Gender Comparisons Between Unilateral and Bilateral Landings

Joshua T. Weinhandl, Mukta Joshi, and Kristian M. O’Connor

The increased number of women participating in sports has led to a higher knee injury rate in women compared with men. Among these injuries, those occurring to the ACL are commonly observed during landing maneuvers. The purpose of this study was to determine gender differences in landing strategies during unilateral and bilateral landings. Sixteen male and 17 female recreational athletes were recruited to perform unilateral and bilateral landings from a raised platform, scaled to match their individual jumping abilities. Three-dimensional kinematics and kinetics of the dominant leg were calculated during the landing phase and reported as initial ground contact angle, ranges of motion (ROM) and peak moments. Lower extremity energy absorption was also calculated for the duration of the landing phase. Results showed that gender differences were only observed in sagittal plane hip and knee ROM, potentially due to the use of a relative drop height versus the commonly used absolute drop height. Unilateral landings were characterized by significant differences in hip and knee kinematics that have been linked to increased injury risk and would best be classified as “stiff” landings. The ankle musculature was used more for impact absorption during unilateral landing, which required increased joint extension at touchdown and may increase injury risk during an unbalanced landing. In addition, there was only an 11% increase in total energy absorption during unilateral landings, suggesting that there was a substantial amount of passive energy transfer during unilateral landings.

Keywords: anterior cruciate ligament, injury, lower limb, knee, biomechanics

In recent years, studies have reported that females have a higher prevalence of noncontact anterior cruciate ligament (ACL) injuries compared with males during athletic competition (Arendt & Dick, 1995; Ferretti et al., 1992). Many have suggested that females perform high demand athletic maneuvers differently than males and in a manner that predisposes them to higher knee joint stress (Cowling & Steele, 2001; Decker et al., 2003; Huston et al., 2001; Kernozek et al., 2005; McLean et al., 2007; McLean et al., 2005; Wojtys et al., 2002). Kirkendall and Garrett (2000) reported ACL injuries occurring in basketball and soccer were most often noncontact in nature (64 of 72 injuries, 88%) and a result of a deceleration type of movement (landing from a jump was reported in 30 of the 72 injuries, 41%). Others have reported that landing from a jump is one of the primary noncontact mechanisms for ACL injury in female basketball and volleyball players (Ferretti et al., 1992; Kirkendall & Garrett, 2000). In light of these observations, controlled laboratory experiments have investigated joint mechanics and energetics in landings tasks (Coventry et al., 2006; Decker et al., 2003; DeVita & Skelly, 1992; Kernozek et al., 2005; McLean et al., 2007; Nagano et al., 2009; Pappas et al., 2007; Zhang et al., 2000).

While ACL tears can occur during bilateral landings (Boden et al., 2000), unilateral landings are considered more dangerous due to a decreased base of support and the increased demands of landing on the musculature of only one lower extremity. Ireland (1999) suggested that ACL injuries commonly occur when landing on one leg with knee valgus. Boden et al. (2000) described the mechanism of ACL tear during unilateral landings as the result of knee abduction and a strong contraction of the quadriceps. Olsen et al. (2004) reported that a common mechanism of injury was unilateral jump shot landings while no injuries occurred during bilateral jump shot landings.

Pappas et al. (2007) compared bilateral and unilateral landings and found that, in unilateral landings, subjects performed high-risk kinematics with increased knee abduction, decreased knee flexion, and decreased relative hip adduction. Pappas et al. (2007) also reported that females landed with increased knee abduction and vertical ground reaction force (VGRF) during both types of landing compared with male subjects. Nagano et al. (2009) reported increased knee flexion at contact and peak knee abduction during bilateral drop jumps compared with unilateral landings performed by female athletes. While there is little evidence comparing the biomechanics of unilateral and bilateral landings, studies have suggested
that there are gender differences during bilateral landings that may place females at greater risk of injury in regards to knee flexion (Decker et al., 2003; Dufek & Bates, 1992), knee valgus (Hewett et al., 2004; Kermozek et al., 2005), and VGRF (Kermozek et al., 2005), and during unilateral landings in regard to hip adduction (Hewett et al., 2006), and knee valgus (Russell et al., 2006).

