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Lower Extremity Energy Absorption and Biomechanics During Landing, Part II: Frontal-Plane Energy Analyses and Interplanar Relationships

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Context: Greater sagittal-plane energy absorption (EA) during the initial impact phase (INI) of landing is consistent with sagittal-plane biomechanics that likely increase anterior cruciate ligament (ACL) loading, but it does not appear to influence frontal-plane biomechanics. We do not know whether frontal-plane INI EA is related to high-risk frontal-plane biomechanics.

Objective: To compare biomechanics among INI EA groups, determine if women are represented more in the high group, and evaluate interplanar INI EA relationships.

Design: Descriptive laboratory study.

Setting: Research laboratory.

Patients or Other Participants: Participants included 82 (41 men, 41 women; age = 21.0 ± 2.4 years, height = 1.74 ± 0.10 m, mass = 70.3 ± 16.1 kg) healthy, physically active volunteers.

Intervention(s): We assessed landing biomechanics with an electromagnetic motion-capture system and force plate.

Main Outcome Measure(s): We calculated frontal- and sagittal-plane total, hip, knee, and ankle INI EA. Total frontal-plane INI EA was used to create high, moderate, and low tertiles. Frontal-plane knee and hip kinematics, peak vertical and posterior ground reaction forces, and peak internal knee-varus moment (pKVM) were identified and compared across groups using 1-way analyses of variance. We used a χ^2 analysis to evaluate male and female allocation to INI EA groups. We used simple, bivariate Pearson product moment correlations to assess interplanar INI EA relationships.

Results: The high–INI EA group exhibited greater knee valgus at ground contact, hip adduction at pKVM, and peak hip adduction than the low–INI EA group ($P < .05$) and greater peak knee valgus, pKVM, and knee valgus at pKVM than the moderate– ($P < .05$) and low– ($P < .05$) INI EA groups. Women were more likely than men to be in the high–INI EA group ($\chi^2 = 4.909$, $P = .03$). Sagittal-plane knee and frontal-plane hip INI EA ($r = 0.301$, $P = .006$) and sagittal-plane and frontal-plane ankle INI EA were associated ($r = 0.224$, $P = .04$). No other interplanar INI EA relationships were found ($P > .05$).

Conclusions: Greater frontal-plane INI EA was associated with less favorable frontal-plane biomechanics that likely result in greater ACL loading. Women were more likely than men to use greater frontal-plane INI EA. The magnitudes of sagittal- and frontal-plane INI EA were largely independent.

Key Words: anterior cruciate ligament, kinetics, kinematics

Key Points

- Greater frontal-plane energy absorption (EA) during the initial 100 milliseconds of landing (INI) was associated with a less favorable frontal-plane biomechanical profile that may contribute to greater anterior cruciate ligament (ACL) loading.
- Women were 3.6 times more likely than men to use a landing strategy with greater frontal-plane INI EA.
- The magnitude of sagittal-plane INI EA generally did not influence the magnitude of frontal-plane INI EA during double-legged jump landings, suggesting the lack of an interplanar EA relationship during the 100 milliseconds immediately after initial ground contact.
- Individuals landing with elevated sagittal- and frontal-plane INI EA may be at increased risk of noncontact ACL injury because they would experience greater combined sagittal- and frontal-plane ACL loading.

The risk of noncontact anterior cruciate ligament (ACL) injury is 2 to 8 times greater in females than in males.[1,2] Accordingly, much research has been focused on identifying neuromechanical differences between sexes to discover the underlying mechanism for noncontact ACL injury.[3–7] Investigators[4,6,8] have identified greater knee-valgus angle at initial ground contact (IGC) and peak knee-valgus angle during landing in females than males. Furthermore, frontal-plane knee loading has been shown both in vivo using biomechanical modeling[9,10] and in vitro[11] to contribute to greater ACL loading, and knee-valgus angle and frontal-plane knee moment have been identified as predictors of noncontact ACL injury risk.[12] Consequently, researchers[13] have advocated limiting these frontal-plane biomechanical factors to decrease ACL injury risk.

We demonstrated in part I of this investigation that greater sagittal-plane lower extremity energy absorption (EA) during the initial impact phase (INI; ie, 100 milliseconds after IGC) of double-legged jump landings resulted in a sagittal-plane biomechanical profile that appeared to contribute to greater ACL loading due to sagittal-plane mechanisms.[14] However, we did not identify differences in frontal-plane knee kinematics or kinetics among INI EA groups, indicating that the magnitude of sagittal-plane INI EA did not directly influence frontal-plane landing biomechanics thought to contribute to ACL loading.[14] Furthermore, women, who represent greater ACL injury risk, were not more likely to be in the high sagittal-plane INI EA group than men.[14] This equal likelihood for men and women to use a higher-risk sagittal-plane landing strategy suggests that the greater ACL injury risk in females may be attributable at least partially to differences in frontal-plane landing strategies. Therefore, we wanted to expand this energetic analysis beyond the sagittal plane to evaluate whether frontal-plane INI EA influences ACL loading attributable to frontal-plane mechanisms.

