variables in SSL. In addition, the differences and associations between the time to peak tibial axial forces and GRF were analyzed for each condition using paired-samples t tests and Pearson product moment correlation analyses. These analyses were conducted because Cerulli et al[14] showed that peak ACL strain occurred almost simultaneously with peak GRF in their in vivo case study. We set the α level at .05. We used SPSS statistical software (version 22; IBM Corp, Armonk, NY) for analysis.

RESULTS

Kinematics

Degrees of freedom, F values, and P values for 2-way RMANOVAs for kinematic variables are presented in

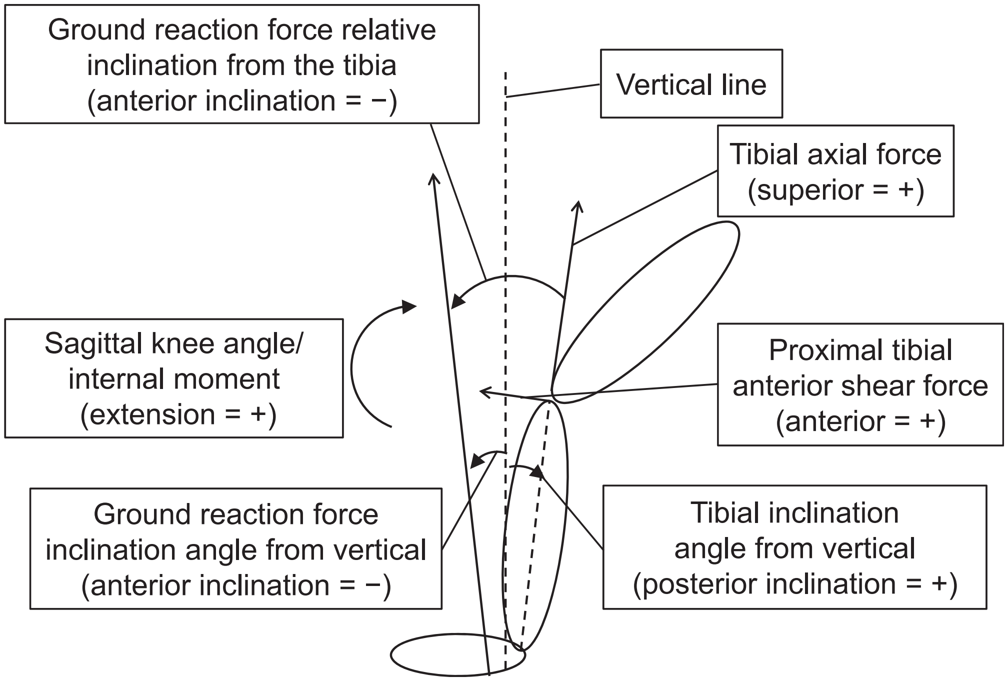

Figure 1. Directional conventions for targeted kinematic and kinetic variables.

Table 1. Lower extremity kinematics (mean ± SD) across the landing styles are presented in Table 2. From 1 participant, representative kinematic data normalized to 100 data points are shown in Figure 2.

The COG positions relative to the ankle joint indicated that COG was positioned the most posteriorly and anteriorly in URL and LFL, respectively, at the peak tibial axial forces, but those at the foot contact between SSL and URL did not differ. However, tibial inclination angles were different across landing styles. Specifically, the tibia was 1.4° ± 4.2° and 5.5° ± 4.8° posteriorly inclined relative to vertical in the SSL and URL, respectively, and 4.4° ± 6.0°

Table 1. Degrees of Freedom and *F* and *P* Values for 2-Way Repeated-Measures Analyses of Variance for Kinematic Variables

	Main Effects			Landing × Sex	
Variable	Degrees of Freedom	*F* Value	*P* Value	*F* Value	*P* Value
Center-of-gravity positions relative to the ankle joint at foot contact					
Men					
Women					
All	1.341	9.346	.003	0.115	.81
Center-of-gravity positions relative to the ankle joint at peak tibial axial force					
Men					
Women					
All	2	60.242	<.001	0.258	.77
Sagittal-plane ground reaction force angle from vertical					
Men					
Women					
All	1.358	0.737	.439	0.473	.567
Sagittal-plane tibial inclination angle from vertical[a]					
Men	1.156	11.866	.001		
Women	2	39.723	<.001		
All	1.468	47.943	<.001	6.427	.01
Relative ground reaction force angle to tibia[a]					
Men	1.068	12.861	.005		
Women	2	31.207	<.001		
All	1.448	42.204	<.001	4.092	.04
Knee-flexion angle[a]					
Men	1.357	24.111	<.001		
Women	1.145	28.479	<.001		
All	1.388	37.376	<.001	4.062	.043

[a] Values for 1-way repeated-measures analyses of variance for each sex are shown when a landing × sex interaction was observed.

Table 2. Descriptive Statistics (Mean ± SD) and Results for 2-Way Repeated-Measures Analyses of Variance for Kinematic Variables Across Landing Styles

