Lower Extremity Energy Absorption and Biomechanics During Landing, Part I: Sagittal-Plane Energy Absorption Analyses

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Abstract:

Context: Eccentric muscle actions of the lower extremity absorb kinetic energy during landing. Greater total sagittal-plane energy absorption (EA) during the initial impact phase (INI) of landing has been associated with landing biomechanics considered high risk for anterior cruciate ligament (ACL) injury. We do not know whether groups with different INI EA magnitudes exhibit meaningful differences in ACL-related landing biomechanics and whether INI EA might be useful to identify ACL injury-risk potential.

Objective: To compare biomechanical factors associated with noncontact ACL injury among sagittal-plane INI EA groups and to determine whether an association exists between sex and sagittal-plane INI EA group assignment to evaluate the face validity of using sagittal-plane INI EA to identify ACL injury risk.

Design: Descriptive laboratory study.

Setting: Research laboratory.

Patients or Other Participants: A total of 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 individuals volunteered.

Intervention(s): We assessed landing biomechanics using an electromagnetic motion-capture system and force plate during a double-legged jump-landing task.

Main Outcome Measure(s): Total INI EA was used to group participants into high, moderate, and low tertiles. Sagittal- and frontal-plane knee kinematics; peak vertical and posterior ground reaction forces (GRFs); anterior tibial shear force; and internal hip extension, knee extension, and knee varus moments were identified and compared across groups using 1-way analyses of variance. We used a χ2 analysis to compare male and female representation in the high and low groups.

Results: The high group exhibited greater knee-extension moment and posterior GRFs than both the moderate ($P < .05$) and low ($P < .05$) groups and greater anterior tibial shear force than the low group ($P < .05$). No other group differences were noted. Women were not represented more than men in the high group ($\chi^2 = 1.20$, $P = .27$).

Conclusions: Greater sagittal-plane INI EA likely indicates greater ACL loading, but it does not appear to influence frontal-plane biomechanics related to ACL injury. Women were not more likely than men to demonstrate greater INI EA, suggesting that quantification of sagittal-plane INI EA alone is not sufficient to infer ACL injury-risk potential.

Keywords: anterior cruciate ligament | landing biomechanics | kinetics | kinematics

Article:

Key Points

- Landing with greater sagittal-plane energy absorption (EA) during the 100 milliseconds immediately after ground contact (INI) resulted in sagittal-plane knee kinetics and impact forces that likely increased anterior cruciate ligament loading.

- Sex was not associated with the magnitude of sagittal-plane INI EA during landing.

- Sagittal-plane INI EA did not appear to influence frontal-plane knee biomechanics believed to be associated with greater anterior cruciate ligament injury risk.

- Total sagittal-plane INI EA may serve as a surrogate for the multiple, discrete variables commonly used to evaluate high-risk sagittal-plane landing strategies, and elevated sagittal-plane INI EA may be a useful marker of poor sagittal-plane landing biomechanics.

Noncontact mechanisms account for 70% to 80% of all anterior cruciate ligament (ACL) injuries[1,2] and occur most commonly in dynamic activities involving rapid deceleration, cutting, and landing.[3,4] During landing, internal hip-, knee-, and ankle-extension (plantar-flexion) moments must be produced via eccentric muscle contractions to control joint motion and to absorb the kinetic energy of the body.[5] This energy absorption (EA) by the lower extremity musculature can be calculated using energetic analyses in which kinematic (joint angular velocity) and kinetic (net joint moment) data are combined to quantify the energy at each joint that is responsible for producing the observed movement.[6]

Whereas conventional biomechanical analyses used in ACL injury research identify specific joint kinematic and kinetic variables independently and at discrete time points, energetic analyses quantify these data across the landing period. Further, individual contributions of the hip, knee, and ankle to the total lower extremity EA can be combined, providing insight into the coordinated actions of these joints.[7–9] By coupling the kinematics and kinetics of multiple joints, a more comprehensive description of the complex multisegmental mechanics that occur during landing and in proposed ACL injury mechanisms can be generated.[10]

Researchers[5] have suggested that greater EA by the neuromuscular system over the entire landing period during drop landings reduces the loading of passive tissues, such as the ACL. Specifically, greater total lower extremity EA in the sagittal plane has been associated with smaller vertical ground reaction forces (GRFs) and greater knee-flexion displacements during landing.[11,12] However, these results typically have been observed by researchers who have artificially manipulated landing conditions. Devita and Skelly[5] and Zhang et al[12] observed greater EA and smaller peak impact forces in "soft" landings than "stiff" landings when participants were instructed to alter the magnitude of knee-flexion displacement during drop landings. To date, few researchers have directly evaluated the influence of sagittal-plane EA during natural (ie, nonmanipulated) landing conditions on peak impact forces and other biomechanical factors specifically related to noncontact ACL injury.