To our knowledge, only 1 report in which investigators directly evaluated frontal-plane EA during landing has been published. Using a double-legged drop-landing task, Yeow et al[15] observed greater frontal-plane EA at the hip and knee than the ankle and an increase in frontal-plane EA in response to increased landing height. However, the authors did not specifically evaluate the relationships between frontal-plane EA and frontal-plane biomechanics that have been associated with ACL injury.[15]

Pollard et al[16] reported that female athletes exhibiting greater combined peak hip and knee flexion during double-legged drop landings displayed more sagittal-plane EA but less peak knee-valgus angle and average internal knee-varus moment (ie, the internal response to an external knee-valgus moment). They postulated that the greater sagittal-plane EA in the high-flexion group necessitated less frontal-plane knee EA, reducing ACL loading due to frontal-plane mechanisms and indicating a potential interplanar EA relationship whereby greater sagittal-plane knee EA (eccentric contraction of the quadriceps) could enhance frontal-plane support. This notion is supported by Lloyd and Buchanan[17] and Lloyd et al,[18] who demonstrated that the quadriceps and hamstrings musculature, which primarily serve a sagittal-plane function, can resist varus-valgus loading of the knee during both isometric and dynamic tasks primarily via cocontraction. However, Pollard et al[16] reported sagittal-plane EA calculated from IGC to peak knee flexion. Researchers[19] have indicated that greater EA during the initial impact period is unfavorable for ACL loading, whereas greater EA later in the landing phase is more desirable. Furthermore, given that peak ACL loading[20,21] and injury[22] occur within the first 100 milliseconds of landing, evaluating INI EA may be more applicable when assessing factors related to ACL injury risk. Therefore, the longer time interval used by Pollard et al[16] to calculate EA potentially confounds their proposed relationship between sagittal- and frontal-plane EA and frontal-plane biomechanical ACL injury risk factors. In contrast, when EA was calculated over the INI of landing, Norcross et al[14] reported no differences in frontal-plane landing biomechanics among groups displaying different magnitudes of sagittal-plane INI EA and suggested that interplanar EA relationships do not exist. However, the principal limitation of both studies is that neither group of authors specifically calculated frontal-plane EA, leaving the potential influence of frontal-plane EA on frontal-plane landing biomechanics and interplanar INI EA relationships purely speculative.[14,16]

Therefore, the purpose of our study was to expand on our sagittal-plane INI EA analyses by (1) ascertaining if frontal-plane INI EA is associated with meaningful differences in frontal-plane biomechanics by comparing the magnitudes of these biomechanical variables among groups displaying high–, moderate–, and low–frontal-plane INI EA; (2) determining whether women are more likely than men to demonstrate high frontal-plane INI EA; and (3) explicitly evaluating the relationships between the magnitudes of sagittal- and frontal-plane INI EA. We hypothesized that the high–frontal-plane INI EA group would display less desirable biomechanical values than the moderate– and low–frontal-plane INI EA groups, women would be represented more than men in the high INI EA group, and sagittal- and frontal-plane INI EA would be largely independent.

METHODS

We describe our participants and procedures in part I of this study.[14]

Data Sampling and Reduction

Kinematic and kinetic data were sampled, processed, and reduced as described in part I of this study.[14] Using custom computer software (LabVIEW; National Instruments Corporation, Austin, TX), we calculated frontal- and sagittal-plane hip-, knee-, and ankle-joint power curves (P) as the product of angular velocity (ω) and net joint moment (M) ($P = M \times \omega$). The negative portions of the joint power curves were integrated to calculate negative mechanical joint work[23,24–26] during the initial impact[23,27] phase of landing. Negative joint work values, in which the net joint moment and joint angular velocity are in opposite directions, indicate eccentric muscle actions and represent EA by the muscle-tendon unit.[24,28] Next, total sagittal-plane and total frontal-plane joint work were calculated by summing the negative joint work at each joint in their respective planes during this time interval.[25,26,29] By convention, all INI EA values were positive (ie, large positive values represent greater EA). Total INI EA was calculated to quantify the coordinated actions of the lower extremity joints during the period in landing when ACL injury is thought to occur. We used the same custom software to identify frontal-plane knee angle at IGC, peak values for knee-valgus and hip-adduction angles during the interval from IGC to the minimal vertical position of the whole-body center of mass,[29,30] and peak vertical and posterior ground reaction force (GRF) and internal knee-varus moment during the INI of landing. We used different time intervals to identify these variables so peak kinetics that can contribute to joint loading were identified during the time when ACL injury likely occurs[20,22] and the true peak kinematic values during landing were identified correctly because these peak angles generally occur slightly more than 100 milliseconds after IGC. Frontal-plane knee and hip angles at the instant of peak knee-varus moment also were identified. By angular convention, hip adduction and knee varus were positive, whereas hip abduction and knee valgus were negative. The GRFs were normalized to the participant's body weight ($\times$ BW^{-1}), frontal-plane knee moment was normalized to the product of the participant's body weight and height (Ht) ($\times$ [BW $\times$ Ht]$^{-1}$), and INI EA was expressed as a percentage of the product of the participant's body weight and height (% BW $\times$ Ht). Before statistical analysis, we averaged all dependent variables across the 5 jump-landing trials of each participant.

