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Kinetic and kinematic differences between first and second landings of a drop vertical jump task: Implications for injury risk assessments☆

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Abstract

Background—Though the first landing of drop vertical jump task is commonly used to assess biomechanical performance measures that are associated with anterior cruciate ligament injury risk in athletes, the implications of the second landing in this task have largely been ignored. We examined the first and second landings of a drop vertical jump for differences in kinetic and kinematic behaviors at the hip and knee.

☆All authors were fully involved in the study and preparation of the manuscript and the material within has not been and will not be submitted for publication elsewhere.

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Methods—Acohort of 239 adolescent female basketball athletes (age = 13.6 (1.6) years) completed drop vertical jump tasks from an initial height of 31 cm. A three dimensional motion capture system recorded positional data while dual force platforms recorded ground reaction forces for each trial.

Findings—The first landing demonstrated greater hip adduction angle, knee abduction angle, and knee abduction moment than the second landing (P-values < 0.028). The second landing demonstrated smaller flexion angles and moments at the hip and knee than the first landing (P-values < 0.035). The second landing also demonstrated greater side-to-side asymmetry in hip and knee kinematics and kinetics for both the frontal and sagittal planes (P-values < 0.044).

Interpretation—The results have important implications for the future use of the drop vertical jump as an assessment tool for anterior cruciate ligament injury risk behaviors in adolescent female athletes. The second landing may be a more rigorous task and provides a superior tool to evaluate sagittal plane risk factors than the first landing, which may be better suited to evaluate frontal plane risk factors.

Keywords

Kinematics; Kinetics; Drop vertical jump; ACL injury risk; Knee biomechanics

1. Introduction

Each year in the United States over 120,000 people suffer an anterior cruciate ligament (ACL) injury (Huston et al., 2000). Female athletes are 4 to 6 times more likely to sustain ACL tears than their male counterparts playing similar high risk landing and pivoting sports (Hewett et al., 2005). These injuries are costly and debilitating, as up to 90% of ACL rupture patients exhibit symptoms of early onset arthritis within 10 years of injury (Lohmander and Roos, 1994; Lohmander et al., 2007). Most athletes who sustain ACL ruptures also experience a decrease in quality of life with knee symptoms within 15 years postinjury (Lohmander et al., 2004; von Porat et al., 2004). As costly reconstructive surgeries exhibit no long term benefits towards the reduction of osteoarthritis at the knee (Lohmander and Roos, 1994), injury prevention is likely the best method to reduce the negative consequences of an ACL rupture.

Approximately 70% of ACL injuries occur in non-contact situations as the result of a rapid deceleration or change in direction (Krosshaug et al., 2007; McNair et al., 1990; Myklebust et al., 1998). In regards to basketball, the most commonly reported mechanism of ACL rupture is rebounding, a task that involves a rapid, and often unstable, deceleration as athletes land from a maximal vertical jump (Powell and Barber-Foss, 2000). Jump landings produce high, sudden ground reaction forces that translate into large external torques at the knee that can rupture the ACL (Boden et al., 2000; Hewett et al., 1999). Research with threedimensional motion capture systems has identified a number of mechanical factors that contribute to ACL injury risk during athletic tasks such as excessive knee abduction (Ford et al., 2003; Hewett et al., 2005), knee compression forces (Fleming et al., 2001; Meyer and Haut, 2008), internal tibial rotation (Meyer and Haut, 2008; Shin et al., 2011), and insufficient hip and knee flexion (Chappell and Limpisvasti, 2008; Pollard et al., 2010). The prevalence of these mechanical variables during athletic tasks can be attributed to an athlete's level of neuromuscular control (Hewett et al., 2005). Therefore, training protocols designed to enhance neuromuscular control and target injury risk deficits are effective in altering biomechanics and reducing the incidence of ACL injury within an athletic population (Chappell and Limpisvasti, 2008; Hewett et al., 1999; Pollard et al., 2006).

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One task commonly used to evaluate injury risk biomechanics is the drop vertical jump (DVJ), which simulates the mechanics of rebounding a basketball (Ford et al., 2011; Hewett et al., 2005; Kernozek et al., 2005; Myer et al., 2011; Paterno et al., 2007). The DVJ requires an athlete to drop off a static box, land, immediately execute a maximal vertical jump toward a target, and finish with a second landing. Based on kinematic and kinetic performance traits and anatomical variables, an algorithm has been designed using the DVJ to evaluate an individual's cumulative risk of sustaining an ACL rupture (Myer et al., 2011). This algorithm is designed around the evaluation of a subject's first landing. However, during rebounding tasks, ACL injuries are most often endured as athletes land following a maximal vertical jump to secure the basketball (Powell and Barber-Foss, 2000). Accordingly, the second landing of a DVJ may provide a better simulation of injury risk mechanics.