Variable	Landing Style		
	SSL	LFL	URL
Center-of-gravity positions relative to the ankle joint at foot contact, m			
Men	−0.125 ± 0.018	−0.100 ± 0.02	−0.130 ± 0.018
Women	−0.124 ± 0.029	−0.091 ± 0.044	−0.123 ± 0.049
All[a]	−0.125 ± 0.023	−0.096 ± 0.034[c,d]	−0.127 ± 0.036
Center-of-gravity positions relative to the ankle joint at peak tibial axial force, m			
Men	−0.054 ± 0.059	−0.023 ± 0.069	−0.099 ± 0.051
Women	−0.086 ± 0.053	−0.044 ± 0.054	−0.127 ± 0.045
All[a]	−0.070 ± 0.057	−0.034 ± 0.061[c,e]	−0.113 ± 0.049[g]
Sagittal-plane ground reaction force angle from vertical, °			
Men	−0.7 ± 2.6	0.2 ± 1.6	−0.5 ± 3.2
Women	2.3 ± 2.1	2.7 ± 2.8	3.0 ± 2.7
All	0.8 ± 2.8	1.4 ± 2.6	1.3 ± 3.4
Sagittal-plane tibial inclination angle from vertical, °			
Men[a]	3.0 ± 4.3	−1.8 ± 5.9[c,e]	4.5 ± 5.0[h]
Women[a]	−0.1 ± 3.7	−6.9 ± 5.2[c,e]	6.4 ± 4.7[g]
All[a,b]	1.4 ± 4.2	−4.4 ± 6.0[c,e]	5.5 ± 4.8[g]
Relative ground reaction force angle to tibia, °			
Men[a]	3.7 ± 4.9	−2.0 ± 5.8[c,e]	5.0 ± 5.0[h]
Women[a]	−2.4 ± 4.6	−9.6 ± 6.2[c,e]	3.4 ± 5.0[g]
All[a,b]	0.7 ± 5.6	−5.8 ± 7.1[c,e]	4.2 ± 5.0[g]
Knee-flexion angle, °			
Men[a]	−21.6 ± 4.8	−27.0 ± 7.9[e,f]	−19.7 ± 4.6[h]
Women[a]	−23.1 ± 6.8	−30.4 ± 8.3[c,e]	−16.3 ± 8.0[g]
All[a,b]	−22.4 ± 5.8	−28.7 ± 8.1[c,e]	−18.0 ± 6.6[g]

Abbreviations: LFL, landing with body leaning forward; SSL, landing with self-selected style; URL, landing with upright posture.
[a] Indicates main effect ($P < .01$).
[b] Indicates interaction ($P < .05$).
[c] Indicates difference between LFL and SSL ($P < .01$).
[d] Indicates difference between URL and LFL ($P < .05$).
[e] Indicates difference between URL and LFL ($P < .01$).
[f] Indicates difference between LFL and SSL ($P < .05$).
[g] Indicates difference between URL and SSL ($P < .01$).
[h] Indicates difference between URL and SSL ($P < .05$).

anteriorly inclined relative to vertical in the LFL. Similarly, GRF angles relative to the tibia differed across landing styles. The GRF vector was 0.7° ± 5.6° and 4.2° ± 5.0° anteriorly inclined relative to the tibia in SSL and URL, respectively, and was 5.8° ± 7.1° posteriorly inclined relative to the tibia in LFL.

We observed interactions in the time to peak tibial axial forces, tibial inclination angles relative to vertical, GRF angles relative to the tibia, and knee-flexion angle. Whereas an interaction was observed in these variables, the directionality of differences among landing styles was the same in men and women.

Kinetics

Degrees of freedom, *F* values, and *P* values for 2-way RMANOVAs for kinetic variables are presented in Table 3. Lower extremity kinetics (mean ± SD) across the landing styles are presented in Table 4. From 1 participant, representative kinetic data normalized to a 100% cycle are shown in Figure 3. Differences in the directions of GRF and tibiofemoral forces relative to the tibia between URL and LFL are presented in Figure 4.

On average, the time to peak tibial axial forces during URL was less than 40 milliseconds after ground contact, which was shorter than the time during SSL and LFL. In contrast, during LFL, the participants took 1.3 and 1.8 times longer to reach peak tibial axial forces than during SSL and URL, respectively.

At the time of the peak tibial axial forces, we observed no differences in GRF angles relative to vertical across landing styles. We found main effects of peak tibial axial and proximal tibial shear forces with no interactions. Peak tibial axial forces reached 4.9 BW in URL, which was 1.1 and 1.4 times greater than in SSL and LFL, respectively. In contrast, LFL produced lower peak tibial axial forces than the other landing styles. The proximal tibial anterior shear forces in URL were an average of 5 times larger than in SSL, whereas LFL resulted in posteriorly directed proximal tibial shear forces at peak tibial axial forces.

We observed main effects for GRF_V. The GRF_V was 1.1 and 1.3 times greater in URL than in SSL and LFL, respectively, and was smaller in LFL than in the other landing styles. No main effect or interaction was found for GRF_{AP} and knee-extensor moment.

Associations Between Variables

A correlation matrix for kinematic and kinetic variables in SSL is shown in Table 5. We noted strong positive associations between tibial inclination and knee-flexion angles, indicating that smaller knee-flexion angles were associated with smaller tibial anterior-inclination angles. Smaller tibial anterior-inclination angles and greater GRF angles relative to vertical were associated with greater GRF vector of anterior inclination relative to the tibia, which is strongly associated with greater proximal tibial anterior and axial forces as well. The directionalities of these relationships were consistent with the aforementioned results of ANOVAs.

Differences and Associations Between Time to Peak Tibial Axial Force and GRF

The means and standard deviations for time to peak GRF during SSL were 52 ± 8 milliseconds, during LFL were 66 ± 14 milliseconds, and during URL were 36 ± 14 milliseconds. We noted differences in the time to peak tibial axial force and peak GRF between SSL and URL (both $P < .01$), whereas no difference was found between these variables during LFL ($P = .34$). Correlation analyses showed very high positive correlations between the 2 time variables in all conditions (SSL: $r = 0.989$; LFL: $r = 0.992$; URL: $r = 0.997$; all $P < .001$).

Whereas differences existed in the time to peak tibial axial force and peak GRF between SSL and URL, they were extremely small and systematic: the average differences between SSL and URL were both 1 millisecond (range = –2 to 3 milliseconds for SSL and –1 to 4 milliseconds for URL). The mean and range of the time difference during LFL were 0 milliseconds and –4 to 4 milliseconds, respectively. Therefore, although differences were present in time to peak during SSL and URL, the peak tibial axial forces and peak GRF occurred almost simultaneously in all conditions.