Statistical Analysis

The magnitude of total frontal-plane INI EA was used to create 3 distinct frontal-plane INI EA groups (tertiles): high, moderate, and low. We used separate 1-way analysis of variance (ANOVA) models to make static comparisons across INI EA groups for each biomechanical factor. When ANOVA models were significant, we calculated pairwise comparisons with the Tukey honestly significant difference test to identify specific group differences for these dependent variables, and used a Pearson χ^2 test of association to determine if more women than men were represented in the high versus low INI EA groups. Simple, bivariate Pearson correlation coefficients were used to assess the relationships among total, hip, knee, and ankle INI EA in the frontal and sagittal planes.

After our primary analyses, we conducted several post hoc analyses using only the participants who were assigned to both high–sagittal-plane[14] and high–frontal-plane (high-high) INI EA groups and the participants who were assigned to both low–sagittal-plane[14] and low–frontal-plane (low-low) INI EA groups. A Pearson χ^2 test of association and the Mantel-Haenszel common odds ratio were used to determine if sex and high-high versus low-low INI EA group allocation were associated and if women were more likely than men to demonstrate both high–sagittal-plane and high–frontal-plane INI EA, respectively. Finally, the low-low and high-high INI EA groups were compared across key ACL-related biomechanical variables using independent-samples *t* tests.

All analyses were conducted using SPSS (version 17.0; IBM Corporation, Armonk, NY), and the α level was set a priori at equal to or less than .05.

RESULTS

Descriptive statistics and frequency counts by sex for the 3 frontal-plane INI EA groups are presented in Table 1. The means and standard deviations for frontal- and sagittal-plane INI EA are displayed in Table 2.

Participant allocation to tertiles based on total frontal-plane INI EA was successful in creating 3 distinct groups demonstrating high, moderate, and low frontal-plane EA ($F_{2,79} = 55.501$, $P < .001$; Table 1). The 1-way ANOVA detected INI EA group differences for frontal-plane knee angle at IGC ($F_{2,79} = 5.782$, $P = .005$), frontal-plane knee ($F_{2,79} = 12.947$, $P < .001$) and hip ($F_{2,79} = 4.890$, $P = .01$) angles at peak knee-varus moment, peak frontal-plane knee ($F_{2,79} = 19.874$, $P < .001$) and hip ($F_{2,79} = 4.529$, $P = .01$) angles, peak posterior GRF ($F_{2,79} = 4.030$, $P = .02$), and peak knee-varus moment ($F_{2,79} = 16.978$, $P < .001$) but no group differences for peak vertical GRF ($F_{2,79} = 0.424$, $P = .66$; Table 3). Post hoc testing revealed that the high INI EA group landed with greater knee-valgus angle at IGC than the low INI EA group ($P = .001$) and displayed greater peak knee-valgus angles during landing than both the moderate ($P < .001$) and low ($P < .001$) INI EA groups. The high INI EA group also exhibited greater peak knee-varus moment and greater knee-valgus angle at peak knee-varus moment than both the moderate ($P = .001$ and $P = .005$, respectively) and low ($P < .001$ for both) INI EA groups and greater hip adduction at peak knee-varus moment ($P = .01$) and greater peak hip-adduction angle ($P = .007$) than the low INI EA group. Women were more likely than men to be in the high INI EA group ($\chi^2 = 4.909$, $P = .03$). Correlation coefficients between sagittal- and frontal-plane EA during the INI phase of landing are displayed in Table 4. Greater sagittal-plane knee INI EA was associated with greater frontal-plane hip INI EA ($r = 0.301$, $P = .006$), and greater sagittal-plane ankle INI EA was associated with greater frontal-plane ankle INI EA ($r = 0.224$, $P = .04$). No other relationships between frontal- and sagittal-plane INI EA were identified ($P > .05$).

We did not identify an association between sex and high-high (2 men, 8 women) versus low-low (6 men, 3 women) INI EA group allocation in our post hoc analysis using a test of association ($\chi^2 = 4.232$, $P = .07$). However, the Mantel-Haenszel common odds ratio revealed that women were 8 times more likely than men to demonstrate both high sagittal-plane and high frontal-plane INI EA ($P = .05$).