The objective of this study was to examine the kinetic and kinematic differences between the first and second landings in a DVJ. Anecdotal evidence indicates that athletes display greater lower extremity neuromuscular control deficits during the second landing. Therefore, the hypothesis tested was that lower extremity biomechanical deficits associated with increased ACL injury risk would be greater in the second landing than the first landing. Specifically, we evaluated whether study participants demonstrated greater knee abduction, greater hip adduction, reduced knee and hip flexion and increased side-to-side asymmetry in the second landing relative to the first landing of a DVJ.

2. Methods

This study examined middle (n = 162; age = 12.6 (0.9) years) and high school (n = 77; age = 15.6 (0.9) years) female basketball players from a cohort in a prospective, longitudinal study. Study participants were tested immediately preceding their upcoming season. Procedures were approved by the institutional review board and informed written consent was obtained from the parent or legal guardian of each subject prior to testing. Each subject assented to participation prior to testing.

Participants were evaluated for anatomical measures and maximal vertical jump height prior to motion testing. Height was measured with a stadiometer while the subject stood barefoot (height = 1.60 (0.09) m). To evaluate body mass, participants stood barefoot on a calibrated physician scale (mass = 55.4 (13.2) kg). Shoe size and maximal countermovement vertical jump height were also measured individually for each subject.

For 3-D biomechanical motion analysis, participants wore athletic shorts and tee shirts that were taped in a manner that exposed skin around the greater trochanter of the hip the lower lumbar and abdominal regions and were instrumented with 43 retroreflective markers for 3-D biomechanical analysis. Markers were arranged in a modified Helen Hayes marker set with a backpack (Skeeter CamelBak, Petluma, CA) to define the superior torso and previously marked shoes (Supernova, Adidas, Herzogenaurach, Germany) to standardize footwear (Bates et al., 2013). A static trial was captured to anatomically define each body segment and determine neutral alignment for each subject. 3D motion was collected with a 10-camera motion capture system (Eagle cameras, Motion Analysis Corporation, Santa Rosa, CA) that sampled at 240 Hz. vertical ground reaction force (vGRF) was sampled at 1200 Hz and collected by dual, in-ground, multi-axis force platforms (AMTI, BP600900 Watertown, MA) such that each platform corresponded with a single leg of each subject.

Participants each performed three trials of the DVJ task (Ford et al., 2007). The DVJ began with each subject standing on top of a 31 cm box with feet positioned 35 cm apart and arms held at their sides. The box was aligned such that when a subject dropped straight down from the box, they would land with each foot on a separate force platform. Participants

proceeded to drop straight down from the box and complete a first landing on the force platforms. Upon landing, participants immediately transitioned into a maximal vertical jump toward a provided target, which was followed by a second landing. The provided target was a basketball suspended at the maximal vertical jump height recorded previously for each subject. Prior to execution of the DVJ, participants were instructed to drop straight down from the box without any vertical launch, execute a maximal vertical jump upon contact with the force platforms, and attempt to reach for and bring down the provided target. No specific instructions were provided for the execution of the second landing. If participants failed to land with both feet contained in separate force platforms on the first landing then the trial was repeated. If participants failed to land with both feet contained in separated force platforms on the second landing then the trial was excluded from analysis. Of the 239 participants, 33 failed to complete a successful trial and were excluded.

3-D biomechanical motion data were processed through Visual3D (version 4.0, C-Motion, Inc., Germantown, MD) with custom MATLAB (version 2010b, The Mathworks, Inc., Natick, MA) code for both the first and second landing phases of the DVJ. Landing phase was defined as the moment of initial contact (IC) with the force platform, where the vGRF first exceeded 10 N, through the lowest point of center of gravity during stance (Bates et al., 2013). vGRF data were filtered through a fourth-order, low-pass, digital filter with a cutoff frequency of 100 Hz for vGRF calculations, while marker trajectories and vGRFs were filtered at a cutoff frequency of 12 Hz for kinetic and kinematic calculations (Ford et al., 2010b). For data analysis, each individual subject was represented by the mean of all of her successful trials. All moments were reported as external joint moments derived from the GRFs created during contact with the force platforms.

A 2-by-2 analysis of variance (side: right versus left and landing type: first versus second) examined the relationships between each kinetic and kinematic variable. Post-hoc Student's *t*-tests assessed statistical differences in peak values between the first and the second landing when warranted. *t*-tests were also used to evaluate between landing differences at initial contact and the time point corresponding to the maximal vGRF. Side-to-side asymmetry was assessed through absolute differences in peak values for each kinetic and kinematic variable. Statistical analyses were performed in MATLAB and statistical significance was established a priori at $P < 0.05$.