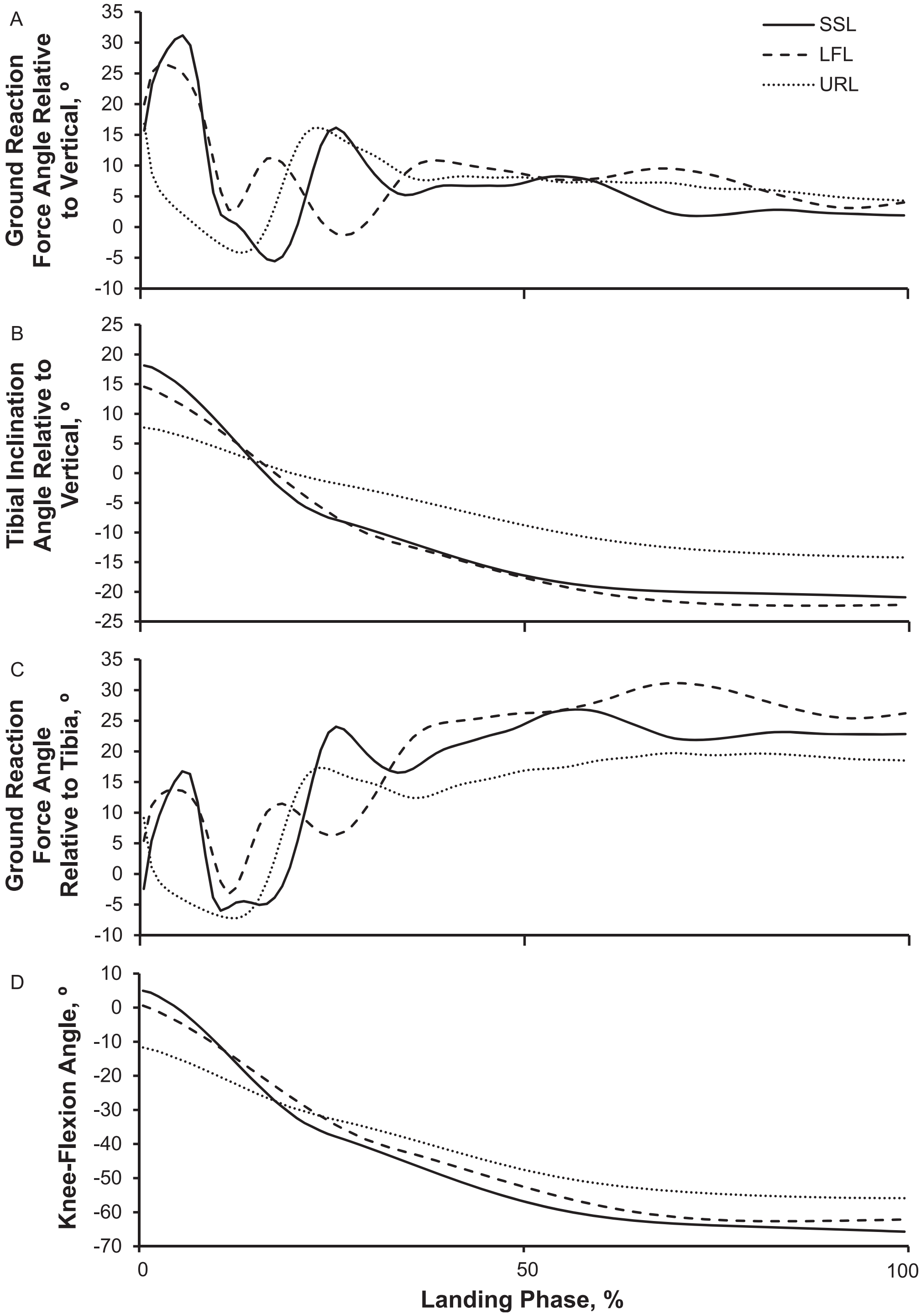

Figure 2. Representative percentage time series data for, A, sagittal-plane ground reaction force angles relative to vertical, B, shank inclination angles relative to vertical, C, ground reaction force angles relative to shank, and D, knee-flexion angles in 3 different landing styles in 1 participant. Abbreviations: LFL, landing with body leaning forward; SSL, landing with self-selected style; URL, landing with upright posture.

DISCUSSION

Changes in sagittal-plane body position altered not only the amount but also the direction of impact and tibiofemoral forces relative to the tibia in the sagittal plane during single-legged landings. Participants successfully modified their landing styles as their COG became more or less posterior to the ankle-joint centers in URL and LFL than SSL, respectively. These results indicate that URL and LFL produced more "leaning-back" and "leaning-forward" landings, respectively, than SSL. Changes in sagittal-plane body position resulted in most of the examined kinematic and kinetic variables in URL and LFL differing from those in SSL.

Furthermore, the directionality of these differences from SSL was generally opposite between URL and LFL (ie, if 1 variable increased in URL, it decreased in LFL and vice

Table 3. Degrees of Freedom and *F* and *P* Values for 2-Way Repeated-Measures Analysis of Variance for Kinetic Variables

	Main Effects			Landing × Sex	
Variable	Degrees of Freedom	*F* Value	*P* Value	*F* Value	*P* Value
Time to peak tibial axial force[a]					
Men	2	24.111	<.001		
Women	2	31.207	<.001		
All	1.554	54.437	<.001	4.032	.03
Peak tibiofemoral axial force					
Men					
Women					
All	1.507	60.62	<.001	0.304	.68
Proximal tibial shear force					
Men					
Women					
All	2	40.649	<.001	1.562	.22
Vertical ground reaction force					
Men					
Women					
All	1.441	53.414	<.001	0.463	.57
Horizontal ground reaction force					
Men					
Women					
All	1.32	0.859	.39	1.826	.19
Knee-extensor moment					
Men					
Women					
All	1.289	5.642	.02	0.777	.42

[a] Values for 1-way repeated-measures analyses of variance for each sex are shown when a landing × sex interaction was observed.