Table 1. Frontal-Plane Initial Impact-Phase Energy-Absorption Frequency Counts by Sex and Group Descriptives

	Initial Impact-Phase Energy-Absorption Group		
Characteristic	High	Moderate	Low
Participants, No.			
Women	20	9	12
Men	7	19	15
Total	27	28	27
Age, y			
Women	21.05 ± 3.17	20.00 ± 1.50	20.92 ± 2.58
Men	21.14 ± 3.44	20.95 ± 1.62	21.20 ± 2.15
Total	21.07 ± 3.17	20.64 ± 1.62	21.07 ± 2.30
Height, m			
Women	1.66 ± 0.07	1.68 ± 0.06	1.67 ± 0.04
Men	1.80 ± 0.05	1.81 ± 0.06	1.82 ± 0.07
Total	1.69 ± 0.09	1.77 ± 0.09	1.75 ± 0.10
Mass, kg			
Women	59.46 ± 10.00	64.06 ± 8.68	61.91 ± 8.11
Men	78.80 ± 6.88	81.03 ± 19.61	77.86 ± 15.57
Total	64.47 ± 12.59	75.58 ± 18.54	70.77 ± 14.95
Total initial impact-phase energy absorption, % body weight × height			
Women[a,b]	3.07 ± 1.47	1.11 ± 0.21	0.60 ± 0.21
Men[a,b]	2.07 ± 0.72	1.29 ± 0.23	0.52 ± 0.20
Total[a,b]	2.81 ± 1.37	1.23 ± 0.24	0.55 ± 0.21
95% confidence interval for total initial impact-phase energy absorption			
Women	2.38, 3.76	0.95, 1.27	0.46, 0.73
Men	1.40, 2.74	1.17, 1.40	0.40, 0.63
Total	2.27, 3.35	1.14, 1.32	0.47, 0.63

[a] Indicates high–initial impact-phase energy-absorption group was different from low–initial impact-phase energy-absorption group ($P < .05$).

[b] Indicates high–initial impact-phase energy-absorption group was different from moderate–initial impact-phase energy-absorption group ($P < .05$).

Table 2. Sagittal- and Frontal-Plane Initial Impact-Phase Energy-Absorption Descriptives (Mean ± SD)

Initial Impact-Phase Energy Absorption, % body weight × height	Sagittal Plane[a]	Frontal Plane
Total	13.62 ± 3.02	1.53 ± 1.24
Hip	2.26 ± 1.34	0.20 ± 0.26
Knee	8.98 ± 2.69	1.05 ± 1.08
Ankle	2.37 ± 1.64	0.28 ± 0.32

[a] Adapted from Norcross et al.[14]

When compared across key ACL-related biomechanical variables, the high-high INI EA group demonstrated 7.5° greater knee valgus at IGC, 11.3° greater peak knee-valgus angle, 71% greater peak posterior GRF, 49% greater peak knee- extension moment, 33% greater peak anterior tibial shear force, and 117% greater peak knee-varus moment than the low-low INI EA group (all $P < .05$).

DISCUSSION

The principal findings of this investigation were that greater frontal-plane EA during the initial 100 milliseconds of landing was associated with a less favorable frontal-plane biomechanical profile that may contribute to greater ACL loading and that women were 3.6 times more likely than men to use a landing strategy with greater frontal-plane INI EA. In addition, the magnitude of sagittal-plane INI EA generally did not influence the magnitude of frontal-plane INI EA during double-legged jump landings, suggesting the lack of an interplanar EA relationship during the 100 milliseconds immediately after IGC.

The differences in biomechanical variables across the frontal-plane INI EA groups generally agreed with our hypotheses. We identified greater knee-valgus angle at IGC, peak knee-valgus and hip-adduction angles, peak knee-varus moment, knee-valgus and hip-adduction angles at peak knee-varus moment, and peak posterior GRF in the high than low INI EA group, and only peak vertical GRF did not differ among the INI EA groups (Table 3). The lack of group differences in peak vertical GRF was not surprising given our previous sagittal-plane INI EA analysis[14] and the generally equivocal results of investigations[23,25,31,32] in which researchers have compared peak vertical GRF between sexes (ie, higher and lower ACL injury risk). Regarding frontal-plane knee biomechanics, the mean differences between INI EA groups also appeared to be consequential when compared with findings reported in previous investigations. Hewett et al[12] noted that females who went on to sustain noncontact ACL injuries demonstrated 8.4° more knee valgus at IGC, 7.6° greater peak knee-valgus angle, and about 2.5 times more frontal-plane knee moment than uninjured females. By comparison, the high INI EA group displayed 6.6° more knee valgus at IGC, 14.4° greater peak knee-valgus angle, and about 2.1 times more frontal-plane knee moment than the low INI EA group. Whereas we clearly are limited in drawing any conclusions regarding injury outcome, the high INI EA group displayed frontal-plane knee biomechanics that are