Norcross et al[13] reported the first direct associations between EA and biomechanical factors related to noncontact ACL injury in individuals using their preferred landing styles. Their exploratory analysis involving just 27 participants suggested that both the magnitude and the timing of EA during landing influence these biomechanical factors. Specifically, greater total lower extremity EA in the sagittal plane during the initial impact phase (INI) of landing (ie, 100 milliseconds immediately after initial ground contact [IGC]) was associated with greater peak vertical GRF, anterior tibial shear force, and internal hip-extension moment—factors generally considered to be unfavorable with respect to ACL injury risk.[14,15] However, greater total EA during the terminal phase of landing (ie, 100 milliseconds after IGC to the minimal vertical position of the whole-body center of mass) was associated with smaller peak values of these same biomechanical factors.[13] Therefore, they suggested that analyzing EA during the INI of landing may serve to quantify combined multijoint movement strategies that could result in greater ACL injury risk.[13] However, this previous investigation had 2 principal limitations. First, whereas they identified important relationships among lower extremity EA and key ACL-related biomechanical factors, we do not know whether groups performing different amounts of sagittal-plane INI EA during landing demonstrate meaningful differences on these ACL-related biomechanical factors. Second, whereas quantification of sagittal-plane INI EA appears to accurately synthesize an overall sagittal-plane biomechanical landing profile, we do not know whether quantification of sagittal-plane INI EA might be a useful mechanism to identify individuals at greater risk for noncontact ACL injury. Females are well documented to display a greater likelihood than males for sustaining noncontact ACL injuries[1,16] despite males sustaining a greater absolute number of ACL injuries.[17–19] Therefore, greater sagittal-plane INI EA potentially could be more prominent in females and serve as a more effective and discrete means of prospectively identifying high-risk athletes than sex.

Therefore, the purpose of our study was to address these aforementioned limitations by (1) determining whether meaningful differences exist among high–, moderate–, and low–sagittal-plane INI EA groups in various biomechanical factors that are associated with noncontact ACL injury and (2) evaluating the face validity of using sagittal-plane INI EA to identify ACL injury risk by determining whether an association exists between sex and sagittal-plane INI EA group assignment. We hypothesized that individuals in the high INI EA group would display less favorable values across all biomechanical variables than those in the moderate and low INI EA groups and that a greater proportion of women than men would be represented in the high versus low INI EA group.

METHODS

Participants

A total of 82 individuals (41 men, 41 women; age = 21.0 ± 2.4 years, height = 1.74 ± 0.10 m, mass = 70.3 ± 16.1 kg) volunteered. All participants were required to be *physically active*, which was defined as participating in at least 30 minutes of physical activity 3 times per week, and generally healthy with no history of ACL injury, neurologic disorder, lower extremity surgery, or lower extremity injury within the 6 months preceding data collection. All participants provided written informed consent, and the study was approved by the University of North Carolina at Chapel Hill Biomedical Institutional Review Board.

Procedures

The height and mass of each participant were recorded before data collection and used to generate a biomechanical model and normalize the dependent variables. Lower extremity and trunk kinematics were assessed using an electromagnetic motion-capture system (Motion Star; Ascension Technology Corp, Burlington, VT). The 6-degrees-of-freedom electromagnetic tracking sensors were positioned over the third metatarsal of the foot, anteromedial aspect of the shank, and lateral thigh of the dominant lower extremity, which was defined as the lower extremity used to kick a ball for maximal distance; sacrum; and C7 spinous process of the trunk. These sensors were placed over areas of minimal muscle mass and secured with prewrap and athletic tape to reduce motion artifact. We established global and segment axis systems and designated the positive x-axis as forward/anteriorly, the positive y-axis as leftward/medially, and the positive z-axis as upward/superiorly. A segment-linkage model of the dominant lower extremity, pelvis, and thorax was created using The MotionMonitor motion analysis software (Innovative Sports Training, Inc, Chicago, IL) by digitizing the ankle-, knee-, and hip-joint centers and the T12 spinous process (Figure). Knee- and ankle- -joint centers were defined as the midpoints of the digitized medial and lateral malleoli and medial and lateral femoral condyles, respectively. The hip-joint center was predicted using external landmarks on the pelvis as described by Bell et al.[20]