Table 4. Descriptive Statistics and Results for 2-Way Repeated-Measure Analyses of Variance for Kinetic Variables Across Landing Styles (Mean ± SD)

	Landing Style		
Variable	SSL	LFL	URL
Time to peak tibial axial force, ms			
Men[a]	50 ± 7	60 ± 8[c,d]	39 ± 12[e]
Women[a]	57 ± 8	73 ± 14[c,d]	35 ± 16[e]
All[a,b]	53 ± 8	67 ± 13[c,d]	37 ± 14[e]
Peak tibial axial force, × BW			
Men	4.7 ± 0.5	3.9 ± 0.5	5.2 ± 0.6
Women	3.9 ± 0.6	3.3 ± 0.6	4.6 ± 0.7
All[a]	4.3 ± 0.7	3.6 ± 0.7[c,d]	4.9 ± 0.7[e]
Proximal tibial shear force, × BW			
Men	0.3 ± 0.4	−0.1 ± 0.5	0.6 ± 0.5
Women	−0.1 ± 0.4	−0.5 ± 0.4	0.4 ± 0.4
All[a]	0.1 ± 0.4	−0.3 ± 0.5[c,d]	0.5 ± 0.5[e]
Vertical ground reaction force, × BW			
Men	5.0 ± 0.6	4.2 ± 0.5	5.4 ± 0.7
Women	4.1 ± 0.6	3.5 ± 0.6	4.8 ± 0.8
All[a]	4.5 ± 0.7	3.8 ± 0.6[c,d]	5.1 ± 0.8[e]
Horizontal ground reaction force, × BW			
Men	−0.1 ± 0.2	0.0 ± 0.1	−0.1 ± 0.3
Women	0.2 ± 0.2	0.1 ± 0.1	0.3 ± 0.2
All	0.0 ± 0.2	0.1 ± 0.2	0.1 ± 0.3
Knee-extensor moment			
Men	−0.057 ± 0.084	−0.002 ± 0.076	−0.071 ± 0.125
Women	0.026 ± 0.110	0.036 ± 0.086	−0.031 ± 0.133
All	−0.016 ± 0.104	0.017 ± 0.081	−0.051 ± 0.127

Abbreviations: BW, body weight; LFL, landing with body leaning forward; SSL, landing with self-selected style; URL, landing with upright posture.
[a] Indicates main effect ($P < .01$).
[b] Indicates interaction ($P < .05$).
[c] Indicates difference between LFL and SSL ($P < .01$).
[d] Indicates difference between URL and LFL ($P < .01$).
[e] Indicates difference between URL and SSL ($P < .01$).

versa). The directionalities of the relationships in the correlation analyses were also consistent with the results we found using the analyses of variance. Specifically, in URL, participants demonstrated reduced shock-attenuation capacities, resulting in greater GRF, less knee-flexion angle, and smaller tibial anterior-inclination angles at peak tibial axial forces. Consequently, the GRF vector was directed more anterior to the tibial longitudinal axis in the sagittal plane, resulting in the greatest tibial axial and anterior shear forces among the 3 landing styles (Figure 4). In contrast, in LFL, participants had increased shock-attenuation capacity, which resulted in the lowest GRF and largest knee-flexion and tibial anterior-inclination angles at peak tibial axial forces. Therefore, in LFL, participants received posteriorly directed tibial shear forces at the peak tibial axial forces (Figure 4). In addition, peak tibial axial forces in LFL were the lowest among the 3 landing styles. All changes in the targeted kinematic and kinetic variables across the landing conditions were supported by associations among these variables during SSL, which participants performed without specific instructions (eg, earlier times to peak tibial axial forces were associated with smaller knee-flexion angles, which were also associated with less anterior tibial inclination relative to vertical). Therefore, such changes in lower extremity kinematic and kinetic variables across conditions were not due to the instructions for performing specific landing styles but were due to the intersegmental kinematic and kinetic links (ie, the kinetics and kinematics of 1 segment influenced the kinetics and kinematics of the adjacent segment).

We found differences in tibial inclination angles relative to vertical and GRF angles relative to the tibia across conditions but no difference in GRF angles from vertical across conditions. These results indicate that the biomechanical factor to modulate the GRF angle relative to the tibia at the peak tibial axial force should not be the GRF angles from vertical but the tibial inclination angles from vertical. Given that tibial inclination angles and knee-flexion angles were highly associated, increasing the knee-flexion angles when absorbing the GRF must be an important biomechanical factor that modifies the GRF angles relative to the tibia and, thus, the direction of tibiofemoral forces at the time of the impact phase of the landing.

Sex-specific interactions were observed in some kinematic data; however, the directionalities of sex-specific differences among landing styles were almost the same between men and women. Therefore, modifying landing styles led to similar kinetic and kinematic changes between men and women. These results generally support our

