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Original Research

Temporal Kinematic Differences throughout Single and Double-Leg Forward Landings

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Abstract

Screening methods for anterior cruciate ligament (ACL) injuries often involve double-leg landings, though the majority of ACL injuries occur during single-leg landings. Differences in kinematic temporal characteristics between single-leg and double-leg landings are poorly understood. The purpose of this study was to examine discrete and temporal kinematics associated with functional valgus collapse during single-leg and double-leg landings (LAND_{SL} and LAND_{DL}). Three-dimensional kinematics were obtained during the landing phases of LAND_{SL} and LAND_{DL} in ninety participants (45 females: 20.1±1.7 yr, 165.2±7.6 cm, 68.6±13.1 kg; 45 males: 20.7±2.0 yr, 177.7±8.5 cm, 82.8±16.3 kg). Peak joint angles and time series curves for frontal and transverse plane hip and knee kinematics were analyzed with an RMANOVA (discrete variables) and Statistical Parametric Mapping (SPM) paired t-tests (time series). LAND_{SL} elicited greater knee abduction than LAND_{DL} from 0-35% (0-73ms) but greater knee adduction from 54-100% (112-207ms). Peak knee abduction was 2.0° greater during LAND_{DL} than during LAND_{SL} ($p<.001$). LAND_{SL} elicited greater hip adduction than LAND_{DL} from 2-33% (4-69ms) and greater hip abduction from 49-100% (102-207ms). Peak hip adduction was 4.6° greater during LAND_{SL} than during LAND_{DL} ($p<.001$). LAND_{SL} elicited less knee internal rotation from 0-3% and greater hip internal rotation from 52-75% of the landing phase. Peak transverse plane joint angles did not differ between tasks. During the time frame in which ACL injuries are thought to occur, LAND_{SL} elicited frontal plane knee and hip movement consistent with risky biomechanics. Researchers and clinicians should be cognizant of how a chosen screening task alters observed kinematic effects.

Introduction

Of the more than 350,000 anterior cruciate ligament (ACL) injuries that occur annually in the United States, an estimated 72% occur through non-contact mechanisms (Moses, Orchard, & Orchard, 2012; Wojtys & Brower, 2010). Functional valgus collapse, a movement pattern consisting of frontal and transverse plane hip and knee motion (specifically greater knee [tibial] abduction, knee [tibial] internal rotation, hip [femoral] adduction, and hip [femoral] internal rotation), is thought to increase the risk of sustaining a non-contact ACL injury (Boden, Torg, Knowles, & Hewett, 2009; Hewett et al., 2005; Ireland, 1999; Tron Krosshaug et al., 2007). For clinical screening purposes, functional valgus collapse is often assessed during double-leg landings (Cesar, Tomasevicz, & Burnfield, 2016; Fox, Bonacci, McLean, Spittle, & Saunders, 2016; Hewett et al., 2005). However, these methods have been criticized for their lack of ecological validity (Fox, Spittle, Otago, & Saunders, 2013) and lack of utility in predicting ACL injury (T. Krosshaug et al., 2016; Smith et al., 2012).

Given that approximately 72% of ACL injuries occur during a single-leg stance (Boden, Torg, Knowles, & Hewett, 2009), it is reasonable to contend that single-leg landings may present a greater challenge to maintaining safe lower extremity mechanics. Thus, screening tasks incorporating single-leg landings may better expose problematic movement patterns. To support this, the use of more demanding single-leg landings has proven more effective in identifying risky biomechanics than studies using less challenging double-leg landings (Fox, Bonacci, McLean, Spittle, & Saunders, 2014; Hollman et al., 2009; Homan, Norcross, Goerger, Prentice, & Blackburn, 2013; Howard, Fazio, Mattacola, Uhl, & Jacobs, 2011; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007; Lawrence, Kernozek, Miller, Torry, & Reuteman, 2008; Nguyen, Shultz, Schmitz, Luecht, & Perrin, 2011; Sigward, Ota, & Powers, 2008; Taylor, Ford, Nguyen,

& Shultz, 2016; Thijs, Van Tiggelen, Willems, De Clercq, & Witvrouw, 2007; Weinhandl, Irmischer, & Sievert, 2015; Willson, Ireland, & Davis, 2006). Even so, comprehensive biomechanical differences between single-leg and double-leg landings have yet to be elucidated. Examining this methodological distinction and its impact on injury screening processes is necessary so that future studies can unearth underlying factors that specifically contribute to kinematics occurring during single-leg activities where the ACL is more vulnerable.

