Changing sagittal plane body position during single-leg landings influences the risk of non-contact anterior cruciate ligament injury

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Abstract:

Purpose. To examine the effects of different sagittal plane body positions during single-leg landings on biomechanics and muscle activation parameters associated with risk for anterior cruciate ligament (ACL) injury.
Methods. Twenty participants performed single-leg drop landings onto a force plate using the following landing styles: self-selected, leaning forward (LFL) and upright (URL). Lower extremity and trunk 3D biomechanics and lower extremity muscle activities were recorded using motion analysis and surface electromyography, respectively. Differences in landing styles were examined using 2-way Repeated-measures ANOVAs (sex × landing conditions) followed by Bonferroni pairwise comparisons.
Results. Participants demonstrated greater peak vertical ground reaction force, greater peak knee extensor moment, lesser plantar flexion, lesser or no hip extensor moments, and lesser medial and lateral gastrocnemius and lateral quadriceps muscle activations during URL than during LFL. These modifications of lower extremity biomechanics across landing conditions were similar between men and women.
Conclusions. Leaning forward while landing appears to protect the ACL by increasing the shock absorption capacity and knee flexion angles and decreasing anterior shear force due to the knee joint compression force and quadriceps muscle activation. Conversely, landing upright appears to be ACL harmful by increasing the post-impact force of landing and quadriceps muscle activity while decreasing knee flexion angles, all of which lead to a greater tibial anterior shear force and ACL loading. ACL injury prevention programmes should include exercise regimens to improve sagittal plane body position control during landing motions.

Keywords: Anterior cruciate ligament injury | Injury prevention | Electromyography | Sagittal plane biomechanics | Lower extremity

Article:

INTRODUCTION

An estimated 200,000 anterior cruciate ligament (ACL) injuries occur annually in the United States alone [21]; more than 70 % of these are non-contact in nature [5]. Non-contact ACL injuries often occur during sudden deceleration activities, such as landing (e.g. single-leg landing), immediately after touchdown with the knee at a shallow flexion angle [23, 29]. Video analyses and prospective studies have shown that leaning the trunk backward [5, 27] or laterally while decelerating [14] and inability to control frontal plane trunk position [36, 37] are often associated with ACL injuries. Therefore, it is important to assess how trunk control affects lower extremity biomechanical parameters that may be associated with ACL injuries.

Sagittal plane body positions have been shown to affect lower extremity biomechanics and the risk of ACL injury [3, 4, 9, 10, 16, 30]. Blackburn and Padua [3, 4] examined the differences in lower extremity biomechanics (joint angles and GRF) and quadriceps muscle activation produced by two sagittal plane trunk flexion positions (preferred vs. flexed) during drop landings and reported that flexing the trunk increased the maximum knee flexion angles and reduced quadriceps muscle activation during the entire descending phase of the landing, producing a smaller peak vertical GRF compared with that of the preferred trunk position. Shimokochi et al. [30] reported that leaning the whole-body forward could simultaneously reduce the knee extensor moment and increase the ankle and hip extensor moments during single-leg landings, suggesting that leaning forward may be ACL protective. The effects of sagittal plane foot-landing techniques on the biomechanical risk of ACL injury were examined by Cortes et al. [9, 10]. They found that how the foot was used could significantly affect the risk of ACL injury during landing [10], sidestep cutting and pivoting tasks [9]. The relative amounts of ACL loading were compared between soft and stiff landing techniques using a computer simulation model [16]. Soft landings with hip and knee flexion angles larger than those observed during stiff landings were found to result in less ACL loading than did stiff landings [16].

Although the abovementioned studies provide preliminary information on how the position of the body's centre of mass influences the risk of ACL injury, they have certain limitations. For example, Blackburn and Padua [3, 4] examined only average vastus lateralis muscle activation and maximum knee flexion angles during the entire descending phase of landing. However, as non-contact ACL injuries occur immediately after foot contact [29], it is important to examine muscle activation and lower extremity kinematics and kinetics with the knee at a shallow flexion angle. Moreover, lower extremity joint moments or hamstring muscle activation, which are important for knee stability [2, 32], were also not examined. Additionally, although Shimokochi et al. [30] discussed the relationships among lower extremity sagittal plane joint moments during landing, they analysed only simple correlations among lower extremity joint moments and not sagittal plane body position, muscle activation, joint kinematics or GRF. Other studies have examined the effects of different landing techniques on lower extremity biomechanics and ACL injury risk [9, 10, 16] but did not investigate the effect of changing the whole-body sagittal plane position on lower extremity biomechanics and knee loading. Furthermore, few studies have examined the biomechanical characteristics of landing in an upright position, which has been suggested to be risky for the ACL [5, 27]. Overall, there have been some limited reports but few comprehensive examinations of the actual effects of different sagittal plane body positions on lower extremity kinetics, kinematics and muscle activations during landing and their effects on ACL injury risk factors.