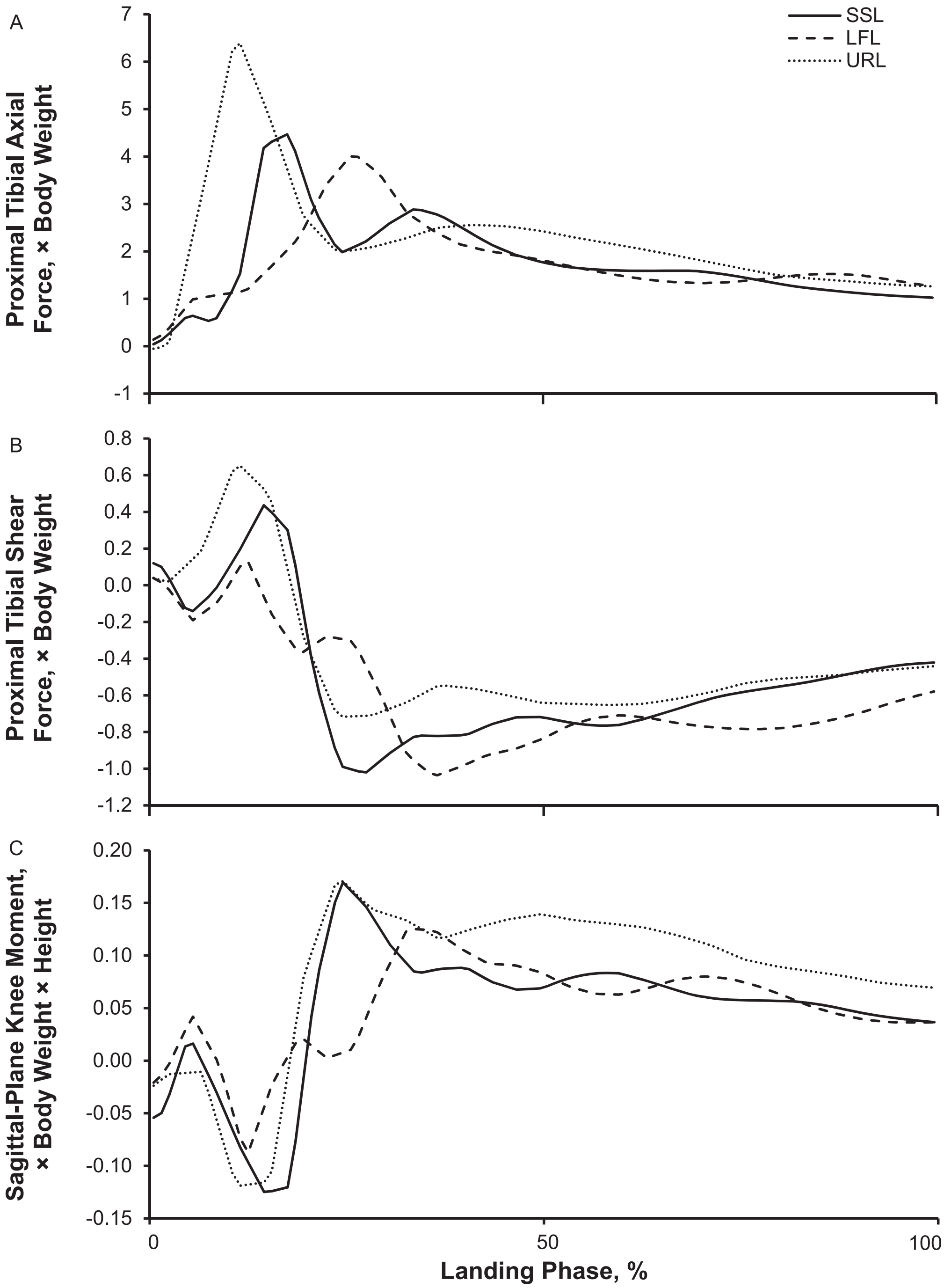

Figure 3. Representative percentage time series data for, A, proximal tibial axial forces, B, proximal tibial shear forces, C, sagittal-plane internal knee moments, D, vertical ground reaction forces, and E, horizontal ground reaction forces in 3 different landing styles in 1 participant. Data were normalized by body weight or body weight and height in meters. Abbreviations: LFL, landing with body leaning forward; SSL, landing with self-selected style; URL, landing with upright posture.

hypothesized intersegmental kinematic and kinetic links during landing.

We analyzed the time point at which the tibial axial forces peaked because authors[18,19] of cadaveric studies have demonstrated that tibial axial force is one of the critical forces that cause noncontact ACL injuries. Meyer and Haut[18,19] investigated whether tibial axial force ruptured the ACL in human cadaver knees and found that increasing tibial axial force increased tibial anterior shift and internal tibial rotation relative to the femur. The ACL of each specimen was ruptured at 5.4 ± 2 kN of applied tibial axial force. The authors theorized that tibial axial force is transformed into tibial anterior shear force due to the posterior slope of the lateral tibial plateau. This proximal tibial anterior shear force shifts the lateral tibial plateau anteriorly, leading the ACL to eventually rupture.[18] Supporting this potential mechanism of ACL injury, several investigators[31–35] have shown relationships between the posterior slope of the tibial plateau and knee biomechanics or ACL injury risk. Meyer and Haut[18] suggested that sustaining a large GRF during sudden decelerating motions could harm the ACL because it increases the amount of