Table 3. Frontal-Plane Initial Impact-Phase Energy-Absorption Group Comparisons for Biomechanical Variables

Variable	Initial Impact-Phase Energy-Absorption Group	Mean ± SD	95% Confidence Interval	$F_{2,79}$	P	η^{2a}
Frontal-plane knee angle at initial ground contact, °	High[b]	−10.34 ± 7.81	−13.43, −7.25	5.782	.005	0.128
	Moderate	−6.38 ± 7.69	−9.36, −3.40			
	Low	−3.73 ± 5.89	−6.06, −1.40			
Frontal-plane knee angle at peak knee-varus moment,°	High[b,c]	−18.88 ± 8.02	−22.05, −15.71	12.947	<.001	0.247
	Moderate	−11.50 ± 9.84	−15.31, −7.68			
	Low	−7.30 ± 7.29	−10.18, −4.41			
Peak knee-valgus angle, °	High[b,c]	−25.41 ± 8.66	−28.83, −21.98	19.874	<.001	0.503
	Moderate	−14.75 ± 10.31	−18.75, −10.75			
	Low	−11.04 ± 6.69	−13.68, −8.39			
Frontal-plane hip angle at peak knee-varus moment, °	High[b]	3.36 ± 7.24	0.50, 6.23	4.890	.01	0.110
	Moderate	−0.45 ± 5.65	−2.64, 1.74			
	Low	−1.77 ± 5.82	−4.07, 0.53			
Peak hip-adduction angle, °	High[b]	6.25 ± 7.74	3.19, 9.32	4.529	.01	0.103
	Moderate	1.90 ± 6.88	−0.77, 4.57			
	Low	0.76 ± 6.59	−1.85, 3.37			
Peak vertical ground reaction force, × body weight^{-1}	High	2.97 ± 0.67	2.71, 3.24	0.424	.66	0.011
	Moderate	2.96 ± 0.95	2.60, 3.33			
	Low	2.80 ± 0.73	2.51, 3.09			
Peak posterior ground reaction force, × body weight^{-1}	High[c]	0.91 ± 0.27	0.80, 1.02	4.030	.02	0.093
	Moderate	0.75 ± 0.23	0.66, 0.84			
	Low	0.76 ± 0.20	0.68, 0.84			
Peak knee-varus moment, × (body weight × height)$^{-1}$	High[b,c]	0.12 ± 0.05	0.10, 0.14	16.978	<.001	0.301
	Moderate	0.08 ± 0.04	0.06, 0.09			
	Low	0.06 ± 0.03	0.05, 0.07			

[a] Indicates effect size.
[b] Indicates high–initial impact-phase energy-absorption group was different from low–initial impact-phase energy-absorption group ($P < .05$).
[c] Indicates high–initial impact-phase energy-absorption group was different from moderate–initial impact-phase energy-absorption group ($P < .05$).

Table 4. Simple Bivariate Correlations Between Frontal- and Sagittal-Plane Initial Impact-Phase Energy Absorption During the Double-Legged Jump-Landing Task

	Frontal-Plane Initial Impact-Phase Energy Absorption							
	Total		Hip		Knee		Ankle	
Sagittal-Plane Initial Impact-Phase Energy Absorption	*r*	*P*	*r*	*P*	*r*	*P*	*r*	*P*
Total	−0.015	.89	0.139	.21	−0.054	.63	0.010	.93
Hip	−0.095	.40	−0.117	.30	−0.096	.39	0.050	.65
Knee	0.002	.99	0.301	.006[a]	0.025	.82	−0.151	.18
Ankle	0.046	.68	−0.141	.21	0.019	.86	0.224	.04[a]

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

sufficiently different from those of the low INI EA group to potentially result in greater frontal-plane knee loading. Accordingly, we propose that landing strategies with greater total frontal-plane INI EA are likely to cause greater ACL loading.