3. Results

3.1. Sagittal plane

Comparison of sagittal plane kinetic and kinematic values revealed significant side versus landing type interactions at the hip for maximum flexion angle and maximum extension moment (P-values < 0.035). Between landing differences were present in sagittal plane kinetics and kinematics for both the hip and knee. Specifically, the study participants demonstrated reduced peak flexion angles at both joints during the second landing (P-values < 0.001; Table 1). The hip and knee demonstrated less flexion angle during the second landing at IC (P-values< 0.001) and at the time of maximum vGRF (P-values < 0.001). Specifically, the study participants demonstrated 8.4° less knee flexion at the time of IC (Table 3) and 8.5° less knee flexion at the time of maximum vGRF (Table 4). The magnitude of change in flexion angle differences between landings increased when moving proximally up the kinematic chain, as the hip extension was increased by 18.5° at IC and 20.2° at peak vGRF in the second landing. During both landings, maximum flexion angle was achieved faster in both joints during the second landing (P-values < 0.001) (Table 2). Relative to the first landing kinetics, the second landing influenced reduced sagittal plane peak flexion moments at the hip and knee (P-values < 0.001), increased peak extension moments at the hip ($P < 0.001$), and unchanged peak extension moments at the knee ($P =$

0.074). Specifically, participants demonstrated 40.0 N * m less hip flexion, 14.8 N * m less knee flexion, and 18.4 N * mgreater hip extension moments during the second landing. Flexion moments were greater in both the hip and knee at peak vGRF during the second landing (P-values < 0.001), but demonstrated no difference at IC (P = 0.627&0.063).

3.2. Frontal plane

In the frontal plane, there were no significant side versus landing type interactions at either the hip or knee (P-values > 0.050). However, between landing differences were present in hip adduction and knee abduction. While participants demonstrated greater peak hip adduction angle in the first landing (P < 0.001) there were no differences in peak hip adduction moment (P = 0.174; Table 1) between the landings. Peak knee abduction angles and moments were minimally reduced by an average of 1.8° and 1.9 N * m, respectively, in the second landing relative to the first landing (P-values < 0.028). While the magnitude of between landing differences in the frontal plane were similar for both peak angles and moments at hip adduction and knee abduction, these values occurred closer to the time of initial contact during the second landing than the first landing (P-values < 0.001; Table 2). Between landing differences in frontal plane kinetics and kinematics were not present at IC (Table 3) or peak vGRF (P > 0.050; Table 4).

3.3. Asymmetry

Side-to-side asymmetry in terms of absolute differences between peak values generally increased from the first to second landing (Fig. 1). The second landing exhibited increased side-to-side asymmetry for hip sagittal and transverse plane rotation angles, hip sagittal plane and adduction moments, knee flexion angle, and knee sagittal plane and adduction moments (P-values < 0.044). Conversely, the first landing showed increased peak side-to-side differences for hip internal rotation, knee extension, and knee external rotation moment (P-values < 0.019). The average magnitudes of side-to-side asymmetries within each landing are displayed in Table 5.

Ensemble averages of hip and knee moments between landings demonstrated divergent kinematic and kinetic strategies between the primary and secondary landing during the last 20% of landing phase (Fig. 2). While the knee flexion moment was increased during the first landing, knee flexion moment decreased over the same relative time period in the second landing. Similarly divergent patterns were observed in knee abduction, hip flexion, and hip abduction, which led to smaller second landing magnitudes in knee flexion, abduction, and internal rotation moments as well as hip flexion, adduction, and external rotation moments at 100% of landing phase. Divergent behaviors at the end of landing phase were not as apparent in hip and knee flexion angles (Fig. 3).

4. Discussion

The use of the DVJ to screen for injury risk biomechanics and deficiencies in neuromuscular control is well documented (Ford et al., 2011; Hewett et al., 2005; Kernozek et al., 2005; Myer et al., 2011; Paterno et al., 2007). However, previous investigations have focused on the first landing of a DVJ despite that the second landing may more closely resemble the rebounding task responsible for a majority of non-contact ACL ruptures in basketball athletes (Powell and Barber-Foss, 2000). This study examined 3D biomechanics of the second landing in a DVJ in comparison to those of the first landing. The second landing demonstrated higher risk sagittal plane and lower risk frontal plane biomechanics than did the first landing. Specifically, hip and knee flexion and knee abduction were reduced on the second landing.

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Relative to the first landing, hip and knee flexion observed during the second landing of the DVJ were indicative of reduced neuromuscular control and the absorption patterns of a more physically demanding fall. Literature indicates that post pubertal females tend to exhibit less neuromuscular control than their male counterparts (Ford et al., 2010a; Hass et al., 2005; Hewett et al., 2005; Kernozek et al., 2005; Myer et al., 2005). As a result, females execute landings in a more erect position with less hip and knee flexion than males (Decker et al., 2003; Kernozek et al., 2005). Within a female population, athletes who compete at a higher level of sport demonstrate better muscular control and larger magnitudes of knee flexion during landing (Smith et al., 2007). In the present study hip and knee flexion angles and moments decreased from the first to second landing. These between landing differences increased at each joint when moving proximally up the kinetic chain, such that they were most prominent at the proximal joints (i.e. hip).