A potential limitation of previous landing studies is the use of an absolute drop height for all subjects, male and female. As landing height (LH) is manipulated there is a divergence in knee joint kinematics between gender for both unilateral (Fagenbaum & Darling, 2003) and bilateral (Huston et al., 2001) landing tasks. The results of these studies indicate that the relative demand of the landing task may vary across individuals, and it may be beneficial to evaluate gender differences in landing mechanics from a height relative to each individual’s maximum jumping ability. In addition, although it is generally accepted that external and internal forces can be mediated by manipulating the lower extremity joint kinematics during landing, no consensus has been reached regarding gender differences in the primary energy absorption strategy. However, the alteration of joint positions at ground contact and throughout the landing can influence the magnitudes and temporal relationships of the peak joint moment and power profiles and thus, mediate stresses placed on internal knee structures (Zhang et al., 2000). Gender differences in landing performances therefore require investigation beyond the kinematic level of analysis to understand the underlying neuromuscular performance criteria by which genders select an energy absorption strategy during landing.

Currently, only two studies have compared unilateral and bilateral landings (Nagano et al., 2009; Pappas et al., 2007), only one having investigated gender differences (Pappas et al., 2007), and neither have reported joint kinetics or energetics. Therefore, the purpose of this study was to determine whether females and males use different landing strategies (kinematic, kinetic and energetic) during unilateral and bilateral landings from a height equal to their individual jumping abilities.

**Methods**

Sixteen recreationally active (defined as physically active for 30 min at least 3 days per week) males (23 ± 3 years, 84.8 ± 11.4 kg, 1.81 ± 0.09 m) and 17 recreationally active females (21 ± 2 years, 62.9 ± 5.9 kg, 1.68 ± 0.06 m) volunteered to participate in this study. A power analysis was performed based on pilot data to detect a between-subject 2° frontal plane difference with an alpha of 0.05 and 80% power. The clinical significance of this kinematic threshold was supported by the results of Withrow et al. (2006). Thirty total subjects (15 each gender) were sufficient to detect this difference, and additional subjects were collected to account for possible dropouts. A background questionnaire to screen for health status and an informed consent form were read and signed by all volunteers before the beginning of the study. The protocol was approved by the university Institutional Review Board (IRB). Volunteers were accepted in the study if they had not suffered a knee injury requiring surgery and had been free of any other injury within the previous six months that could interfere with normal movements. Subjects were also asked how many years they competed (if any) in competitive team sports. Males participated 11 ± 3 years and females participated 6 ± 4 years.

To track lower extremity kinematics, twenty-five light-reflecting skin markers were placed on the subjects. Markers used exclusively for the standing calibration trial (calibration markers) included the left and right iliac crests and greater trochanters, and dominant leg lateral and medial femoral epicondyles, lateral and medial malleoli, first and fifth metatarsal heads. Additional tracking markers were placed on the left and right antero-superior iliac spines, the sacrum, two four-marker plates attached to elastic Velcro straps on the thigh and shank segments, and a marker triad secured on the heel counter of the shoe. Dominant leg was defined as the subject’s preferred jumping leg, which resulted in 29 right-limb and four left-limb dominant subjects. The calibration markers were removed after standing calibration trial completion. Subjects wore standard laboratory footwear (Saucony Jazz, Lexington, MA). Marker trajectories were collected at 200 Hz with a seven-camera Motion Analysis Eagle System (Santa Rosa, CA, USA), and ground reaction forces were collected synchronously at 1000 Hz with an AMTI OR6–5 force plate (Watertown, MA, USA).

There were two sessions for this study. During the first (practice) session, subjects completed three maximum two-leg countermovement jumps on the forceplate to calculate jump height. Custom-made software calculated countermovement jump height from the ground reaction force (GRF) curve (Luhtanen & Komi, 1978). The highest jump height was used to set the platform height for both tasks, unilateral and bilateral landings. The subjects next practiced unilateral and bilateral landings until they could comfortably complete the two tasks.

The second (testing) session was scheduled within one week of the practice session. Following a five-minute jogging warm-up, markers were placed on the subjects, and a two-second calibration trial was collected. Each subject then completed five trials of each landing task. Unilateral and bilateral landings were initiated from a raised platform, placed approximately 10 cm behind the back edge of the forceplate and set to match each subject’s maximal vertical jump height, with the dominant foot leading forward. During bilateral landings subjects landed with their dominant foot upon the forceplate and the other landed next to the forceplate on the landing platform. During each landing task, subjects were encouraged to land with their dominant foot on the center of the force platform. Unsuccessful landings because of loss of balance, or touching the floor with the contralateral limb (during the unilateral landings) were immediately repeated. These two tasks were part of a larger data collection protocol where several unilateral and bilateral landing, jumping, and cutting activities were recorded during
a session that lasted approximately 90 min (O’Connor & Bottum, 2009; O’Connor et al., 2009). The order of tasks was randomized to minimize effects of fatigue.