Despite the greater risk of noncontact ACL injury in females[2] and the association between sagittal-plane INI EA and ACL loading mechanisms, we did not identify a relationship between sex and sagittal-plane INI EA group in part I of this investigation.[14] In contrast, we identified an association between sex and frontal-plane INI EA group in this study ($\chi^2 = 4.909$, $P = .03$), finding women were 3.6 times more likely than men to be in the high INI EA group. This suggests that whereas men and women have an equal likelihood of landing with greater sagittal-plane INI EA and, therefore, an equal likelihood of experiencing greater ACL loading due to sagittal-plane mechanisms, women are more likely to absorb greater energy in the frontal plane during initial impact, so they are more likely to experience greater ACL loading caused by frontal-plane mechanisms. Moreover, given that multiplanar knee loading has been shown to result in greater ACL strain than pure sagittal-plane or frontal-plane loading,[11,33] we suggest that this increased likelihood of greater frontal-plane INI EA coupled with a similar chance of landing with greater sagittal-plane INI EA in women may contribute to their increased risk of ACL injuries. This notion is supported by the results of our post hoc analyses in which we compared the 10 (2 men, 8 women) participants assigned to both high–sagittal-plane[14] and high–frontal-plane and the 9 (6 men, 3 women) participants assigned to both low–sagittal-plane[14] and low–frontal-plane INI EA groups. Although we could not identify an association between sex and high-high versus low-low INI EA group allocation because of small cell frequencies, we did find that women were 8 times more likely than men to demonstrate both high–sagittal-plane and high–frontal-plane INI EA. In addition, the high-high INI EA group demonstrated greater knee valgus at IGC, peak knee-valgus angle, peak posterior GRF, peak knee-extension moment, peak anterior tibial shear force, and peak knee-varus moment than the low-low INI EA group. We suggest that identifying individuals who have greater magnitudes of INI EA in both the sagittal and frontal planes during landing may be a means of accurately discriminating individuals who display high-risk landing biomechanics in multiple planes without assessing the numerous discrete variables currently used for this purpose. Whereas this type of EA analysis still requires motion-capture methods similar to those currently used, we believe that using a single variable for each plane of motion that represents the biomechanical profiles of interest (ie, INI EA) is advantageous because it would simplify the analysis of high-risk landing biomechanics that must be done by comparing multiple individual variables. In addition, these results are clinically important because they suggest that facilitating a decrease in the magnitude of energy absorbed in the sagittal and frontal planes during initial impact in individuals who exhibit the largest magnitudes of combined multiplanar INI EA may be an appropriate intervention to reduce ACL injury risk. Energy absorption is influenced by factors that affect either joint angular velocities or net joint moments during landing[24,34,35]; therefore, INI EA might be changed by targeting not one but potentially several different specific modifiable factors, such as muscular strength and activation, joint positioning, and the magnitude of joint motion during landing.

Finally, the lack of a consistent association between frontal- and sagittal-plane INI EA agreed with our hypothesis and suggested that sagittal- and frontal-plane INI EA are generally independent (Table 4). Apart from relatively weak associations between sagittal-plane knee and frontal-plane hip ($r = 0.301$, $P = .006$) and between sagittal- and frontal-plane ankle ($r = 0.224$, $P = .04$) EA, we identified no other relationships between the magnitudes of sagittal- and frontal-plane INI EA. Whereas the results of our study confirm our previous findings,[14] they conflict with research in which investigators[16] have postulated that greater sagittal-plane EA would limit frontal-plane EA and thus frontal-plane knee loading. Though it appears counterintuitive that the magnitudes of sagittal- and frontal-plane INI EA are independent, all individuals do not have to absorb a fixed magnitude of energy during such a limited portion (100 milliseconds) of landing. The total energy of the system during these landings was relatively standardized among participants; however, in addition to energy being absorbed via eccentric contraction and accounted for in the EA calculation, it may be transformed into translational and rotational kinetic energy and into potential energy in each segment of the body and in each plane during landing.[28,36] Therefore, the magnitude of energy that needs to be absorbed during this critical time varies and depends on the motion of the individual segments. As a result, the magnitude of sagittal-plane INI EA does not necessarily influence the magnitude of frontal-plane INI EA and likely results in the lack of association confirmed by our investigation.

Limitations

Our study had limitations. First, the time interval used to identify peak kinematic angles was different from the time interval during which EA was calculated and the peak kinetic variables identified. We identified peak values for the kinetic variables during the initial 100 milliseconds of landing to align with both the EA calculation and the time when ACL injury most likely occurs.[20–22] Whereas this time interval differs from the one Hewett et al[12] used to identify peak frontal-plane knee moment as a predictor of ACL injury, we believe we can directly compare our results given the similarity of the landing tasks used and given that even when identified using the time from initial contact to peak knee flexion, the peak frontal-plane moment usually occurs within the first 100 milliseconds of landing (53 ± 37 milliseconds). However, peak kinematic angles do not always occur during the INI of landing (knee valgus = 108 ± 46 milliseconds, hip adduction = 111 ± 63 milliseconds). Therefore, ensuring that we identified the peak kinematic variables during a period that was consistent with previous prospective investigations was important to allow for accurate comparisons among studies. In addition, we do not believe the use of different time intervals for identifying peak kinematics and calculating EA confounds our results because the magnitude of frontal-plane INI EA clearly influences peak frontal-plane kinematics regardless of whether the peak kinematic angles occur slightly more than 100 milliseconds after ground contact.