Another important consideration in the comparison between single and double-leg tasks is the timing of risky kinematics that may influence chance of injury. ACL injuries are thought to occur 30-50 milliseconds (ms) after initial ground contact of a landing task (T Krosshaug, Slauterbeck, Engebretsen, & Bahr, 2007). Current kinematic analyses often focus on peak joint angles, assuming the most extreme value coincides with this riskiest time interval. Though this approach is well-accepted in the literature, comparing peak values ignores potentially important temporal fluctuations occurring throughout the landing phase. Previous research investigating temporal biomechanics has been promising and has rendered a more complete picture of movement patterns (Hébert-Losier et al., 2015; Hébert-Losier, Schelin, Tengman, Strong, & Häger, 2018; Smeets et al., 2018). Statistical Parametric Mapping is one such approach that allows for the comparison of complete time series curves by accounting for the dependency of adjacent time points in its calculation of the appropriate significance threshold (Pataky, Robinson, & Vanrenterghem, 2013), thus circumventing the conventional need to stringently adjust for Type I error rates. Employing a more holistic statistical technique to the study of task differences may provide more complete information of how single-leg and double-leg landings differ and thus better inform clinicians and researchers in the development of more individualized screening and intervention programs.

Therefore, the purpose of this study was to examine and compare frontal and transverse plane hip and knee motions throughout the landing phases of single-leg and double-leg landings (LAND_{SL} and LAND_{DL}) using both conventional, discrete kinematic analysis and statistical parametric mapping (SPM) analysis. We hypothesized that, compared to the double-leg landing, the single-leg landing would elicit greater functional valgus collapse, operationally defined as the angles associated with greater knee abduction and internal rotation and hip adduction and internal rotation, and that these differences would be most pronounced during the time frame 30-50 milliseconds after initial ground contact.

Methods

Participants. Ninety (45 females: 20.1±1.7 yr, 165.2±7.6 cm, 68.6±13.1 kg; 45 males: 20.7±2.0 yr, 177.7±8.5 cm, 82.8±16.3 kg) participants were recruited for a single testing session. This was part of a larger study examining comprehensive biomechanical patterns (Hogg et al, in prep) To ensure a homogenous sample, the study was limited to healthy adults who were: 1) between the ages of 18 and 25 and 2) who scored two or more (at least “one time in a week”) on categories 2-4 (“cutting,” “decelerating,” and “pivoting”) of the Marx activity rating scale (Marx, Stump, Jones, Wickiewicz, & Warren, 2001). Participants were excluded if they had: 1) any history of knee surgery, 2) any history of ligamentous or meniscal knee injury, 3) any lower extremity injury within the previous 6 months, 4) history or diagnosis of a vestibular condition affecting balance, or 5) history or diagnosis of any cardiovascular condition precluding exercise. Each participant provided written informed consent as approved by the university’s Institutional Review Board.

Biomechanical Instrumentation. Prior to digitization for motion capture, participants were outfitted with standardized shoes (Adidas Uraha 2, Adidas AG, Herzogenaurach, Bavaria)

to eliminate potential confounding effects caused by different shoe-surface interactions. Participants were adorned with six marker clusters, placed at each of the following locations: lateral aspect of the left foot, lateral aspect of the left lower leg (mid-shaft), medial and lateral proximal tibial flares, the lateral left thigh (mid-shaft), the L5-S1 junction, and the postero-superior thorax (C7-T1 spinous processes). Participants were then digitized using the MotionMonitor software (Innovative Sports Training, Chicago, IL). Joint centers for the knee and ankle were defined as the midway point between the medial and lateral femoral condyles and medial and lateral malleoli, respectively. The hip joint center was determined using the Bell method (Bell, Brand, & Pedersen, 1989). A segment-based coordinate system was used to define each body segment. The X-axis was defined as the anterior-posterior axis (adduction/abduction), the Y-axis was the distal-proximal axial axis (internal/external rotation), and the Z-axis was defined as the medial-lateral axis (flexion/extension). Motions for each joint were calculated using Euler's equations (Z Y' X'') (Kadaba, Ranakrishnan, Wootten, Gainey, & Cochran, 1989). Kinematic data were obtained with an 8-camera optical LED system (Impulse, Phase Space; San Leandro, CA) at a sampling rate of 240 Hz. Kinetic data were obtained using an embedded forceplate (Type 4060-130; Bertec Corporation, Columbus OH, USA) and were sampled at a rate of 1000 Hz. Kinematic and kinetic instrumentation were interfaced with MotionMonitor software and manually synced by a pulse trigger during each trial. Following acquisition, the kinematic data were interpolated to 1000 Hz to allow synchronization of data sources.