Data reduction was implemented with Visual3D (v3.89, C-Motion, Inc. Rockville, MD). The raw three-dimensional coordinate data of all markers and GRF data were filtered using a fourth-order, zero lag, recursive Butterworth filter with a cutoff frequency of 18 Hz (Bisseling & Hof, 2006). Right-handed Cartesian local coordinate systems (LCS) for the pelvis, thigh, shank and foot segments of the support leg were defined to describe position and orientation of each segment. Three-dimensional ankle, knee, and hip angles were calculated using a joint coordinate system approach (Grood & Suntay, 1983). Joint centers were given by the midpoint between the medial and lateral calibration markers for the knee and ankle joints (Grood & Suntay, 1983; Wu et al., 2002) and one quarter the distance between the greater trochanter markers in the medial direction (Holden & Stanhope, 1998). Body segment parameters were estimated from Dempster (1955), and joint kinetics were calculated using a Newton-Euler approach (Bresler & Frankel, 1950) and reported in the distal segment reference frame. Initial ground contact was determined by the instant when the vertical GRF exceeded 20 N. The end of the landing phase was defined as the instant of maximum knee flexion. Processed data were time normalized to 101 data points.

To assess the relationship of the lower extremity joint dynamics between the two landings, the touchdown (TD) angles and range of motion (ROM) of the hip, knee, and ankle were reported in all three planes. The ROM values were calculated from touchdown to the peak flexion, abduction, and internal rotation angles. Internal peak extension, adduction, and internal rotation moments were calculated as well. In addition, net joint work in the sagittal plane was calculated for the hip, knee, and ankle for the duration of the landing phase by integrating the respective joint power curve. Positive and negative work values indicate energy production and absorption, respectively. Total energy absorption was calculated by summing the hip, knee, and ankle energy absorption values. Joint moments and the VGRF were normalized to body mass times the square root of LH. We chose to normalize by the square root of LH because for a given body mass the average ground reaction force during the landing phase is approximately proportional to the square root of LH based on the impulse-momentum relationship and the properties of uniformly accelerated motion (Hass et al., 2005). Energy absorption at each joint was normalized by body mass \( \times \) LH because total mechanical energy is directly proportional to LH.

A 2 \( \times \) 2 ANOVA (task \( \times \) gender) was performed for each of the dependent variables. Gender was a between-subjects factor and task was a within-subjects factor. Initially, years of playing experience was included as a covariate, but no influence was found for any variable. Therefore the previously described ANOVA design was used. Significant findings were subsequently examined with Tukey’s post hoc analysis. Significance for all tests was set at \( p < .05 \), and all statistical analyses were performed using SPSS (v16.0, SPSS, Inc. Chicago, IL). Effect sizes for ANOVA (\( \eta^2 \)) were reported.

Results

The mean LH for males was 44.0 ± 7.7 cm, and 28.2 ± 5.4 cm for females. The main effect for landing style showed a 44% larger peak VGRF during unilateral landings compared with bilateral landings \( [p < .001, \eta^2 = 0.929] \) (Figure 1). There were no significant task \( \times \) gender interactions \( [p = .342, \eta^2 = 0.031] \) or gender effects \( [p = .146, \eta^2 = 0.071] \).

In the sagittal plane, there was no significant task \( \times \) gender interactions (Table 1 and Figure 2). The main effect for gender showed that women landed with a significant decrease of 11.3° hip flexion ROM and 8.2° knee flexion ROM. The main effect for landing style showed significant decreases of 2.2° in hip flexion and 5.0° in knee flexion, and a significant increase of 4.4° in plantar flexion at TD during unilateral landings. Unilateral landings were also performed with significant decreases of 10.5° hip flexion ROM, and 14.1° knee flexion ROM.