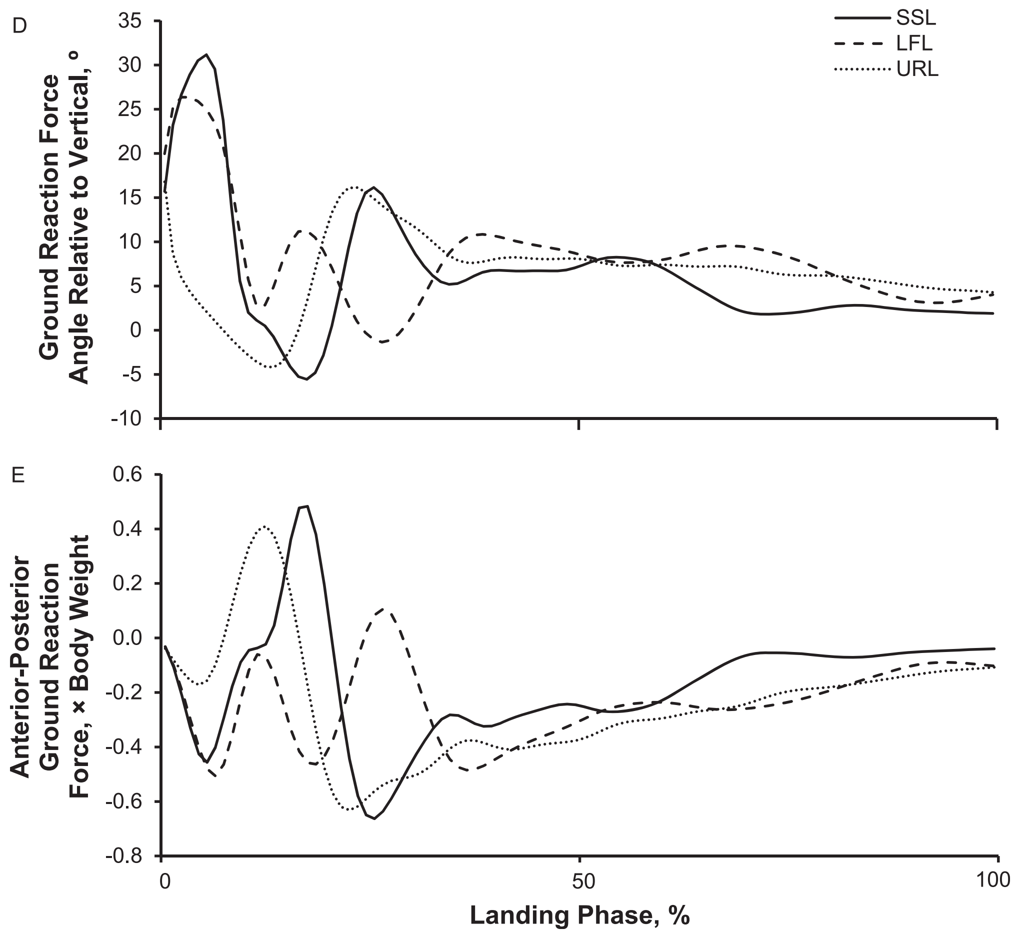

Figure 3. Continued from previous page.

tibial axial force and, thus, the proximal tibial anterior shear force due to the posterior slope of the tibial plateau. Therefore, comparing lower extremity kinetics and kinematics at peak tibial axial forces among the 3 landing styles and examining the relationships among these variables should be important in understanding ACL injury mechanisms.

The ACL directs restraint against tibial anterior shear forces, especially at shallow knee-flexion angles (ie, <30°),[36] and researchers have examined factors associated with tibial anterior shear force magnitude during sudden deceleration motions. The peak tibial anterior shear forces during sharp decelerating tasks range from 0.2 to 0.8 (× BW).[37–42] Whereas the tasks and time points used for the analyses differ somewhat among these studies and our study, all of these tasks involve sudden deceleration motions, and the tibial anterior shear forces that we generally observed in SSL and URL are similar to those in the literature.

The tibial anterior shear forces in our study were expressed as the horizontal component of the tibiofemoral forces, which are parallel to the anteroposterior axis of the shank coordinate system. Given that this anteroposterior axis is perpendicular to the tibial longitudinal axis, the tibial anterior shear force in our study represents the tibial anterior shear force when no posterior slope exists on the tibial plateau. Theoretically, when a tibial posterior slope exists, the tibial anterior shear force magnitude is largely influenced by the magnitude of tibial axial force. For example, Shao et al[43] compared the maximum tibial anterior shear forces during walking with different posterior tibial plateau slopes. They found that the maximal tibial anterior shear forces during walking increased from 0.28 to 0.58 BW when the tibial posterior slope increased from 4° to 8°.

The URL resulted in the largest tibial anterior shear force (0.5 BW), which was much smaller than the reported ultimate load for ACL ruptures (2160 ± 157 N, corresponding to approximately 3.1 BW based on the average BW of participants).[44] However, using the average values for peak tibial axial (4.9 BW) and shear (0.5 BW) forces observed in URL and the Pythagorean theorem, the resultant tibiofemoral force was 4.9 BW with a 5.8° anterior inclination relative to the tibial longitudinal axis (ie, derived from $\tan^{-1}$ [0.5/4.9]; Figure 5). In their cadaveric study, Meyer et al[45] showed that the average posterior slope of the tibial lateral plateau was 14.6° ± 4.8° and in the posterior part of the tibial plateau was 17.8° ± 4.7°. Therefore, if participants had a tibial plateau with a 15° posterior slope, for example, the average tibial anterior shear force would increase from 0.5 to 1.74 BW in URL. Whereas this is a simplistic estimate of the effects of posterior slope of the tibial plateau, it is logical that larger and more anteriorly inclined resultant tibiofemoral forces would magnify tibial anterior shear forces in the presence of a tibial posterior slope. Therefore, an individual who