Procedure for Single and Double-Leg Forward Landing. Participants were familiarized to each task prior to data collection and were allowed up to three practice trials. For both LAND_{SL} and LAND_{DL} tasks, tape was placed on the ground at a distance equal to 40% of each participant's height away from the front edge of the force plate. A foam barrier equal to

15% of the participant's height (Jacobs et al., 2007) was placed halfway between the tape and the edge of the forceplate. For the LAND_{SL}, participants were instructed to stand behind the tape and jump from two legs over the foam barrier, landing on their left foot. The procedure for the LAND_{DL} was the same in every way except the participants landed on both feet, with the left foot landing entirely on the forceplate, and the right foot landing completely off the forceplate. Trials were discarded if the participant double-hopped upon landing, hit the barrier, didn't clear the barrier with both feet, didn't land with the entire left foot on the forceplate, or used their contralateral limb for additional support (LAND_{SL} only). Five clean trials for each task were collected and used for analysis. The order of the two landing tasks was counterbalanced.

Data Handling and Processing. Motion capture was collected two seconds before initial ground contact, defined as the point at which the vertical ground reaction force exceeded 10N, and continued for three seconds after initial ground contact, for a total of five seconds. All flexions, adductions, and internal rotations were defined as positive angles. All extensions, abductions, and external rotations were defined as negative angles. To determine the appropriate filter for kinematic and kinetic data, a residual analysis was conducted on the ground reaction force (GRF) in a subset of the trials, according to the procedure described by Winter (Winter, 1990). The signal to noise ratio was deemed to be most optimum at 10Hz. Therefore, all kinematic and kinetic data were filtered with a 10 Hz low-pass, zero-lag, 2nd order Butterworth filter. Data were then exported to MATLAB (MathWorks, Inc., Natwick, MA) for further reduction using custom-written script.

For conventional analysis using discrete variables, MATLAB script was written to extract peak joint angles from the landing phase (interval encompassing the point at which vertical ground reaction force exceeded 10N through the point of maximum knee flexion). Peak hip and

knee angles were operationally defined as the maximum respective joint angle occurring during the landing phase.

For SPM analyses, all data points of the landing phase (initial ground contact until the point of maximum knee flexion) were extracted for all kinematic data. Custom MATLAB script was written to create time series curves of 101 data points (representing 0-100% of the landing phase), equally spaced, for each biomechanical variable, resulting in separate ensemble curves for each variable by task.

Statistical Approach. Because time series curves were registered to 100%, a paired sample t-test was conducted to verify that LAND_{SL} and LAND_{DL} were completed in comparable amounts of time.

For conventional analysis of discrete variables, a single repeated measures MANOVA was used to detect differences between tasks in hip and knee peak joint angles. Statistical significance was set at 0.05 ($\alpha = 0.05$). SPSS (Version 24, IBM Corp, Armonk, NY) was used for repeated measures analysis. Cohen's *d* effect sizes were also calculated to aid in interpretation of results (small > 0.2, medium > 0.5, large > 0.8) (Cohen, 1988).

For SPM analysis, separate SPM paired t-tests were used to compare the following biomechanical time series curves between tasks: knee adduction/abduction angle, knee internal/external rotation angle, hip adduction/abduction angle, and hip internal/external rotation angle. All SPM analyses were conducted using MATLAB script developed by Todd Pataky (Pataky, T., 2016, www.spm1d.org).

Results

Landing Phase Lengths. Landing phase lengths were not different between landing tasks ($p = .42$) where time of completion (initial ground contact through maximum knee flexion)

was 205.6±34.5 ms (range: 152.5 – 303.4 ms) for LAND_{SL} and 209.4±41.8 ms (range: 139.5 – 327.5 ms) for LAND_{DL}.

Omnibus Repeated Measures. Peak hip and knee joint angles differed by task ($\lambda=.44$, $p<.001$, $\eta^2 = .56$).

Table 1. Means ± Standard Deviations (°) for Hip and Knee Peak Joint Angles.