Therefore, the purpose of current study was to examine the effects of different sagittal plane body positions on lower extremity biomechanics and muscle activation during single-leg landings. It was hypothesised that the most ACL-protective biomechanical parameters would be associated with leaning forward landings (LFL) and the most harmful with upright landings (URL) and that all variables examined would be intermediate between LFL and URL during self-selected landings (SSL). The following specific hypotheses were tested: (1) at the point of foot contact during landing, URL would result in less plantar flexion with more upright posture than would LFL, (2) URL would result in a greater peak vertical GRF and reduced ankle plantar flexor moment in less time than that required for LFL, which would show the opposite trend, (3) URL would result in a greater peak knee extensor moment with reduced knee flexion angles as well as hip extensor and ankle plantar flexor moments at the peak knee extensor moment than would LFL, and (4) URL would result in greater quadriceps and reduced gastrocnemius and hamstring muscle activation than would LFL during the early post-foot-contact phase of landing.

MATERIALS AND METHODS

Twenty recreationally active participants (10 men and 10 women, 23.4 ± 3.6 years of age, 171.0 ± 9.4 cm tall, and 73.3 ± 12.7 kg in weight) were recruited. All participants signed the university-approved Institutional Review Board consent form before participating. None reported any history of knee ligament injury or lower extremity pain at the time of participation.

Test protocol

The participants performed all landings wearing their own running shoes. Prior to actual data collection, the participants were asked to perform double-leg counter-movement jumps and to land on a single leg to determine which leg was most comfortable for single-leg landing (the dominant leg), which was then used for all subsequent tests [12].

The participants were then familiarised with a single-leg drop landing from a 30-cm (women) or 45-cm (men) box. These heights were chosen during pilot testing as the highest drop heights from which female and male participants were comfortable manipulating their landing styles; the largest possible effects of different landing styles could thus be observed while ensuring that the participants could still land successfully.

The participants were instructed to stand on the box on their dominant legs with both arms across their chests and then to drop off of the box and land on the centre of the force plate using their dominant legs. The edges of the boxes were aligned to a line that was 10 cm away from the edge of the force plate. They were instructed not to jump up or lower their bodies when dropping off of the box and to remain balanced on the dominant leg for at least 2 s after landing. Prior to actual data collection, the participants practiced each landing style until they were comfortable with all.

Surface electromyography (sEMG)

Two bipolar Ag/Ag–Cl surface electrodes (centre-to-centre distance 20 mm; Blue Sensor N-00-S, Ambu Products, Ølstykke, Denmark) were attached over the medial and lateral gastrocnemius (MG and LG, respectively), hamstring (MH and LH, respectively) and quadriceps (MQ and LQ, respectively) muscles on areas of skin prepared using previously documented procedures [25]. Manual muscle tests were performed to confirm crosstalk elimination. All sEMG data were recorded at 1,000 Hz using a 16-channel Myopac telemetric system (amplification 1 mV/V, frequency bandwidth 10–1,000 Hz, CMRR 90 dB min at 60 Hz, input resistance 1 MΩ, and internal sampling rate 8 kHz; Run Technologies, Mission Viejo, CA, USA).

Biomechanical instrumentation and digitisation procedures

A non-conductive force plate (Type 4060, Bertec Corporation, Columbus, OH, USA) and 3D electromagnetic tracking system (Ascension Star Hardware, Ascension Technology, Burlington, VT; Motion Monitor Software, Innovative Sports Training, Chicago, IL, USA) were used to obtain kinetic and kinematic data at 1,000 Hz and 120 Hz, respectively. The 3D electromagnetic tracking system was found to have a 2–5 mm (position) and 0.3°~0.5° (orientation) measurement error in static condition at a 1.5 m distance. All tasks were performed within this distance from the transmitter. The examiners attached the motion sensors directly to the skin above the dorsum of the foot, tibial shaft, iliotibial band on the lateral aspect of the thigh, and sacrum using double-sided tape, athletic underwrap and white tape. For the foot dorsum, the participants removed their socks, attached the sensors to the dorsal surfaces of their feet, replaced their shoes and fastened their shoelaces to secure the sensors. A sensor was also attached to the posterior aspect of the thorax using a manufacturer-made bracket (Fig. 1). Thereafter, participants' dominant legs, sacra and trunks were digitised and their joint centres estimated as previously reported [31]. Internal joint moments were calculated using a standard Newtonian inverse-dynamics approach [35].