To perform double-legged jump landings, participants stood atop a 30-cm-high box that was set at a distance equal to 50% of their height away from the edge of a nonconductive force plate (type 4060-NC; Bertec Corporation, Columbus, OH) with an axis system aligned with the global axis system. We instructed participants to jump down and forward toward the force plate, contact the ground with both feet at the same time with the dominant foot near the center of the force plate and the nondominant foot positioned next to the force plate, and immediately jump up for maximal height using both legs. In addition, they were encouraged to use whatever technique they desired to jump as high as possible and were not provided feedback on their chosen landing techniques during the testing. Participants performed 3 practice trials and 5 successful testing trials with 30 seconds of rest between trials to minimize the potential effects of fatigue. Trials were deemed successful if participants jumped from the box and landed with both feet at the same time, completely contacted the force plate with only the dominant foot, and performed the landing task and subsequent maximal jump in a fluid motion.

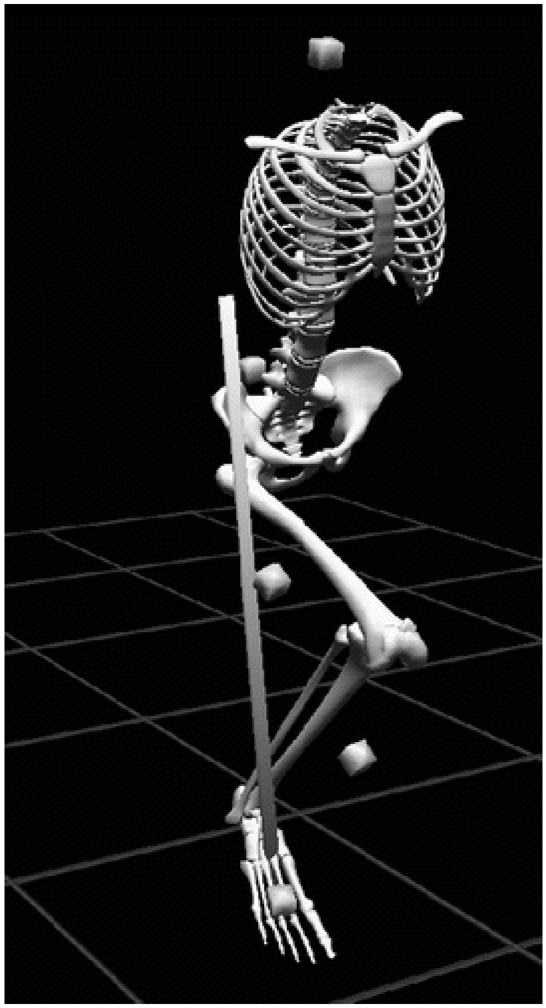

Figure.

Representative biomechanical model of a single participant studied for kinematic and kinetic analysis. Note the location of electromagnetic sensors depicted as white cubes around the model skeleton.

Data Sampling and Reduction

Kinematic and kinetic data were sampled at 120 and 1200 Hz, respectively, using The MotionMonitor motion-analysis software. Raw kinematic data were low-pass filtered using a fourth-order, zero-phase-lag Butterworth filter with a cutoff frequency of 10 Hz,[21] time synchronized with the kinetic data, and resampled at 1200 Hz. We calculated joint angular positions based on a right-hand convention using Euler angles in a YX′Z″ rotation sequence and calculated instantaneous joint angular velocities as the first derivative of angular position. We defined motion about the hip as the thigh relative to the pelvis, about the knee as the shank relative to the thigh, and about the ankle as the foot relative to the shank. Kinetic data were low-pass filtered at 60 Hz with a fourth-order, zero-phase-lag Butterworth filter[22] and combined with kinematic and anthropometric data to calculate the net internal joint moments of force at the hip, knee, and ankle and the net internal force on the shank at the knee joint using an inverse dynamics solution.[23]