In the frontal plane, there was a significant task \( \times \) gender interaction for frontal plane touchdown angle at the hip \( [p = .009, \eta^2 = 0.211] \) (Table 2). During bilateral landings men and women landed with similar amounts of hip abduction. However, men performed unilateral landings with a 5.0° increase in hip abduction TD angle compared with a 1.6° increase by women. The main effect for landing style showed significant decreases of 0.8° in knee abduction, and 2.4° in ankle inversion at

**Figure 1** — Group mean VGRF curves during the landing phase from females unilateral (DLSF: solid gray line) and bilateral (DLTF: dashed gray line), and males unilateral (DLSM: solid black line) and bilateral (DLTM: dashed black line).
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TD during unilateral landings. Unilateral landings were also performed with a significant increase of 8.3° in hip abduction ROM, a significant decrease of 3.3° in knee abduction ROM, and a significant increase of 6.4° in frontal plane ankle ROM.

There were no significant task × gender interactions or significant gender effects for peak moments (Table 3 and Figure 3). Unilateral landings resulted in significant increases in hip extension, hip abduction, knee extension, knee abduction and ankle plantar flexion peak moments compared with bilateral landings. There was a significant task × gender interaction for knee energy absorption \( p = 0.067 \) (Figure 4). While the energy absorption at the knee decreased for both men and women during unilateral landings, women exhibited a 0.71 J·kg\(^{-1}\)·LH\(^{-1}\) decrease compared with the 0.06 J·kg\(^{-1}\)·LH\(^{-1}\) decrease exhibited by men. The main effect of landing style showed that unilateral landings were performed with a 1.11 J·kg\(^{-1}\)·LH\(^{-1}\) increase in ankle energy absorption \( p < .001, \eta^2_p = 0.602 \) and a 0.76 J·kg\(^{-1}\)·LH\(^{-1}\) increase in total energy absorption \( p = .008, \eta^2_p = 0.218 \).

**Discussion**

The purpose of this study was to examine gender differences in unilateral and bilateral drop landings. The results of this study indicated that gender differences were only observed in sagittal plane hip and knee ROM, potentially due to the use of a relative drop height. Irrespective of gender, the lower extremity landing mechanics observed in this study were generally consistent with previous studies (Decker et al., 2003; DeVita & Skelly, 1992; Dufek & Bates, 1992; Hewett et al., 2004; Kernozek et al., 2005). Unilateral landings were characterized by increases in TD angles were accompanied by significantly less sagittal and frontal plane hip and knee ROM and an increase in frontal plane ankle motion. Unilateral landings were also characterized by increases in peak hip extension and abduction moments, peak knee extension and abduction moments, and peak ankle plantar flexion moments. These changes in landing strategy, accompanied by the observed increase in utilization of the ankle musculature for impact absorption during unilateral landings may increase knee injury risk during an unbalanced landings (Decker et al., 2003).

A primary goal of this study was to determine if males and females differed in how they adjust to a unilateral landing, which would be detected by a significant task × gender interaction. We found significant task × gender interactions for hip abduction angle at TD and
energy absorption at the knee, suggesting that males and females used different landing strategies when adjusting to unilateral landings. During bilateral landings, both males and females landed with similar amounts of hip abduction and limited ROM. However, during unilateral landings subjects demonstrated a 3.1° increase in hip abduction at TD and an 8.3° increase in subsequent adduction throughout the landing, with this effect being exacerbated in male subjects who demonstrated a 3.4° greater increase in hip abduction TD angle compared with female subjects. The finding of increased hip abduction at TD during unilateral landings is in agreement with Pappas et al. (2007) although they did not report a significant task × gender interaction. The authors concluded that the increased abduction at TD during unilateral landings allows the hip to move through a greater ROM toward adduction, keeping the gluteus medius closer to its resting length and allowing increased control of deceleration (Pappas et al., 2007).

Gender differences were only observed in sagittal plane hip and knee ROM, which is contrary to previous reports of gender differences during bilateral landings.
Table 2  Effects of landing style and gender on mean (SD) for touchdown angle (TD) and range of motion (ROM) of the hip, knee and ankle in the frontal plane. Positive sagittal plane TD angles represent hip flexion, knee extension, and ankle dorsiflexion. ROM values represent the excursion from TD to the prominent peak values in the direction of the arrows in Figure 2.