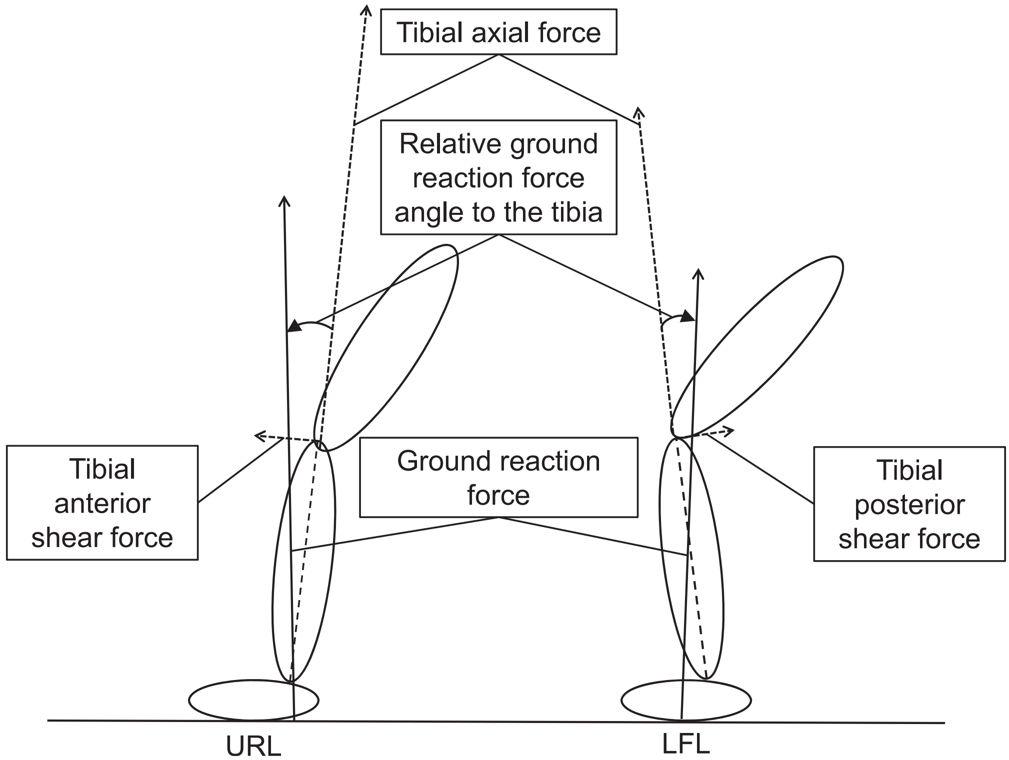

Figure 4. Differences in the directions of ground reaction and tibiofemoral forces relative to the tibia between landing flat-footed with the body upright (URL) and landing on the toes with the body leaning forward (LFL).

sustains large tibial axial forces and has a tibia with a greater posterior slope would theoretically experience magnified proximal tibial anterior shear forces.

We observed no difference in knee-extensor moment at the peak tibial axial forces. This result is not consistent with the work of Shimokochi et al,[16] who found the largest and smallest peak knee-extensor moments in URL and LFL, respectively. As a possible explanation for this disparity between the work of Shimokochi et al[16] and our work, we hypothesize that peak tibial axial forces occur much earlier than the peak knee-extensor moment and that the knee has insufficient time to develop knee-extensor moments. To examine this explanation, we compared time to peak tibial axial force and time to peak knee-extensor moment from the kinetic data for each landing style using paired-samples t tests. We observed that time to peak tibial axial forces occurred 20, 28, and 26 milliseconds earlier than time to peak knee-extensor moment in SSL ($t_{19} = 4.139$, $P < .001$, effect size = 0.80), LFL ($t_{19} = 6.606$, $P < .001$, effect size = 1.13), and URL ($t_{19} = 4.129$, $P < .001$, effect size = 1.18), respectively. Therefore, at peak tibial axial forces, knee-extensor muscles may not have had sufficient time to develop knee-extensor moments, resulting in no difference in knee-extensor moments among the various landing styles. This notion may be supported by a simulation study in which researchers[46] reported that the quadriceps forces at 50 milliseconds after foot contact at landing were much lower than the maximum quadriceps forces, even with maximum pre-activation of the quadriceps muscles before foot contact. In addition, the electromechanical delay in

Table 5. Correlation Matrix for Kinetic and Kinematic Variables in Self-Selected Landing

Variable	Tibial Shear Forces	Knee-Extensor Moment	Vertical GRF	Anteroposterior GRF	Tibial Inclination Angles From Vertical	GRF Angles From Vertical	GRF Angles From the Tibia	Knee-Flexion Angles	Time to Peak Tibial Axial Forces
Peak tibial axial forces	0.75[a]	−0.38	0.99[a]	−0.54[b]	0.70[a]	−0.59[a]	0.82[a]	0.51[b]	−0.78[a]
Tibial shear forces		−0.80[a]	0.76[a]	−0.63[a]	0.83[a]	−0.67[a]	0.96[a]	0.60[a]	−0.69[a]
Knee-extensor moment			−0.40	0.59[a]	−0.56[b]	0.61[a]	−0.72[a]	−0.35	0.46[b]
Vertical GRF				−0.60[a]	0.68[a]	−0.64[a]	0.83[a]	0.50[b]	−0.82[a]
Anteroposterior GRF					−0.19	0.99[a]	−0.64[a]	−0.03	0.40
Tibial inclination angles from vertical						−0.25	0.88[a]	0.78[a]	−0.74[a]
GRF angles from vertical							−0.69[a]	−0.09	0.46[b]
GRF angles from the tibia								0.63[a]	−0.78[a]
Knee-flexion angles									−0.63[a]

Abbreviation: GRF, ground reaction force.

[a] Indicates difference ($P < .01$).

[b] Indicates difference ($P < .05$).