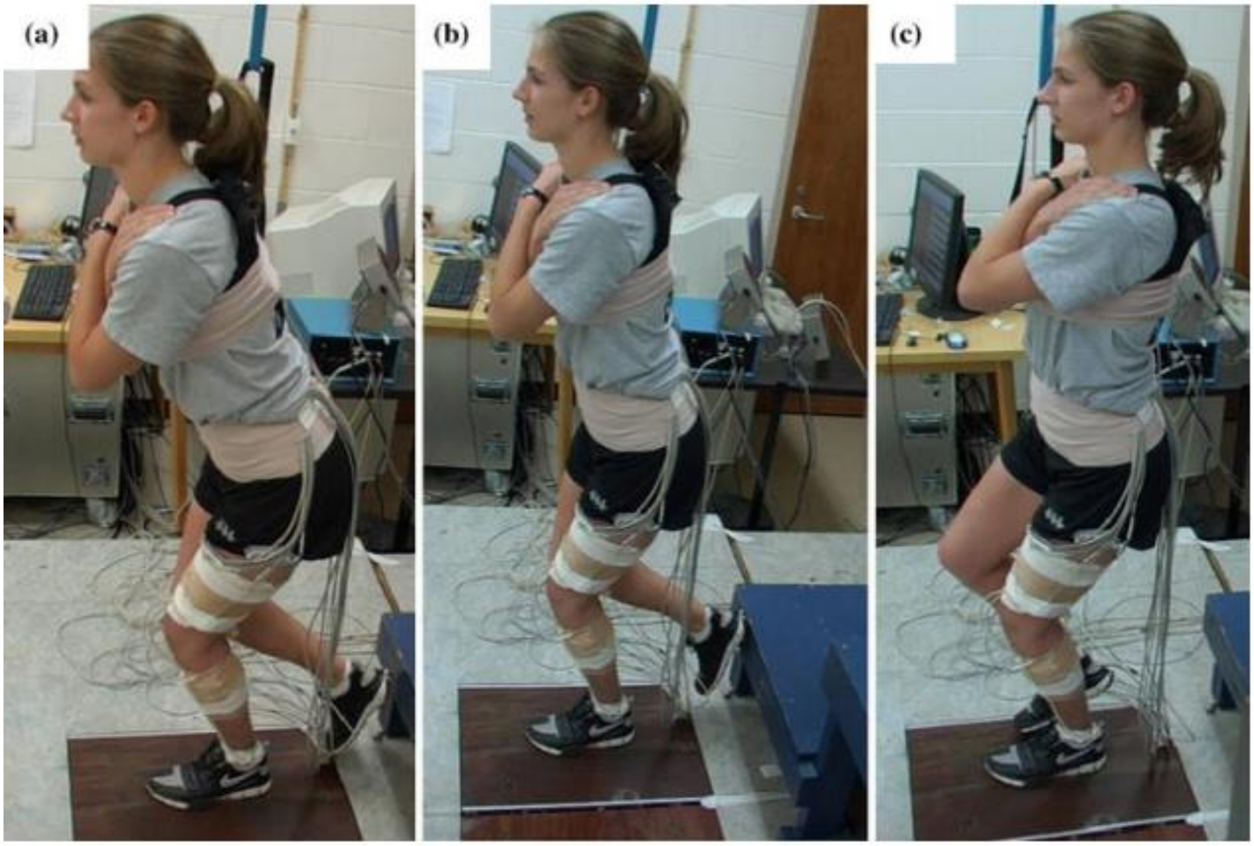

Fig. 1

Exemplar landing styles (a) leaning forward landing (LFL), (b) self-selected landing (SSL) and (c) upright landing (URL)

Landings

After digitisation, participants performed 5 landings in each landing style (Fig. 1): (1) SSL, (2) LFL and (3) URL. SSL was performed first, followed by LFL and URL in a counterbalanced order (Fig. 1).

All participants received the same instructions: for SSL, ‘just land the way you want’, for LFL, ‘lean forward when you land and land on your toes’ and for URL, ‘land with your body as upright as possible and land on your heel’. A trial was discarded and repeated if the participant (1) jumped up or slowly lowered his or her body when dropping off the box, (2) could not remain standing using only the dominant leg for 2 s after landing or (3) did not cross his or her arms across the chest throughout the landing task.