We used custom computer software (LabVIEW; National Instruments Corporation, Austin, TX) to multiply sagittal-plane joint angular velocities (ω) and net joint moments (M) to generate hip-, knee-, and ankle-joint power curves (P) for each landing trial ($P = M \times \omega$). The negative portion of the joint power curves (ie, when net joint moment and joint angular velocity are in opposite directions and indicate eccentric muscle action) were integrated to calculate negative mechanical joint work[11,21,24,25] during the INI of landing (the 100 milliseconds immediately after IGC [vertical GRF > 10 N]).[21,26] Finally, total negative lower extremity joint work was calculated by summing the negative joint works calculated at the hip, knee, and ankle.[11,12,25] This value represents the total sagittal-plane lower extremity EA because negative joint work indicates EA by the muscle-tendon unit.[6,24] We calculated total EA to quantify the combined actions of the lower extremity joints during landing. Further, we focused on the INI of landing for 2 reasons: (1) previous results identified a temporal relationship between EA and high-risk landing biomechanics in which greater INI EA and lesser terminal-phase EA were considered unfavorable[13] and (2) peak ACL strain and injury likely occur during this period.[27–29]

The same custom software was used to identify sagittal- and frontal-plane knee angles at IGC and peak values for vertical and posterior GRF; anterior tibial shear force; internal hip-extension, knee-extension, and knee-varus moments; and knee-flexion and -valgus angles. In addition, sagittal- and frontal-plane knee angles were identified at the instants of peak knee-extension moment and anterior tibial shear force and of peak knee-varus moment, respectively.

Peak kinetics proposed to contribute to ACL loading usually occur during the initial 100 milliseconds of landing and, therefore, were identified during the INI of landing. However, peak angular values for knee flexion and knee valgus generally occur more than 100 milliseconds after IGC. Therefore, peak kinematic values were identified during the total landing phase (IGC to the minimal vertical position of the whole-body center of mass) to identify the true peak values during landing and allow for comparison with previous studies in which investigators[12,14,22] used the total landing phase when reporting peak knee-flexion and -valgus angles. We normalized GRFs and segmental forces to the participant's body weight (BW) ($\times BW^{-1}$) and net joint moments 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 × Ht). All dependent variables were averaged across the 5 jump-landing trials of each participant before statistical analysis.

Statistical Analysis

Total INI EA data were arranged into tertiles to create 3 distinct INI EA groups: high, moderate, and low. Static comparisons across INI EA groups for each biomechanical factor were made using separate 1-way analysis of variance models. For significant models, we performed post hoc Tukey honestly significant difference testing to identify group differences for these dependent variables. We constructed a 2 × 2 contingency table using sex and INI EA group (high and low) as categorical variables and used a Pearson χ^2 test of association to determine whether a greater proportion of women than men was represented in the high or low INI EA group. All analyses were conducted using commercially available software (SPSS 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 INI EA groups are displayed in Tables 1 and 2. The INI EA group assignment by tertile successfully created 3 groups with different sagittal-plane EA during INI ($F_{2,79}$ = 133.093, P < .001; Table 1). For biomechanical variables related to ACL injury, we observed differences among groups for peak knee-extension moment ($F_{2,79}$ = 10.537, P < .001), anterior tibial shear force ($F_{2,79}$ = 5.813, P = .004), and posterior GRF ($F_{2,79}$ = 10.582, P < .001; Table 3). Post hoc testing revealed that the high group landed with greater peak knee-extension moment than the moderate (P = .007) and low (P < .001) groups. However, we did not detect a difference in knee-extension moment between the moderate and low INI EA groups (P = .34; Table 3). The high group demonstrated greater peak anterior tibial shear force than the low group (P = .004); however, we did not note differences between the high and moderate groups (P = .07) or the moderate and low groups (P = .50; Table 3). Peak posterior GRF also was greater in the high than moderate (P = .001) or low (P < .001) groups, but the posterior GRFs of the moderate and low groups were not different (P = .84; Table 3). We did not note INI EA group differences for any other biomechanical variable of interest (P > .05; Tables 3 and 4). We did not find an association between sex and high or low INI EA group assignment (χ^2 = 1.20, P = .27;Table 1).