<table>
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<tr>
<th></th>
<th>Hip</th>
<th>Knee</th>
<th>Ankle</th>
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<tr>
<td></td>
<td>TD (deg)‡</td>
<td>RoM (deg)*</td>
<td>TD (deg)*</td>
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<tr>
<td>1 Leg</td>
<td></td>
<td></td>
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</tr>
<tr>
<td>Male</td>
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<td>11.7 (5.8)</td>
<td>0.9 (2.8)</td>
</tr>
<tr>
<td>Female</td>
<td>–7.3 (7.5)</td>
<td>12.2 (5.0)</td>
<td>0.2 (2.8)</td>
</tr>
<tr>
<td>Mean</td>
<td>–9.1 (7.9)</td>
<td>12.0 (5.2)</td>
<td>0.5 (2.8)</td>
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<tr>
<td>2 Legs</td>
<td></td>
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<tr>
<td>Male</td>
<td>–6.4 (6.1)</td>
<td>3.3 (2.7)</td>
<td>0.1 (2.8)</td>
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<tr>
<td>Female</td>
<td>–5.7 (5.0)</td>
<td>3.9 (2.5)</td>
<td>–0.7 (3.1)</td>
</tr>
<tr>
<td>Mean</td>
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<td>3.7 (2.5)</td>
<td>–0.3 (2.9)</td>
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<td>Overall</td>
<td></td>
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<tr>
<td>Male</td>
<td>–8.9 (7.4)</td>
<td>7.5 (6.1)</td>
<td>0.5 (2.8)</td>
</tr>
<tr>
<td>Female</td>
<td>–6.5 (7.5)</td>
<td>8.1 (5.0)</td>
<td>–0.3 (2.8)</td>
</tr>
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</table>

| Task × Gender | p = 0.012 | p = 0.936 | p = 0.809 | p = 0.201 | p = 0.635 | p = 0.835 |
| Task         | p = 0.001 | p = 0.004 | p < 0.001 | p < 0.001 | p < 0.001 | p < 0.001 |
| Gender       | p = 0.316 | p = 0.679 | p = 0.448 | p = 0.921 | p = 0.736 | p = 0.773 |

‡Indicates significant task × gender interaction (p < .05).
*Indicates significant task main effect (p < .05).

Table 3  Effects of landing style and gender on mean (SD) peak joint moments of the hip, knee, and ankle during the landing phase for the sagittal and frontal planes. Positive sagittal, and frontal plane moments represent internally applied hip flexion/adduction, knee extension/adduction, and ankle dorsiflexion/inversion moments.

<table>
<thead>
<tr>
<th></th>
<th>Hip</th>
<th>Knee</th>
<th>Ankle</th>
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<tbody>
<tr>
<td></td>
<td>Extension*</td>
<td>Abduction*</td>
<td>Extension*</td>
</tr>
<tr>
<td></td>
<td>Nm·kg⁻¹·LH⁻¹/²</td>
<td>Nm·kg⁻¹·LH⁻¹/²</td>
<td>Nm·kg⁻¹·LH⁻¹/²</td>
</tr>
<tr>
<td>1 Leg</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Male</td>
<td>–2.9 (1.2)</td>
<td>–3.0 (0.8)</td>
<td>3.6 (1.0)</td>
</tr>
<tr>
<td>Female</td>
<td>–3.0 (1.6)</td>
<td>–3.4 (0.7)</td>
<td>3.4 (1.2)</td>
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<tr>
<td>Mean</td>
<td>–2.9 (1.4)</td>
<td>–3.2 (0.7)</td>
<td>3.5 (1.1)</td>
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<tr>
<td>2 Legs</td>
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<tr>
<td>Male</td>
<td>–2.3 (1.0)</td>
<td>–1.5 (0.7)</td>
<td>3.0 (0.9)</td>
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<tr>
<td>Female</td>
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<td>–1.5 (0.5)</td>
<td>3.0 (1.3)</td>
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<tr>
<td>Mean</td>
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<td>–1.5 (0.6)</td>
<td>3.0 (1.1)</td>
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<tr>
<td>Overall</td>
<td></td>
<td></td>
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</tr>
<tr>
<td>Male</td>
<td>–2.6 (1.2)</td>
<td>–2.3 (1.1)</td>
<td>3.3 (1.0)</td>
</tr>
<tr>
<td>Female</td>
<td>–2.7 (1.4)</td>
<td>–2.4 (1.1)</td>
<td>3.2 (1.2)</td>
</tr>
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</table>

| Task × Gender | p = 0.780 | p = 0.110 | p = 0.090 | p = 0.134 | p = 0.374 | p = 0.297 |
| Task         | p < 0.001 | p < 0.001 | p < 0.001 | p < 0.001 | p < 0.001 | p = 0.255 |
| Gender       | p = 0.833 | 0.523     | 0.836     | 0.779     | 0.967     | 0.252     |