Data processing and reduction

The kinetic and kinematic data were processed with a low-pass fourth-order zero-lag digital Butterworth filter at 40 and 12 Hz, respectively. The position and orientation data were synchronised with the GRF data. The joint angles were calculated as previously reported by

Shultz and Schmitz [31]. To compare trunk inclination across the landing conditions, sagittal plane angles of the thorax and sacrum were calculated relative to the vertical axis of the global coordinate system and were defined as the thoracic and sacral inclination angles, respectively. The participants may have modified their sacral and thoracic inclination angles differently while landing due to different vertebral articulations in the lumbar spine. Therefore, both the sacral and thoracic inclination angles were examined to determine how the participants modified their trunk positions during single-leg landings.

The thoracic and sacral inclination angles as well as the ankle dorsi/plantar flexion angles at initial foot contact were extracted to ensure that the participants successfully modified their landing styles across the 3 conditions. To test hypotheses of current study, the peak vertical GRFs and peak plantar flexor moments as well as the times to these peaks were extracted. The peak knee extensor moment, sagittal plane hip and ankle moments, and knee flexion angles at the instant of peak knee extensor moment were subsequently extracted. The peak vertical GRFs were normalised by the participant's body weight (N) and the joint moments by both the weight (N) and height (m) of each participant (all kinetic variables thus have no unit) [8]. In order to increase the reproducibility of the biomechanical data, the values of each variable from 5 successful single-leg landings of each style were averaged to determine representative values for each participant [33].

The sEMG data were bandpass-filtered (10–350 Hz), using a fourth-order, zero-lag Butterworth filter and a centred Root Mean Square algorithm with a 25-ms time constant. To test the third hypothesis, the mean sEMG amplitudes for each muscle during the first 100 ms after touchdown were ensemble-averaged across 5 trials of each landing style to determine the representative values for each participant.

Statistical analyses

All variables were examined using a single multivariate repeated-measures analysis of variance (RMMANOVA) to reduce the chance of type II error. If the main effect for landing style was significant in the RMMANOVA, separate univariate 2-way repeated-measures ANOVAs (sex × landing conditions) were examined for each variable to examine the possible sex-specific neuromechanical adaptations to different landing styles. Pairwise comparisons with Bonferroni corrections were then conducted to determine where the significant differences occurred. If the sex-by-landing interaction was significant, a separate repeated-measure ANOVA (RMANOVA) followed by Bonferroni pairwise comparisons was conducted for each sex. The independent variable was the landing style, while the dependent variables were the kinetic, kinematic and sEMG variables. The alpha level was set at 0.05.

RESULTS

The RMMANOVA results indicated a significant main effect with a large effect size (Wilks' $\lambda = 0.022$, $F = 7.43$, $P < 0.01$, Partial $\eta^2 = 0.852$, Power = 1). Table 1 describes the means, standard deviations and results of the RMANOVA with post hoc comparisons for the first hypothesis. The thoracic inclination angles were the most upright and the most anteriorly inclined during the URL and LFL, respectively, for all participants. The thoracic inclination angles also showed a significant sex-by-landing interaction, with the thoracic inclination angles being significantly different among all conditions in men while difference only found between SSL and URL and between LFL and URL in women. No significant interaction was observed for the sacral inclination angle. Although the sacral inclination angle at foot contact was the most anteriorly inclined in LFL, no differences in sacral anterior inclination angles were found between URL and SSL. No significant sex-by-landing interaction was found in the plantar flexion angles at initial contact, with the least plantar flexion occurring in URL and the most in LFL.

Table 1 Descriptive statistics and results of 2-way and 1-way repeated-measures ANOVAs and post hoc comparisons of sagittal plane thoracic and sacral inclination angles and ankle flexion angles (°) at the time of foot contact

	LFL	SSL	URL	Post hoc comparisons
Thoracic inclination angle				
Males**	−5.0 ± 7.5	4.0 ± 5.4	14.1 ± 7.2	LFL < SSL‡‡, LFL < URL‡‡, SSL < URL‡‡
Females**	3.5 ± 8.5	8.0 ± 6.4	14.0 ± 7.1	LFL < URL‡‡, SSL < URL‡
All**, †	−1.0 ± 9.0	6.0 ± 6.1	14.1 ± 6.9	LFL < SSL‡‡, LFL < URL‡‡, SSL < URL‡‡
Sacral inclination angle				
Males	6.9 ± 8.9	9.6 ± 6.0	11.2 ± 6.5	
Females	4.2 ± 8.0	7.6 ± 6.5	8.2 ± 8.3	
All**	5.5 ± 8.4	8.6 ± 6.2	9.7 ± 7.4	LFL < SSL‡‡, LFL < URL‡‡
Initial ankle plantar flexion angle				
Males	−39.7 ± 6.0	−37.8 ± 6.0	−23.0 ± 18.6	
Females	−41.2 ± 6.9	−38.4 ± 4.7	−6.4 ± 21.2	
All**	−40.4 ± 6.3	−37.6 ± 5.3	−14.7 ± 21.2	LFL < SSL‡‡, LFL < URL‡‡, SSL < URL‡‡