Table 1. Sagittal-Plane Initial Impact-Phase Energy-Absorption Frequency Counts by Sex and Group Descriptives (Mean ± SD)

	Initial Impact-Phase Energy-Absorption Group		
Characteristic	High	Moderate	Low
Participants			
Women	14	17	10
Men	13	11	17
Total	27	28	27
Age, y			
Women	20.07 ± 1.49	21.00 ± 2.78	21.40 ± 3.72
Men	20.69 ± 1.60	20.91 ± 1.38	21.47 ± 2.83
Total	20.37 ± 1.55	20.96 ± 2.30	21.44 ± 3.12
Height, m			
Women	1.66 ± 0.05	1.65 ± 0.05	1.69 ± 0.08
Men	1.80 ± 0.06	1.80 ± 0.07	1.83 ± 0.06
Total	1.72 ± 0.09	1.71 ± 0.10	1.78 ± 0.10
Mass, kg			
Women	57.63 ± 5.80	62.87 ± 10.43	63.29 ± 10.10
Men	74.04 ± 7.48	77.76 ± 14.66	84.78 ± 20.88
Total	65.53 ± 10.61	68.72 ± 14.10	76.82 ± 20.39
Total initial impact-phase energy absorption, % body weight × height			
Women[a,b]	17.22 ± 1.53	13.30 ± 0.81	10.87 ± 1.47
Men[a,b]	16.74 ± 2.19	13.49 ± 0.75	10.28 ± 1.62
Total[a,b]	16.99 ± 1.85	13.37 ± 0.78	10.50 ± 1.57
95% confidence interval for total initial impact-phase energy absorption			
Women	16.34, 18.10	12.89, 13.71	9.82, 11.93
Men	15.41, 18.06	12.99, 13.99	9.45, 11.11
Total	16.25, 17.72	13.07, 13.68	9.88, 11.12

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

Table 2. Group Descriptives for Magnitude of Individual Joint Initial Impact-Phase Energy Absorption and Joint Contribution to Total Initial Impact (Mean ± SD)

Joint Initial Impact-Phase Energy Absorption	Initial Impact-Phase Energy-Absorption Group		
	High	Moderate	Low
Hip			
% Body weight × height	2.64 ± 1.57	2.08 ± 1.22	2.08 ± 1.19
% Total initial impact-phase energy absorption	15.5 ± 9.2	15.5 ± 9.1	20.0 ± 11.6
Knee			
% Body weight × height	11.03 ± 2.70	8.90 ± 1.94	7.02 ± 1.71
% Total initial impact-phase energy absorption	64.9 ± 14.0	66.62 ± 14.4	67.0 ± 13.7
Ankle			
% Body weight × height	3.31 ± 1.55	2.40 ± 1.60	1.40 ± 1.21
% Total initial impact-phase energy absorption	19.6 ± 9.3	17.9 ± 11.9	13.1 ± 10.9

DISCUSSION

Our primary finding was that individuals absorbing a greater magnitude of energy in the sagittal plane during the INI of landing used a movement strategy that may result in greater ACL loading. The high INI EA group exhibited greater peak knee-extension moment, anterior tibial shear force, and posterior GRF than the low INI EA group without differences in sagittal-plane knee kinematics.

The greater knee extension moment and anterior tibial shear force demonstrated by the high INI EA group supported our hypotheses and have been proposed[30,31] in previous research as important contributors to ACL loading. During landing, the lower extremity joints must use internally generated extension moments to resist rapid joint flexion introduced by impact forces.[5,32] At the knee, the internal-extension moment is generated by a quadriceps contraction, which has been identified as the primary contributor to anterior tibial shear force.[31] In vitro[33–35] and in vivo[36,37] experiments have demonstrated that quadriceps contraction between 0° and 30° of knee flexion and the ensuing anterior tibial shear force strain the ACL. Further, DeMorat et al[30] successfully induced ACL injury in 6 of 11 cadaver specimens by applying an isolated quadriceps force. Therefore, our findings indicate that movement strategies with greater sagittal-plane EA during the 100 milliseconds immediately after ground contact result in greater knee-extension moment and anterior tibial shear force, thereby resulting in greater quadriceps forces that potentially can induce greater ACL loading.