	Single-leg Landing	Double-leg Landing	<i>Total</i>
Peak Hip Adduction (+)	-0.2±6.8	-4.8±6.5	-2.5±6.2*
Peak Hip IR (+)	.01±5.7	.2±4.6	0.1±4.7
Peak Knee Abduction (-)	-1.2±6.4	-3.2±7.4	-2.2±6.4*
Peak Knee IR (+)	12.3±11.1	13.2±11.4	12.7±10.9

IR = internal rotation; ER = external rotation

*significant difference between tasks ($p < .05$)

Knee Abduction. SPM analysis revealed that LAND_{SL} resulted in greater knee abduction from 0-35% than did the corresponding section of LAND_{DL} ($p=.02$). Conversely, from 54-100%, participants displayed greater knee abduction during LAND_{DL} as compared to

LAND_{SL} ($p=.01$) (Figure 1c). Compared to LAND_{SL}, participants displayed 2.0° greater peak knee abduction angles during LAND_{DL} ($d=.29$; $p<.001$).

Knee Internal Rotation. SPM analysis revealed greater knee internal rotation during 0-3% (0-10 ms) of LAND_{DL} when compared to LAND_{SL} ($p=.05$) (Figure 1d). There were no differences between LAND_{SL} and LAND_{DL} for peak knee internal rotation angle ($d=.08$; $p=.15$).

Hip Adduction. SPM analysis revealed greater hip abduction during the LAND_{SL} occurred from 2-33% ($p=.02$) and greater hip adduction occurred from 49-100% ($p=.01$) when compared to corresponding time frames of the LAND_{DL} (Figure 1a). Compared to LAND_{DL}, subjects displayed 4.6° greater peak hip adduction ($d=.69$; $p<.001$) during LAND_{SL}.

Hip Internal Rotation. SPM revealed greater hip external rotation during LAND_{DL} from 52-75% of the landing phase when compared to the corresponding section of LAND_{SL} ($p=.03$) (Figure 1b). There were no differences between LAND_{SL} and LAND_{DL} for peak hip internal rotation angle ($d=.06$; $p=.66$).

Figure 1

Discussion

We hypothesized that, compared to the double-leg landing, the single-leg landing would elicit greater functional valgus collapse, operationally defined as the angles associated with greater knee abduction and internal rotation and hip adduction and internal rotation, and that these differences would be more pronounced during the time frame 30-50 milliseconds after initial ground contact. Our results partially supported this. In agreement with our hypothesis, SPM demonstrated task differences during the first third of the landing phase for frontal plane hip and knee motion. SPM did identify differences in transverse plane hip movement, but they were limited to the second half of the landing phase, and therefore are not as critical to ACL-injury mechanisms. Peak joint angles, particularly in the frontal plane, were more indicative of valgus collapse during the LAND_{SL}. Peak hip adduction was 4.6° greater and peak knee abduction 2.0° greater during the LAND_{SL}. This is in line with previous work demonstrating greater peak knee abduction during double-leg landings as opposed to single-leg landings (Fox et al., 2014; Taylor et al., 2016). There were no differences in peak transverse plane angles.

Greater knee abduction is often referenced as a risky movement pattern. Though discrete peak knee abduction is frequently used as the sole marker of greater knee abduction, knee abduction occurring during the early phases of landing may be more important than the absolute peak (Barry P Boden et al., 2009; Hewett et al., 2005; Tron Krosshaug et al., 2007). To our knowledge, this is the first study to investigate the timing of knee abduction between tasks. In the current study, the average time for completion of the functional task was 207.50 ± 38.17 ms (LAND_{SL}: 205.6 ± 34.5 ms; LAND_{DL}: 209.4 ± 41.8 ms). During LAND_{DL}, peak knee abduction occurred at approximately 75-80% of the landing phase, or 155-166ms after initial ground

contact, which was much later than what was observed for LAND_{SL} (15-20% of the landing phase, or 31-42ms after initial ground contact) (Figure 1C). Therefore, even though peak knee abduction is greater during the LAND_{DL}, the earlier peak during the LAND_{SL} may be of greater concern. Our SPM analysis demonstrated that from initial ground contact until 73ms (0-35%) after contact, LAND_{SL} knee abduction was 2-3° greater than knee abduction during the LAND_{DL}. Coupled with previous research documenting that 72% of ACL injuries occur during a single-leg stance (B P Boden et al., 2009), this information suggests that the peak knee abduction observed earlier in the landing phase of LAND_{SL} may be more indicative of high-risk biomechanics than the greater absolute knee abduction observed during the second half of the LAND_{DL}.

Although not the primary focus of our research question, it is interesting to note the *rate* of change in frontal plane knee angle. Figure 1C demonstrates that within the first 20% of the landing phase, the slope in the single-leg landing is noticeably steeper than the slope during the double-leg landing, suggesting that movement velocity is different between the two landings. Given the visco-elastic nature of ligaments, displacements occurring over shorter periods of time may influence the ACL differently than similar displacements occurring over longer periods of time. More research is needed to examine the consequences of frontal plane knee velocity on biomechanics and subsequent injury risk.