*Indicates significant task main effect (p < .05).
that may place females at greater risk of injury. Previous studies have reported gender differences in knee flexion (Decker et al., 2003; Dufek & Bates, 1992), knee valgus angle (Hewett et al., 2004; Kernozek et al., 2005), normalized VGRF (Kernozek et al., 2005), and energy absorption (Decker et al., 2003). Gender differences in unilateral landings have been reported for hip adduction (Hewett et al., 2006) and knee valgus angle (Russell et al., 2006). The absence of gender differences in the current study may be attributed to use of a relative drop height versus the commonly used absolute drop height (Decker et al., 2003; Dufek & Bates, 1992; Kernozek et al., 2005; Pappas et al., 2007; Schmitz et al., 2007). This is also consistent with the results of Huston et al. (2001), who found similar joint mechanics between genders at lower drop heights but divergence as drop height increased. It could therefore be hypothesized that the use of an absolute landing height places a greater demand on female subjects and may be a contributing factor in landing mechanics gender differences observed in previous studies.

One advantage to utilizing the same absolute drop height is that it equalizes the total mechanical energy
(when scaled by body mass) at impact across all subjects. This makes absolute comparisons across genders easier. By utilizing different drop heights, the absolute mechanical energies will necessarily differ due to the different velocities at impact. Therefore, a challenge in the current study was to normalize kinetic data to allow for direct comparison. Hass et al. (2005) chose to scale joint moments by body mass, body height, and the square root of drop height. Our decision to omit body height from the scaling was based on our assessment of the appropriate scaling for energetic data. Mechanical energy is explicitly tied to body mass and velocity at impact, but scaling energy by body height was difficult to justify mechanically. Therefore, to report similar units for joint moments and work, body height was not included. The result of this normalization was that there was only one task × gender interaction and no gender differences in any kinetic variable, even though there were some kinematic differences. Based on this result it could be concluded that when faced with the same relative demand, males and females used different kinematic strategies to have essentially the same joint loading.

When comparing the two landing styles independent of gender, touchdown angles for unilateral landings were characterized by increased extension and adduction at the knee, and increased plantar flexion and inversion at the ankle, which would best be classified as a “stiff” landing and have been described as having the muscular and passive tissue structures of the ankle joint absorb the greater shares of energy than during soft landings (DeVita & Skelly, 1992; Zhang et al., 2000). These changes in TD angles were accompanied by significantly less sagittal and frontal plane hip and knee ROM and an increase in frontal plane ankle motion. It appears that while the sagittal plane ROM is restricted, there is compensation at the ankle in the frontal plane to absorb the landing. Unilateral landings were also characterized by increases in peak hip extension and abduction moments, peak knee extension and abduction moments, and ankle plantar flexion moments. The increased sagittal plane moments during stiff landings are consistent with previous findings (DeVita & Skelly, 1992), and the increased hip abduction is a logical consequence of the need to support the trunk.

Subjects also performed unilateral landings with significantly less knee flexion at TD and subsequent ROM, which is consistent with previous findings (Nagano et al., 2009; Pappas et al., 2007). Pappas et al. (2007) speculated that subjects may attempt to prevent falls during unilateral landings by limiting excessive knee flexion while simultaneously increasing the forces in the ACL by landing with the knee closer to full extension at TD. Landing with the knee close to extension has been suggested to be a predisposing factor to knee injury (Decker et al., 2003; Huston et al., 2001). Markolf et al. (2004) demonstrated that ACL forces caused by quadriceps contraction increase in the last 40 degrees of knee extension, peaking close to full extension.

Consistent with the findings of Nagano et al. (2009), unilateral landings were performed with less knee abduction at TD and subsequent ROM. Typically, increased knee abduction during landing is considered to be a predisposing factor to ACL injury (Hewett et al., 2005), and is commonly described as a part of the ACL injury mechanism (Boden et al., 2000; Ireland, 1999; Olsen et al., 2004). Nagano et al. (2009) speculated that knee abduction was limited during unilateral landings to compensate for greater internal tibial rotation and smaller knee flexion to prevent ACL injury. Therefore, it is thought that bilateral landings with greater knee abduction might also pose a risk for ACL injury.