Second, we studied healthy participants who were generally active but who may not have been regularly involved in activities requiring cutting, pivoting, and decelerating. Therefore, our results may not be generalizable to individuals who regularly perform these tasks and exhibit the greatest injury risk. However, given that the generally active population sustains ACL injuries, albeit at reduced rates, we believe that the results from this active population are clinically applicable.

CONCLUSIONS

Although ACL strain is greater under a combination of anterior shear force and frontal-plane knee moment than the isolated application of these components,[11,33] considerable disagreement persists about whether sagittal-plane[37,38] or frontal-plane[9] loading is more responsible for ACL injury. In part I of this investigation,[14] we demonstrated that greater INI EA in the sagittal plane indicated a biomechanical landing profile with greater peak internal knee-extension moment, anterior tibial shear force, and posterior GRF, which may result in greater ACL loading due to sagittal-plane mechanisms. Furthermore, we did not identify an association between sex and sagittal-plane INI EA group, signifying that a similar likelihood exists for men and women to land using this deleterious sagittal-plane strategy. In part II of this study, we reported that greater frontal-plane INI EA indicated frontal-plane landing biomechanics that potentially can increase ACL loading due to purely frontal-plane mechanisms and that women were 3.6 times more likely than men to exhibit higher frontal-plane INI EA during landing. In addition, we identified only 2 relatively weak relationships between the magnitudes of sagittal- and frontal-plane INI EA, indicating that these values generally are independent of one another. Given these findings, we hypothesize that individuals who consistently absorb a higher magnitude of energy in both the sagittal and frontal planes immediately after ground contact would be at highest risk of noncontact ACL injury because they would experience greater combined sagittal- and frontal-plane ACL loading. Considering that inherent variability exists in the magnitude of ACL loading during different movements, these individuals who normally operate at higher ACL-loading levels probably are more likely to experience suprathreshold loading events that actually result in ACL injury. However, future prospective investigation is necessary to test this hypothesis.

In addition, we speculate that the risk of ACL injury in females is increased possibly because they are more likely than males to land with higher frontal-plane INI EA but just as likely to land with higher sagittal-plane INI EA, which increases the likelihood for females to experience greater combined sagittal- and frontal-plane ACL loading. As such, we suggest that identifying biomechanical factors contributing to greater sagittal- and frontal-plane INI EA is paramount and might assist in the design of more efficacious ACL injury-prevention programs. Given that the magnitude of EA during landing is influenced by factors that affect either joint moments or joint angular velocities,[24,34,35] we suggest that changing modifiable variables, such as muscle strength, muscle activation, initial contact joint positions, and the magnitude of joint motion during landing, may successfully alter INI EA and potentially reduce ACL injury risk.

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REFERENCES

1. Dugan SA. Sports-related knee injuries in female athletes: what gives? *Am J Phys Med Rehabil*. 2005;84(2):122–130.
2. Griffin LY, Agel J, Albohm MJ, et al. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *J Am Acad Orthop Surg*. 2000;8(3):141–150.
3. Chappell JD, Yu B, Kirkendall DT, Garrett WE. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med*. 2002;30(2):261–267.
4. Ford KR, Myer GD, Hewett TE. Valgus knee motion during landing in high school female and male basketball players. *Med Sci Sports Exerc*. 2003;35(10):1745–1750.
5. Lephart SM, Ferris CM, Riemann BL, Myers JB, Fu FH. Gender differences in strength and lower extremity kinematics during landing. *Clin Orthop Relat Res*. 2002;401:162–169.
6. Pappas E, Hagins M, Sheikhzadeh A, Nordin M, Rose D. Biomechanical differences between unilateral and bilateral landings from a jump: gender differences. *Clin J Sport Med*. 2007;17(4):263–268.
7. Pollard CD, Sigward SM, Powers CM. Gender differences in hip joint kinematics and kinetics during side-step cutting maneuver. *Clin J Sport Med*. 2007;17(1):38–42.
8. Russell KA, Palmieri RM, Zinder SM, Ingersoll CD. Sex differences in valgus knee angle during a single-leg drop jump. *J Athl Train*. 2006;41(2):166–171.
9. McLean SG, Huang X, Su A, Van Den Bogert AJ. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. *Clin Biomech (Bristol, Avon)*. 2004;19(8):828–838.
10. Chaudhari AM, Andriacchi TP. The mechanical consequences of dynamic frontal plane limb alignment for non-contact ACL injury. *J Biomech*. 2006;39(2):330–338.

11. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res*. 1995;13(6):930–935.
12. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med*. 2005;33(4):492–501.
13. Griffin LY, Albohm MJ, Arendt EA, et al. Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005. *Am J Sports Med*. 2006; 34(9):1512–1532.
14. Norcross MF, Lewek MD, Padua DA, Shultz SJ, Weinhold PS, Blackburn JT. Lower extremity energy absorption and biomechanics during landing, part I: sagittal-plane energy absorption analyses. *J Athl Train*. 2013;48(6):748–756.
15. Yeow CH, Lee PVS, Goh JCH. Effect of landing height on frontal plane kinematics, kinetics and energy dissipation at lower extremity joints. *J Biomech*. 2009;42(12):1967–1973.
16. Pollard CD, Sigward SM, Powers CM. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clin Biomech (Bristol, Avon)*. 2010;25(2):142–146.
17. Lloyd DG, Buchanan TS. Strategies of muscular support of varus and valgus isometric loads at the human knee. *J Biomech*. 2001;34(10): 1257–1267.
18. Lloyd DG, Buchanan TS, Besier TF. Neuromuscular biomechanical modeling to understand knee ligament loading. *Med Sci Sports Exerc*. 2005;37(11):1939–1947.
19. Norcross MF, Blackburn JT, Georger BM, Padua DA. The association between lower extremity energy absorption and biomechanical factors related to anterior cruciate ligament injury. *Clin Biomech (Bristol, Avon)*. 2010;25(10):1031–1036.
20. Cerulli G, Benoit DL, Lamontagne M, Caraffa A, Liti A. In vivo anterior cruciate ligament strain behaviour during a rapid deceleration movement: case report. *Knee Surg Sports Traumatol Arthrosc*. 2003;11(5):307–311.
21. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. The relationship between quadriceps muscle force, knee flexion, and anterior cruciate ligament strain in an in vitro simulated jump landing. *Am J Sports Med*. 2006;34(2):269–274.
22. Koga H, Nakamae A, Shima Y, et al. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med*. 2010;38(11):2218–2225.
23. Decker MJ, Torry MR, Wyland DJ, Sterett WI, Steadman RJ. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clin Biomech (Bristol, Avon)*. 2003;18(7): 662–669.
24. Kulas AS, Schmitz RJ, Shultz SJ, Watson MA, Perrin DH. Energy absorption as a predictor of leg impedance in highly trained females. *J Appl Biomech*. 2006;22(3):177–185.
25. McNitt-Gray JL. Kinetics of the lower extremities during drop landings from three heights. *J Biomech*. 1993;26(9):1037–1046.
26. Schmitz RJ, Kulas AS, Perrin DH, Riemann BL, Shultz SJ. Sex differences in lower extremity biomechanics during single leg landings. *Clin Biomech (Bristol, Avon)*. 2007;22(6):681–688.
27. DeVita P, Janshen L, Rider P, Solnik S, Hortobágyi T. Muscle work is biased toward energy generation over dissipation in non-level running. *J Biomech*. 2008;41(16):3354–3359.
28. Decker MJ, Torry MR, Noonan TJ, Riviere A, Sterett WI. Landing adaptations after ACL reconstruction. *Med Sci Sports Exerc*. 2002; 34(9):1408–1413.
29. Winter DA. *Biomechanics and Motor Control of Human Movement*. 3rd ed. Hoboken, NJ: John Wiley & Sons; 2005:118–145.
30. Zhang SN, Bates BT, Dufek JS. Contributions of lower extremity joints to energy dissipation during landings. *Med Sci Sports Exerc*. 2000;32(4):812–819.
31. Kulas AS, Schmitz RJ, Shultz SJ, Watson MA, Perrin DH. Energy absorption as a predictor of leg impedance in highly trained females. *J Appl Biomech*. 2006;22(3):177–185.
32. Salci Y, Kentel BB, Heycan C, Akin S, Korkusuz F. Comparison of landing maneuvers between male and female college volleyball players. *Clin Biomech (Bristol, Avon)*. 2004;19(6):622–628.
33. McNair PJ, Prapavessis H. Normative data of vertical ground reaction forces during landing from a jump. *J Sci Med Sport*. 1999; 2(1):86–88.
34. Berns GS, Hull ML, Patterson HA. Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. *J Orthop Res*. 1992;10(2):167–176.
35. Mizrahi J, Susak Z. Analysis of parameters affecting impact force attenuation during landing in human vertical free fall. *Eng Med*. 1982;11(3):141–147.
36. Yeow CH, Lee PV, Goh JC. Sagittal knee joint kinematics and energetics in response to different landing heights and techniques. *Knee*. 2010;17(2):127–131.
37. Zatsiorsky VM. *Kinetics of Human Motion*. Champaign, IL: Human Kinetics; 2002:465–487.
38. Lin CF, Gross M, Ji C, et al. A stochastic biomechanical model for risk and risk factors of non-contact anterior cruciate ligament injuries. *J Biomech*. 2009;42(4):418–423.
39. Yu B, Garrett WE. Mechanisms of non-contact ACL injuries. *Br J Sports Med*. 2007;41(1 suppl):i47–i51.

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