Positive directions indicate posterior inclination and dorsiflexion

LFL leaning forward landing, *SSL* self-selected landing, *URL* upright landing, *n.s.* main effect is not significant

* Significant main effect at a level of less than 0.05

** Significant main effect at a level of less than 0.01

† Significant sex-by-landing interaction at a level of less than 0.05

†† Significant sex-by-landing interaction at a level of less than 0.01

‡ Significantly different at a level of less than 0.05

‡‡ Significantly different at a level of less than 0.01

Table 2 shows the descriptive statistics and the results of RMANOVA with post hoc comparisons for the second hypothesis. No significant sex-by-landing interaction was found for any variable with peak GRF. A significant sex-by-landing interaction was found for time to peak GRF. In both sexes, however, the largest peak vertical GRF and the shortest time to peak GRF were observed during URL while the smallest peak vertical GRF and the longest time to peak GRF were observed during LFL. That is, although there was a significant sex-by-landing interaction in time to peak GRF, the trends in time to peak GRF among the landing conditions were the same between sexes, indicating that the significant interaction arose only from differences in probability levels between sexes.

Table 2 Descriptive statistics and results of 2-way and 1-way repeated-measures ANOVAs and post hoc comparisons of normalised peak vertical ground reaction force (GRF) and plantar flexor moment and times to these peaks (msec) across the 3 landing conditions

	LFL	SSL	URL	Post hoc comparisons
Peak vertical GRF				
Males	4.3 ± 0.5	5.1 ± 0.6	5.6 ± 0.7	
Females	3.5 ± 0.6	4.2 ± 0.6	5.0 ± 0.8	
All**	3.9 ± 0.7	4.6 ± 0.8	5.3 ± 0.8	LFL < SSL‡‡, LFL < URL‡‡, SSL < URL‡‡
Peak ankle plantar flexor moment				
Males$^{n.s.}$	−0.20 ± 0.02	−0.19 ± 0.03	−0.16 ± 0.06	
Females**	−0.20 ± 0.04	−0.19 ± 0.06	−0.09 ± 0.05	LFL < URL‡‡, SSL < URL‡
All**, †	−0.20 ± 0.03	−0.19 ± 0.05	−0.13 ± 0.07	LFL < URL‡‡, SSL < URL‡
Time to peak vertical GRF				
Males**	60 ± 9	49 ± 5	37 ± 12	LFL > SSL‡‡, LFL > URL‡‡, SSL > URL‡
Females**	73 ± 16	55 ± 9	32 ± 15	LFL > SSL‡‡, LFL > URL‡‡, SSL > URL‡‡
All**, ††	66 ± 14	52 ± 8	36 ± 14	LFL > SSL‡‡, LFL > URL‡‡, SSL > URL‡‡
Time to peak ankle plantar flexor moment				
Males	46 ± 8	45 ± 11	58 ± 30	
Females	48 ± 13	40 ± 6	66 ± 34	
All*	47 ± 11	42 ± 9	62 ± 31	SSL < URL‡

The positive direction indicates dorsiflexion

LFL leaning forward landing, *SSL* self-selected landing, *URL* upright landing, *n.s.* main effect is not significant

* Significant main effect at a level of less than 0.05

** Significant main effect at a level of less than 0.01

† Significant sex-by-landing interaction at a level of less than 0.05

†† Significant sex-by-landing interaction at a level of less than 0.01

‡ Significantly different at a level of less than 0.05

‡‡ Significantly different at a level of less than 0.01

A significant sex-by-landing interaction was found in the peak ankle plantar flexor moments. Separate RMANOVAs showed no significant differences in peak plantar flexor moment among landing conditions in men but a smaller peak plantar flexor moment in URL than in SSL or LFL in women. Time-to-peak plantar flexor moments were significantly shorter for URL than for LFL, with no significant sex-by-landing interaction.

Table 3 shows the descriptive statistics and results of RMANOVAs and post hoc comparisons for the third hypothesis. Significant main effects with no significant sex-by-landing interactions were found for all peak knee extensor-related variables. The peak knee extensor moment was 56 and 27 % greater during URL than during LFL and SSL, respectively. The peak knee extensor moment was also 18 % smaller during LFL than during SSL. On average, the hip and ankle moments at the time of peak knee extensor moment were directed towards flexion and dorsiflexion during URL but towards extension and plantar flexion during LFL and SSL. The knee flexion angle at the peak knee extensor moment was 10.8° and 8.6° greater in LFL than in URL and SSL, respectively.