The resultant strain on the ACL due to a standardized quadriceps contraction may be influenced by the sagittal-plane position of the knee. Nunley et al[38] reported that the angle between the patellar tendon and tibial shaft decreases as the knee progresses into flexion, resulting in a smaller proportion of the quadriceps force being directed anteriorly relative to the tibia. The *elevation angle* of the ACL,[39–41] defined as the angle between the longitudinal axis of the ACL and the tibial plateau,[40] also decreases with knee flexion, so the ACL is oriented less vertically, a smaller proportion of ACL loading is shear rather than tensile, and less ACL strain occurs with a given anterior shear force.[31] Therefore, under the same quadriceps loading conditions, positioning the knee in more flexion would result in less ACL strain. Accordingly, the high INI EA group possibly exhibited greater knee-extension moment and anterior tibial shear force but in a more-flexed knee position, thereby mediating the effects of the greater quadriceps force and experiencing resultant ACL loading that was comparable with that of the other groups. However, we found no differences in knee-flexion angle at IGC, at peak knee-extension moment, or at peak anterior tibial shear force (Table 4). The lack of kinematic differences between INI EA groups suggests that knee-flexion angle alone does not determine INI EA and that other factors, such as the magnitude of quadriceps activation during landing, may have a greater influence on the magnitude of INI EA. Further, we believe that the greater sagittal-plane knee kinetics observed in the high INI EA group, in concert with the same knee kinematics as the moderate and low INI EA groups, indicate greater ACL loading due to sagittal-plane mechanisms.

Table 3. Sagittal-Plane Initial Impact-Phase Energy-Absorption Group Comparisons for Peak Kinetic Variables

Variable	Initial Impact-Phase Energy-Absorption Group	Mean ± SD	95% Confidence Interval	$F_{2,79}$	P	η^{2a}
Vertical ground reaction force, × body weight^{-1}	High	2.94 ± 0.67	2.68, 3.20	0.096	.91	0.002
	Moderate	2.86 ± 0.89	2.51, 3.20			
	Low	2.94 ± 0.82	2.62, 3.26			
Posterior ground reaction force, × body weight^{-1}	High[b,c]	0.96 ± 0.27	0.86, 1.07	10.582	<.001	0.211
	Moderate	0.74 ± 0.20	0.67, 0.82			
	Low	0.71 ± 0.18	0.64, 0.78			
Anterior tibial shear force, × body weight^{-1}	High[b]	0.97 ± 0.17	0.90, 1.04	5.813	.004	0.128
	Moderate	0.85 ± 0.22	0.76, 0.94			
	Low	0.79 ± 0.20	0.71, 0.87			
Hip-extension moment, × (body weight × height)$^{-1}$	High	0.29 ± 0.13	0.24, 0.35	0.667	.52	0.017
	Moderate	0.28 ± 0.13	0.23, 0.33			
	Low	0.32 ± 0.13	0.27, 0.37			
Knee-extension moment, × (body weight × height)$^{-1}$	High[b,c]	0.21 ± 0.05	0.19, 0.23	10.537	<.001	0.211
	Moderate	0.17 ± 0.05	0.15, 0.19			
	Low	0.16 ± 0.03	0.15, 0.17			
Knee-varus moment, × (body weight × height)$^{-1}$	High	0.08 ± 0.05	0.06, 0.11	0.035	.97	0.001
	Moderate	0.08 ± 0.03	0.07, 0.10			
	Low	0.09 ± 0.05	0.07, 0.11			

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

Table 4. Sagittal-Plane Initial Impact-Phase Energy-Absorption Group Comparisons for Kinematic Variables

Variable	Initial Impact-Phase Energy-Absorption Group	Mean ± SD	95% Confidence Interval	$F_{2,79}$	P	η^{2a}
Sagittal-plane knee angle at initial ground contact, °	High	22.73 ± 6.96	19.98, 25.49	0.015	.99	<0.001
	Moderate	23.11 ± 8.92	19.65, 26.57			
	Low	23.03 ± 9.60	19.23, 26.83			
Frontal-plane knee angle at initial ground contact, °	High	−7.73 ± 8.17	−8.85, −1.85	0.760	.99	0.019
	Moderate	−7.34 ± 5.53	−9.49, −5.20			
	Low	−6.81 ± 7.60	−10.96, −4.50			
Sagittal-plane knee angle at peak knee-extension moment, °	High	47.93 ± 18.33	40.67, 55.18	1.483	.23	0.036
	Moderate	55.56 ± 13.87	50.18, 60.94			
	Low	52.86 ± 17.44	45.96, 59.76			
Sagittal-plane knee angle at peak anterior tibial shear force, °	High	52.35 ± 20.66	44.17, 60.52	2.003	.14	0.048
	Moderate	61.35 ± 14.55	55.71, 67.00			
	Low	57.61 ± 14.33	51.94, 63.28			
Frontal-plane knee angle at peak knee-varus moment, °	High	−10.17 ± 10.73	−14.42, −5.92	1.270	.29	0.031
	Moderate	−13.34 ± 7.49	−16.24, −10.43			
	Low	−14.10 ± 10.36	−18.19, −10.00			
Peak knee-flexion angle, °	High	93.82 ± 14.16	73.91, 99.42	1.143	.32	0.028
	Moderate	91.15 ± 14.75	71.72, 96.87			
	Low	87.74 ± 15.47	61.96, 93.86			
Peak knee-valgus angle, °	High	−14.37 ± 11.15	−18.78, −9.96	1.310	.28	0.032
	Moderate	−18.12 ± 8.86	−21.56, −14.69			
	Low	−18.57 ± 11.35	−23.06, −14.08			