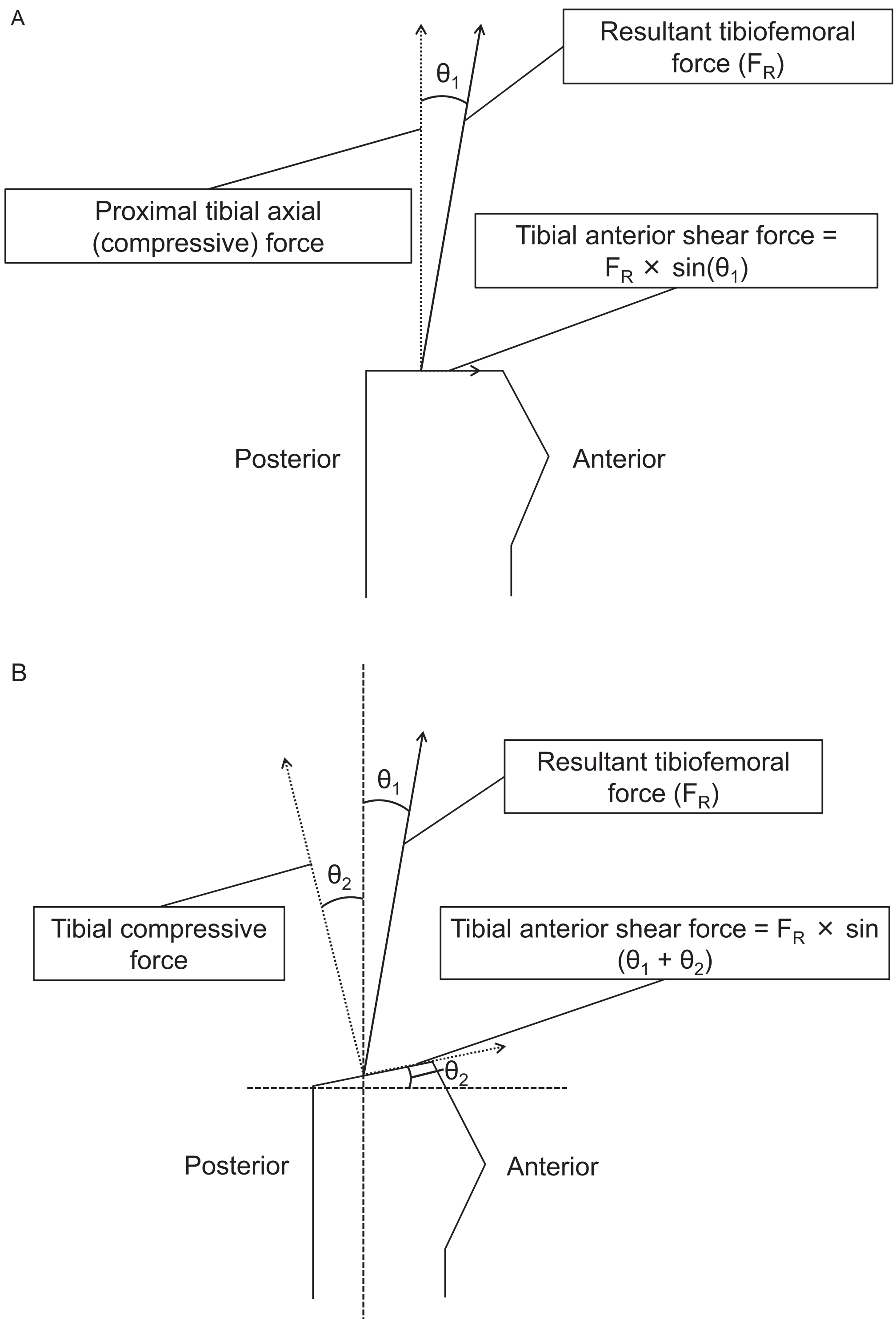

Figure 5. Diagrams describe the comparison of the amount of tibial anterior shear force, A, without and, B, with tibial posterior slope with the same resultant tibiofemoral force (the tibial force acting against the femur). θ_1 indicates the angle between the resultant tibiofemoral force and tibial longitudinal axis, and θ_2 indicates the angle between the tibial plateau and horizontal. In A, the tibial axial force is the force acting parallel to the longitudinal axis of the tibia and tibial plateau and acting perpendicular to the tibial plateau. However, in B, the force acting perpendicular to the tibial plateau is no longer parallel to the longitudinal axis of the tibia because of the posterior slope of the tibial plateau and, thus, was defined as tibial compressive force.

converting reflexive activity (ie, landing) to muscle and mechanical force generation is around 50 to 100 milliseconds.[47] Participants may not have achieved full quadriceps contraction forces to produce peak knee-extensor moments within which the tibial axial force reaches its peak (around 40 milliseconds). In fact, the mean values of knee-extensor moments at peak tibial axial forces (–0.02 to 0.02) were much smaller than peak knee-extensor moments (0.09 to 0.14) in our biomechanical dataset. In a cadaveric study, Wall et al[48] demonstrated that smaller tibial axial forces were necessary to rupture the ACL when quadriceps contraction forces were added to the knee joint. Therefore, when large tibial axial forces and knee-extensor moments (and thus, quadriceps contraction forces) occur simultaneously, the risk of noncontact ACL injuries would increase further. In future studies, researchers should investigate situations that would cause such concurrent combined knee loading during sharp decelerating motions.

We acknowledge that our study had limitations. Whereas knee loadings during noncontact ACL injuries are multiplanar,[3,49] we examined only sagittal-plane kinematics and kinetics. Therefore, other factors and knee loadings in other planes of motion may further increase ACL loadings. Still, we provided important information for understanding mechanisms of noncontact ACL injuries, as Sheehan et al[17] reported sagittal-plane body positions as an influence on noncontact ACL injury mechanisms from their video analyses. We also acknowledge an inherent limitation in synchronizing multiple streaming data sampled at different frequencies and using hardware with different internal clocks. Theoretically, our data may not have matched exactly for up to approximately 4.2 milliseconds on average because we sampled kinematic and kinetic data at 120 and 1000 Hz, respectively, and because the force plate and electromagnetic tracking system use different independent internal clocks. In addition, we used the average data from 5 trials as representative data for each variable in each condition for each participant. Therefore, although we linearly interpolated kinematic data to align with kinetic data and minimize the error, the values of the joint moments and forces were possibly underestimated or overestimated to some degree. In future studies, researchers should aim to identify other factors that may further magnify ACL loadings.

CONCLUSIONS

Our study adds important knowledge about intersegmental kinetic and kinematic links. We showed how changing sagittal-plane body positions would lead to changes in the magnitude and direction of impact that, in turn, affects tibiofemoral force magnitude and direction relative to the tibia during single-legged landing. The URL position resulted in less time to peak tibial axial force, greater GRF, and a more anteriorly inclined GRF vector relative to the tibia. These kinematics and kinetics, in turn, may lead to greater peak tibial axial and anterior shear forces that potentially harm the ACL. Conversely, the LFL position resulted in longer time to peak tibial axial forces, reduced GRF, and a more posteriorly inclined GRF vector relative to the tibia. Therefore, it led to overall lower tibial axial and greater proximal tibial posterior shear forces, which protect the ACL. These results also indicated that modifying kinematic factors, such as increasing knee flexion at the time of peak GRF and tibiofemoral forces, would modulate the GRF angle relative to the tibia and the tibiofemoral forces, thereby reducing the proximal tibial anterior shear forces. Such information may be incorporated in the education provided to athletes by athletic trainers for consistent, safe performance of abrupt decelerating movements and in the design of a rational intervention program that enables athletes to practice these movements and prevent noncontact ACL injuries.

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