Table 3 Descriptive statistics and results of 2-way repeated-measures ANOVAs of normalised peak knee extensor moment, hip extensor/flexor and ankle plantar/dorsiflexor moments, and knee flexion angles at peak knee extensor moment (°) across the 3 landing conditions

	LFL	SSL	URL	
Peak knee extensor moment				
Males	0.11 ± 0.02	0.11 ± 0.02	0.15 ± 0.04	
Females	0.08 ± 0.03	0.10 ± 0.03	0.13 ± 0.04	
All**	0.09 ± 0.03	0.11 ± 0.02	0.14 ± 0.04	LFL < SSL‡, LFL < URL‡‡, SSL < URL‡‡
Hip moment				
Males	−0.04 ± 0.10	0.03 ± 0.12	0.06 ± 0.13	
Females	−0.06 ± 0.07	−0.06 ± 0.05	0.01 ± 0.11	
All*	−0.05 ± 0.08†	−0.02 ± 0.10	0.04 ± 0.12	LFL < URL‡
Ankle moment				
Males	−0.07 ± 0.02	−0.04 ± 0.04	0.04 ± 0.08	
Females	−0.08 ± 0.04	−0.04 ± 0.05	0.02 ± 0.07	
All**	−0.08 ± 0.03	−0.04 ± 0.05	0.03 ± 0.07	LFL < SSL‡‡, LFL < URL‡‡, SSL < URL‡‡
Knee flexion angle at KEMpk				
Males	−37.1 ± 9.4	−26.5 ± 11.3	−24.3 ± 9.7	
Females	−37.9 ± 10.0	−31.5 ± 11.3	−29.0 ± 11.2	
All**	−37.5 ± 9.5	−28.9 ± 11.4	−26.7 ± 10.5	LFL > SSL‡‡, LFL > URL‡‡

Positive directions indicate hip flexion, knee extension, and ankle dorsiflexion

LFL leaning forward landing, *SSL* self-selected landing, *URL* upright landing, *n.s.* main effect is not significant

* Significant main effect at a level of less than 0.05

** Significant main effect at a level of less than 0.01

† Significant sex-by-landing interaction at a level of less than 0.05

†† Significant sex-by-landing interaction at a level of less than 0.01

‡ Significantly different at a level of less than 0.05

‡‡ Significantly different at a level of less than 0.01

Table 4 shows the descriptive statistics and results of RMANOVAs and post hoc comparisons for the fourth hypothesis. LQ muscle activation was 12 and 8 % greater in URL than in LFL and SSL, respectively, while LG and MG muscle activation were 17 and 21 % less, respectively, in URL than in LFL. Also, MG muscle activation in URL was 11 % less than in SSL. There were no significant differences in the other muscles across landing conditions. No significant sex-by-landing interactions were found for all muscle activations.

Table 4 Descriptive statistics and results of 2-way repeated-measures ANOVAs with post hoc comparisons of percent mean root-mean-square amplitudes relative to maximum voluntary isometric contractions during the 100 ms after foot contact across the 3 landing conditions (%)

	LFL	SSL	URL	Post hoc comparisons
Lateral gastrocnemius				
Males	49.1 ± 12.8	43.3 ± 13.7	40.2 ± 13.1	
Females	70.2 ± 25.3	63.9 ± 23.1	59.2 ± 21.9	
All**	59.7 ± 22.3	53.6 ± 21.3	49.7 ± 20.1	LFL > URL‡‡
Medial gastrocnemius				
Males	53.2 ± 17.8	44.5 ± 13.1	39.8 ± 17.4	
Females	72.2 ± 25.6	67.3 ± 25.4	59.7 ± 26.9	
All**	62.7 ± 23.6	55.9 ± 22.9	49.8 ± 24.3	LFL > URL‡‡, SSL > URL‡
Lateral quadriceps				
Males	44.8 ± 13.6	44.9 ± 12.7	46.7 ± 16.6	
Females	61.2 ± 22.0	64.9 ± 29.5	71.6 ± 28.5	
All*	53.0 ± 19.7	54.9 ± 24.4	59.2 ± 26.1	LFL < URL‡‡, SSL < URL‡
Medial quadriceps				
Males	69.9 ± 27.3	67.8 ± 22.6	68.4 ± 28.0	
Females	84.1 ± 27.3	96.0 ± 38.2	106.7 ± 61.9	
All[n.s.]	77.0 ± 27.6	81.9 ± 33.8	87.6 ± 50.7	
Lateral hamstrings				
Males	36.0 ± 11.9	36.1 ± 12.8	34.3 ± 10.7	
Females	43.4 ± 24.7	51.4 ± 36.4	55.8 ± 40.0	
All[n.s.]	39.7 ± 19.2	43.7 ± 27.7	45.1 ± 30.6	
Medial hamstrings				
Males	30.0 ± 14.6	32.4 ± 15.5	32.3 ± 13.0	
Females	37.8 ± 18.4	38.5 ± 17.6	44.3 ± 25.4	
All[n.s.]	33.9 ± 16.6	32.3 ± 13.0	38.3 ± 20.6	