[a] Indicates effect size.

Our results for peak impact forces during landing also were surprising. Whereas the high INI EA group displayed greater peak posterior GRF than both the moderate and low groups, we found no differences among groups for peak vertical GRF (Table 3). This result is in contrast to a previous exploratory investigation in which researchers[13] found an association between peak vertical GRF and total sagittal-plane INI EA; however, only 19.5% of the variance in vertical GRF was explained by sagittal-plane INI EA. Both the posterior and vertical components of the GRF can induce a flexion moment relative to the knee that must be resisted by quadriceps contraction and can increase ACL loading.[31] In a prospective investigation, Hewett et al[14] found that peak vertical GRFs were 20% greater at baseline in females who sustained ACL injuries than in uninjured females. However, accurately comparing the magnitudes of the vertical GRF in our study with this investigation is difficult because the authors did not normalize their measured vertical GRF to account for the participants' mass. In addition, the existing literature in which researchers have compared females and males (ie, representing higher and lower ACL injury risk) on vertical GRF is equivocal. Schmitz et al[11] and Salci et al[42] reported greater peak vertical GRFs in females, whereas McNair and Prapavessis[43] and Decker et al[21] did not observe sex differences in peak vertical GRF during landing. Limited evidence suggests that the posterior component of the GRF is just as important as, if not more important than, the vertical component in explaining knee-joint loading. Yu et al[44] reported anterior tibial shear force and knee-extension moment were associated with peak posterior and vertical GRF. Furthermore, they found that peak posterior GRF occurred at the same time as peak anterior tibial shear force and knee-extension moment and explained 72% and 74%, respectively, of the variance in these same variables compared with only 26% and 32%, respectively, of the variance for vertical GRF.[44] Whereas the variances of knee-extension moment (42% versus 17%) and anterior tibial shear force (34% versus 30%) explained by peak posterior and vertical GRF, respectively, in our investigation were not as large when compared with the findings of Yu et al,[44] these secondary analyses further support the notion that the posterior component of the GRF is important in explaining knee-joint loading. Collectively, these results imply that an increase in posterior GRF likely results in greater ACL loading during landing. As such, the greater peak posterior GRF exhibited by the high INI EA group, even without a concomitant group difference in peak vertical GRF, lends further support to the notion that a movement strategy involving greater lower extremity INI EA increases resultant ACL loading due to sagittal-plane mechanisms.

A lack of INI EA group differences in peak hip-extension moment (Table 3) was unexpected given a previous investigation in which the researchers[13] found a relationship between total INI EA and peak hip-extension moment. However, as with peak vertical GRF, only 18% of the variance in peak hip-extension moment was explained by total sagittal-plane INI EA.[13] In addition, we included a sample size 3 times greater than that in the previous study, thereby decreasing the influence of more extreme values that may have affected the hip-extension moment and vertical GRF findings reported previously.