Previous research has indicated that peak frontal plane hip motion couples with peak frontal plane knee motion during single-leg cutting maneuvers, accounting for as much as 25% of the variance in knee abduction angles (Imwalle, Myer, Ford, & Hewett, 2009). In the current data, a post-hoc bivariate correlation revealed that peak hip adduction accounted for 37% of the variance in peak knee abduction during the LAND_{SL} ($p < .001$), suggesting the hip-knee coupling also occurs in landing tasks. This concept is further supported by the SPM analysis, as task

differences in hip and knee frontal plane motion span similar phases of the landing task (Figures 1a & c). Therefore, evidence demonstrating hip-knee frontal plane coupling has been limited to peak joint angles. The current data demonstrates, as would be expected, that hip-knee coupling is not limited to peak position but extends throughout the landing phase. Because of the hip's frontal plane coupling with the knee, exaggerated hip movement in a single-leg stance could be problematic for knee control (Figure 1c). During the first third of the LAND_{SL} (2-33%), participants' hips were more abducted. This is possibly due to a lateral trunk lean early in the landing phase, resulting in a relative hip abduction. The pattern is reversed during the latter half of the landing phase (49-100%), when participants displayed greater hip adduction in the LAND_{SL} than in the LAND_{DL}. The greater total frontal plane movement during the LAND_{SL} suggests that the hip is freer to adduct during single-leg landing, whereas during a double-leg land there is minimal frontal plane hip movement. With only one limb contacting the ground, a contralateral pelvic drop is more likely, and possibly led to greater movement toward hip adduction later in the LAND_{SL}, while the apparent splinting of the joint during the LAND_{DL} was likely a result of using dual frontal plane supports. Though intuitive, this distinction is nonetheless important. In order to remain upright during a single-limb stance, the body's center of mass must shift laterally, which then initiates a reactive hip adductor moment (Barry P Boden et al., 2009; Hewett & Myer, 2011). Considering the more extreme nature of frontal plane hip motion during LAND_{SL}, this may be an important consideration for maintaining knee control during single-leg activity.

The current evidence has implications for injury screening and intervention. It has been previously noted that frontal plane kinematics are critical to ACL injury (OKane et al., 2016), and can easily be screened prior to activity (Fox et al., 2016). The vertical drop jump task that is

commonly used to assess frontal plane knee kinematics has been criticized for its poor sensitivity and specificity when discriminating between ACL-injured and healthy individuals (T. Krosshaug et al., 2016), and for its lack of ecological validity (Fox et al., 2016). The current evidence suggests that a single-leg task, where frontal plane movement control is more challenging, could be a more appropriate screening task, though prospective research is needed to confirm the efficacy of a single-leg landing task for ACL injury screening.

Limitations. The authors acknowledge limitations in the study design and analyses used. We did not stratify our analysis by sex. Hip and knee movement, particularly knee abduction, is known to be different between sexes, and we acknowledge that it is best practice to analyze within each sex. To address this concern, supplementary sex-stratified analyses are provided in the appendix. In order to accurately compare time series curves between tasks, it was necessary to standardize the landing phase lengths to 100%. The time it took for task completion was variable, and we acknowledge that registering the time series curves to 100% may have masked some temporal inter-subject variability, as well as the absolute timing of when differences occur, which may be relevant to injury risk potential. However, as there were no significant differences in landing phase lengths between tasks, normalization to 100% likely did not have an appreciable effect on our findings. Lastly, because of the relatively homogenous sample (young healthy adults between 18-25 years), our results may not be generalizable to other populations, including adolescents.

In conclusion, frontal plane hip and knee kinematics are appreciably different between single-leg and double-leg landing tasks. During the time frame in which ACL injuries are thought to occur, single-leg landing elicits frontal plane knee movement consistent with functional valgus collapse. In light of the apparent hip-knee coupling in the frontal plane,

exaggerated hip movement observed during LAND_{SL} may be an important consideration for clinicians seeking to screen aberrant movement patterns. Lastly, when screening for faulty biomechanics, clinicians should be cognizant of how the chosen screening task can significantly alter observed biomechanical effects. Further work is needed to determine if these biomechanical task differences represent healthy adaptations to varied movement demands or if they portend an increased risk for injury.

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Conflict of Interest Statement. The authors report no conflicts of interest.

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