LFL leaning forward landing, *SSL* self-selected landing, *URL* upright landing, *n.s.* main effect is not significant

* Significant main effect at a level of less than 0.05

** Significant main effect at a level of less than 0.01

† Significant sex-by-landing interaction at a level of less than 0.05

†† Significant sex-by-landing interaction at a level of less than 0.01

‡ Significantly different at a level of less than 0.05

‡‡ Significantly different at a level of less than 0.01

DISCUSSION

The most important finding of the current study was that altering the sagittal plane body position during landing clearly influences trunk and lower extremity biomechanics and lower extremity muscle activation and may potentially alter ACL injury risk. Specifically, URL increased the vertical GRF, peak knee extensor moment and quadriceps muscle activation while decreasing the knee flexion angle and producing less or no hip extensor moment (or even increasing hip flexor moment) at the time of peak knee extensor moment. Conversely, LFL decreased the vertical GRF, peak knee extensor moment and quadriceps muscle activation while increasing the knee flexion angle and hip extensor moment at the time of peak knee extensor moment. These trends were similar between men and women (i.e. no significant sex-by-landing interactions in these variables), indicating that LFL is more ACL protective and URL more ACL harmful for both sexes.

The results showed the largest vertical peak GRFs in URL and the smallest vertical peak GRFs in LFL. Therefore, the shock-attenuating strategy adopted during URL was less effective and ACL harmful while that used during LFL may be more ACL protective. Meyer and Haut [19, 20] demonstrated in cadaver models that tibial axial loading can rupture the ACL. The lateral tibia has a greater slope on its posterior aspect, with a slightly concave lateral facet, whereas the medial plateau has a greater slope on its anterior aspect, with a convex medial facet. Due to this geometrical joint anatomy, tibiofemoral compression produces a more anterior shift in the lateral than in the medial tibia [20], resulting overall in anterior tibial translation and tibial internal rotation loading that can strain or ultimately rupture the ACL [1, 19, 20, 29]. This theory is supported by an in vivo study by Cerulli et al. [7], who found that peak ACL strain coincided with peak vertical GRF during a single-leg forward-hop landing task. Applying this theory to current results, it was found that the average vertical peak GRF generated was 1.4-fold greater in URL than in LFL. As the GRF is the largest external force producing tibiofemoral joint compression during landing, the ACL may undergo the most strain during URL and the least during LFL.

As differences in the vertical peak GRF magnitude across landing conditions were observed, it was necessary to examine how modifications in shock-attenuating strategies differed by sex. The results showed significant sex-by-landing interactions in thoracic inclination angles at the times of foot contact and peak plantar flexor moment. Previous studies showed that modifying upper body position at initial foot contact during landing considerably influences lower extremity neuromuscular control during the subsequent descending phase [3, 4, 15]. Furthermore, reducing plantar flexor moment during landing was suggested to reduce the capacities of subsequent shock-attenuating modalities during landings [6, 26]. Therefore, comparing changes in these variables between sexes may provide some insight into how participants of each sex modified their shock-attenuating strategies across different landing conditions.

The sex-specific analyses indicated that while men successfully modified their thoracic inclination angles across all landing conditions, women showed significant thoracic inclination angle changes between some pairs but not between SSL and LFL. Although the sacral inclination angles did not differ between SSL and URL for both sexes, these results collectively suggest that men modified their sagittal plane trunk positions across different landing conditions more extensively than did women.

It was also found that while women demonstrated smaller peak plantar flexor moments in URL than in SSL or LFL, men did not have significant differences in this variable across landing conditions. One might suppose from these results that women modified their ankle plantar flexor muscle usage across landing conditions more than did men, but MG and LG had greater muscle activations in LFL than in URL for both sexes. Furthermore, participants demonstrated greater plantar flexion angles at foot contact in LFL than in URL and more flat-footed positions in URL. Collectively, it appeared that both sexes tended to use their plantar flexor muscles to attenuate GRF more during LFL than during URL, with greater plantar flexor muscle activations and larger ankle ROM (secondary to greater plantar flexion angle at foot contact); these adaptations led to better shock attenuation in LFL than in URL.