Our results confirmed exploratory findings that indicated a lack of relationship between total sagittal-plane INI EA and frontal-plane knee kinematics and kinetics.[13] We noted no group differences for knee-valgus angle at IGC, peak knee-valgus angle, peak internal knee-varus moment, or knee-valgus angle at peak knee-varus moment (Tables 3 and 4). These frontal-plane variables are important because knee-valgus angle at initial contact has been reported to differ between individuals who went on to sustain noncontact ACL injury and those who did not, and peak knee-valgus angle and external knee-valgus moment are prospective predictors of noncontact ACL injury.[14] In addition, at knee-flexion angles greater than 10°, an externally applied valgus moment combined with anterior shear force results in greater ACL loading than that produced by anterior shear force alone.[45] Accordingly, limiting frontal-plane knee-valgus motion and moments has been advocated to decrease ACL injury risk.[46] Pollard et al[47] reported that individuals exhibiting more combined peak hip and knee flexion during landing displayed more sagittal-plane hip and knee EA and less peak knee-valgus angle and average internal knee-varus moment. These authors speculated that greater use of sagittal-plane EA may have reduced the magnitude of EA in the frontal plane and thereby influenced frontal-plane knee biomechanics. Pollard et al[47]calculated sagittal-plane EA from IGC to peak knee flexion, which is different from the time frame we used (ie, 100 milliseconds after IGC). We believe that the initial impact period may be a more useful time epoch to evaluate given the previously reported relationships between greater INI EA and high-risk landing biomechanics[13] and the observation that ACL injury likely occurs during this period.[27–29] The failure of the high INI EA group to exhibit a less favorable frontal-plane biomechanical profile than the other groups suggests the magnitude of sagittal-plane INI EA does not influence frontal-plane biomechanics and associated ACL loading caused by frontal-plane mechanisms as had been suggested.[47] We suggest that investigators should more closely examine interplanar INI EA relationships and the direct influence of frontal-plane INI EA on frontal-plane biomechanics.

Finally, our investigation showed isolated quantification of total sagittal-plane INI EA to infer noncontact ACL injury risk most likely is unfounded. Contrary to our hypothesis, we did not find an association between INI EA group assignment (high versus low) and sex. Given the overwhelming evidence indicating the greater risk of ACL injury in females,[1,16] we expected a greater proportion of women would be assigned to the high INI EA group if this measure indeed indicated injury risk. However, this result also indicated that men and women have an equal likelihood of using a landing strategy that results in greater ACL loading due to sagittal-plane mechanisms (ie, high INI EA). As such, we propose that associations between sex and frontal- and transverse-plane landing biomechanics are more likely to contribute to the increased risk of ACL injury in females.

Limitations

Our study had limitations. First, we focused on healthy individuals aged 18 to 30 years who were generally active but did not necessarily participate regularly in activities requiring sudden deceleration, cutting, and pivoting. Therefore, we do not know whether these results can be generalized to individuals who regularly perform these types of movements and are at the greatest risk of ACL injury. However, given that ACL injuries sometimes do occur in individuals not participating in high-risk activities, we believe that our results in this active population are important.

Second, our investigation used different time intervals during which peak kinetic and kinematic variables were identified. We limited the identification of peak kinetics to the INI of landing to align with the EA calculation and to capture the peak values during the period when ACL injury likely occurs[27–29] because these factors are proposed to directly contribute to loading that can result in ACL failure. However, given that peak knee-flexion (mean = 208 ± 54 milliseconds) and valgus (mean = 108 ± 46 milliseconds) angles do not occur consistently during the INI of landing, we used the total landing interval to identify the true peak angular values during landing and also to remain consistent with previous prospective research in which investigators[14] identified these peak kinematic variables over the total landing period.

CONCLUSIONS

Our results provide important information for understanding the ways in which INI EA during landing affects ACL loading. Landing with greater sagittal-plane EA during the 100 milliseconds immediately after ground contact resulted in sagittal-plane knee kinetics and impact forces that likely increased ACL loading. However, we did not identify an association between sex and sagittal-plane INI EA during landing, indicating that the magnitude of sagittal-plane INI EA during landing alone does not explain the greater risk of ACL injury in females. In addition, sagittal-plane INI EA does not appear to influence frontal-plane knee biomechanics thought to be associated with greater ACL injury risk. Nonetheless, these results indicated that total sagittal-plane INI EA may serve as a surrogate for the multiple, discrete variables currently used to evaluate high-risk sagittal-plane landing strategies and that high INI EA may be a useful marker of poor sagittal-plane landing biomechanics. As such, the development or validation of a clinically applicable instrument that can be used to detect high INI EA landing strategies and serve as an objective clinical tool for identifying poor sagittal-plane landing biomechanics is warranted. In addition, researchers should determine which biomechanical factors predict sagittal-plane INI EA and whether sagittal-plane INI EA may be modified via an intervention program to decrease ACL loading attributable to sagittal-plane mechanisms. Further, the relationships among frontal-plane INI EA and frontal-plane biomechanics and sagittal-plane INI EA during landing should be investigated more closely.

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