Comparison of the preferred energy absorption strategies during unilateral and bilateral landing may provide an understanding for the gender disparity in noncontact lower extremity injury mechanism. Therefore, a secondary statistical analysis was performed on the relative energy absorption at each joint during the unilateral and bilateral styles across gender. At the hip there was a significant task × gender interaction for relative energy absorption [p = .017, ηp² = 0.182]. Men performed unilateral landings with 4% less energy absorption at the hip, while women showed a 2% increase during unilateral landings. This may due to the fact that much of the work done by the hip extensor muscles was used to counteract the trunk motion (DeVita & Skelly, 1992). For both the ankle and knee, there were significant main effects for landing style (ankle: p < .001, ηp² = 0.520; knee: p < .001, ηp² = 0.564). For both men and women, unilateral landings shifted greater energy absorption to the ankle and away from the knee. While the knee extensors were the muscle group primarily responsible for reducing the body’s kinetic energy, the relative contributions of the ankle plantar flexors increased with an approximately

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Figure 4 — Group mean hip, knee, ankle, and total sagittal plane joint work during the landing phase from females unilateral (DLS-F: solid gray box) and bilateral (DLT-F: dashed gray box), and males unilateral (DLS-M: solid black box) and bilateral (DLT-M: dashed black box). *Indicates significant task × gender interaction (p < .05). †Indicates significant task main effect (p < .05).
equal reduction in the contribution of the knee extensors during unilateral landings. These results are consistent with previous research (DeVita & Skelly, 1992; Zhang et al., 2000) that reported a greater relative contribution of the ankle plantar flexors as knee extension angles increased at TD.

An interesting finding of the current study was that the summated joint work values were $-7.5$ and $-6.7$ J-kg$^{-1}$-LH$^{-1}$ for unilateral and bilateral landings, respectively. However, the bilateral work value only represents the energy absorption in one limb and therefore, the total energy absorption at all joints would likely be double that value (approx. $-13.4$ J-kg$^{-1}$-LH$^{-1}$) when accounting for left and right sides. In contrast the single limb during 1·LH$^{-1}$ for unilateral and bilateral landings, respectively. That value (approx. –13.4 J·kg$^{-1}$·LH$^{-1}$) when accounting for left and right sides. In contrast the single limb during 1·LH$^{-1}$ for unilateral and bilateral landings, respectively. However, the bilateral work value only represents the energy absorption in one limb and therefore, the total energy absorption in one limb and therefore, the total energy absorption would likely be double that value (approx. $-13.4$ J-kg$^{-1}$-LH$^{-1}$) when accounting for left and right sides. In contrast the single limb during 1·LH$^{-1}$ for unilateral and bilateral landings, respectively. Therefore, an additional 5.9 J·kg$^{-1}$·LH$^{-1}$ must be absorbed passively at the joints through the net joint forces. This result is in agreement with Žatsiorsky and Prilutsky (1987) who found that, in stiff landings from a drop, up to 75% of the energy could be dissipated passively. This also indicates that the passive structures experience much greater energy absorption that may lead to injury.

In the current study, gender differences were only observed in sagittal plane hip and knee ROM, potentially due to the use of a relative drop height versus the commonly used absolute drop height. These results, along with the previously reported divergence in knee joint kinematics between gender as landing height increases for both unilateral (Fagenbaum & Darling, 2003) and bilateral (Huston et al., 2001) landing tasks suggest that relative demand of the landing task is an important factor to consider in future studies. Irrespective of gender, unilateral landings were characterized by significant differences in hip and knee kinematics that have been linked to increased injury risk and would best be classified as “stiff” landings. The ankle musculature was used more for impact absorption during unilateral landing, which required increased joint extension at touchdown and may increase knee injury risk during an unbalanced landing. In addition, there was only an 11% increase in energy absorption by the net joint moments during unilateral landings, suggesting that there was a substantial amount of passive energy transfer during unilateral landings. Based on these results, it is not surprising that noncontact ACL injuries more commonly occur during single-leg impacts.

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References


Comparison of Uni- and Bilateral Landings