The contention that the shock absorption of the ankle joint is lessened in URL was further supported by secondary examination of the relative times to peak GRF and plantar flexor moments across the landing conditions. Specifically, during URL, the peak GRF (36 ± 14 ms) occurred before the peak plantar flexor moment (62 ± 31 ms), whereas the peak GRF (66 ± 14 ms) occurred after the peak plantar flexor moment (47 ± 11 ms) during LFL. This observation indicates that LFL allowed the participants sufficient time to produce the peak ankle plantar flexor moment before the peak GRF occurred, increasing the proportion of GRF absorption by the ankle. Conversely, during URL, the peak GRF occurred before the peak ankle plantar flexor moment. Collectively, these results support previous suggestions that shock-attenuating capacity is higher in LFL than in URL [30].

The results of current study showed that the greatest peak knee extensor moment occurred during URL and the smallest during LFL. Additionally, LQ muscle amplitudes were larger in URL than in LFL. These results collectively indicate that quadriceps muscles were more activated at the time when knee was less flexed in URL than in LFL. The hip extensor moments at the time of peak knee extensor moment were greater for LFL than for URL. As the hamstring muscles extend the hip, greater hip extensor moments were expected during LFL. However, the MH and LH amplitudes did not differ among the landing conditions. Several mechanisms may have contributed in part to this result. First, the co-activation patterns of the quadriceps and hamstring muscles could have differed between participants across the landing conditions. Second, as the hamstrings are 2-joint muscles, the absence of clear differences in sacral inclination across the landing conditions may have influenced hamstring muscle activation. Third, other hip extensors (e.g. the gluteus muscles) may also have modified hip extensor moment across the landing conditions.

Despite the consistency of hamstring muscle activation across the landing conditions, LFL resulted in larger knee flexion angles at peak knee extensor moment than did SSL or URL. A previous study [17] using cadaveric knees demonstrated that the amount of ACL loading decreases with increasing knee flexion angle; the loads due to a quadriceps force of 200 N at knee flexion angles of 30° and 60° were approximately 80 and 40 %, respectively, of that at a knee flexion angle of 15°. Furthermore, a progressive decrease in ACL tensile forces has also been noted due to the co-contraction forces in the hamstrings increasing with knee flexion (~30 % reduction at 15° and ~50 % reduction at 30° and 60° of knee flexion for an ACL tensile force produced by the quadriceps alone). Greater knee flexion angles observed during LFL in the current study thus indicate that the ACL strain is lower in LFL than in URL or SSL.

Although somewhat controversial [13, 18, 29], excessive quadriceps muscle forces without sufficient hamstring muscle co-contraction forces, especially at shallow knee flexion angles, have been reported to increase ACL loading to a degree that may eventually rupture the ACL [11, 29]. However, a recent cadaver study by Wall et al. [34] found that the compressive load required to rupture the ACL when the knee is at 15° of flexion is significantly reduced when combined with a large quadriceps force than without a quadriceps force. Sustaining a large GRF during landing has been thought to be harmful to the ACL as it increases knee joint compressive force [6, 20]. The risk of ACL injuries may be further increased if the larger GRF is associated with a GRF vector that passes through or in front of the centre of the knee joint [24, 28]. Thus, the larger GRF and greater quadriceps activations immediately after foot contact at shallower knee flexion angles observed in URL as compared with in LFL indicate that URL is more ACL harmful. Contrary to URL, LFL may be considered to be more ACL protective as it showed opposite tendencies than URL.

SSL gave less consistent results than did LFL and URL, likely because the participants were asked to land as preferred in SSL. Therefore, not all participants' SSL had neuromechanical patterns 'midway' between those of LFL and URL.

Some limitations of this study should be acknowledged. The current study did not examine actual ACL loading or muscle forces across the different landing conditions. Thus, interpretations about the amount of ACL loading are based on previously proposed theories of ACL strain and fundamental neuromechanical information [4, 11, 19, 20, 22, 29]. Future studies should also validate whether different landing conditions actually produce different ACL loading patterns.

CONCLUSIONS

The findings of the current study indicate that LFL increases the lower extremity shock-attenuating capacity and stabilises the knee by preventing excessive quadriceps contraction while maintaining hamstring muscle contraction and increasing knee flexion angles during the post-impact phase of landing. Conversely, URL decreases lower extremity shock-attenuating capacity and knee flexion angle while increasing quadriceps contraction. LFL may be more ACL protective. Individuals should therefore avoid URL as it may be ACL harmful. ACL injury prevention training programmes or rehabilitation programmes for ACL reconstructed athletes should also include teaching individuals to land with proper sagittal body positions that prevent excessive knee joint compressive forces and improve neuromuscular control to increase knee stability.

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