The present findings indicate that participants landed in a more erect posture during the second landing, which is supported by previous data that found a higher center of mass in the second landing than the first (Bates et al., 2013). Prior literature also demonstrates that the magnitude of flexion incurred when landing decreases relative to increases in drop height, but flexion at initial contact remains the same (Ford et al., 2011; Yeow et al., 2009). However, the drop heights for the first and second landings of our DVJ task are equivalent (Bates et al., 2013). Thus, the second landing presented a maximal flexion pattern representative of a more rigorous landing absorption without an actual increase in drop height. Combined with an unexpected decrease in flexion angles at IC, these data indicate that a different muscle activation pattern and energy absorption technique may be utilized during each landing. The mechanisms enacted in the second landing may result in a better screen for risk factors in the sagittal plane due to a clinician's exacerbated ability to discern neuromuscular deficits during higher level tasks (Ford et al., 2010a).

The magnitude of hip and knee flexion experienced during jump landing has been associated with ACL injury risk (Myers et al., 2011; Paterno et al., 2010; Pollard et al., 2010). Specifically, limited flexion has been linked with increased frontal plane motion and moments that may place athletes at greater risk of ACL injury, while deep flexion is associated with a reduction in valgus torques (Kipp et al., 2011; Pollard et al., 2010). Reduced hip and knee flexion following IC are also indicative of a stiffer joint condition (Myers et al., 2011). Participants who land with stiff joint conditions have been shown to produce greater vGRFs than when landing in a normal condition (DeVita and Skelly, 1992; Myers et al., 2011). Larger ground reaction forces will propagate through the lower extremity and potentially place larger strain on passive restraints such as the ACL, but the exact distribution of these forces remains unknown (DeVita and Skelly, 1992; Myers et al., 2011). It is interesting to note that though the second landing biomechanics demonstrated the reduced hip and knee flexions associated with stiffer joints, the forces incurred between the first and second landings of the DVJ were equivalent (Bates et al., 2013).

One possible mechanism that may have led to the reduced flexion demonstrated in the second landing may be the presence of a quadriceps dominant contraction pattern. Prior studies have demonstrated that females demonstrate larger quadriceps to hamstrings activation ratios than their male counterparts and that these neuromuscular patterns contribute to increases in ACL strain (Ford et al., 2011; Hewett et al., 2008; Padua et al., 2005). The quadriceps serve as the primary knee extensor mechanism and their disproportionate contraction with knee flexors may increase anterior tibial shear force, and therefore ACL load, at flexion angles under 45° (Markolf et al., 1995; Myer et al., 2005). An increase in muscle activation with a knee extensor dominant activation ratio would justify a reduction in joint flexion that increases risk to the ACL in the second landing of a DVJ.

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However, electromyography data was not collected and muscle activation levels were not explicitly examined in this investigation.

Frontal plane dynamics at the hip and knee during landing have been demonstrated to be more accurate predictors of peak landing forces and ACL injury than sagittal measures such as flexion (Hewett et al., 1996, 2005). Specifically, valgus knee loading has been experimentally linked to increases in ACL strain (Markolf et al., 1995) and is considered a primary biomechanical mechanism of non-contact ACL injury (Alentorn-Geli et al., 2009; Hewett et al., 2005). Although not directly linked to ACL injury incidence, hip adduction has been cited as a significant predictor of knee abduction and is described as an intrinsic risk factor (Alentorn-Geli et al., 2009; Imwalle et al., 2009). Therefore, the greater hip adduction and knee abduction magnitudes exhibited in the first landing of a DVJ indicate that it may serve as a better clinical screen tool for frontal plane risk contributors than the second landing. Though the most accurate profile of an athlete's ACL injury risk would be generated by biomechanical data combined from both landings, when assessed individually, the first landing may be a superior overall predictor of ACL injury. As with sagittal plane risk factors, both hip adduction and knee abduction are female biased traits generated from poor neuromuscular control (Ford et al., 2003; Hewett et al., 2005; Kernozek et al., 2005). Also, it is known that females generate maximal hip adduction and knee abduction magnitudes faster than males (Joseph et al., 2011). Therefore, it is likely that decreases in time to peak frontal plane magnitudes may result from poor neuromuscular control throughout landing movements. By this convention, magnitudes of hip and knee frontal plane motions indicated greater ACL injury risk in the first landing, but decreased time to peak values may have signified less muscle coordination in the second landing. Frontal plane dynamics are often targeted by prophylactic interventions as these training protocols have demonstrated the capacity to reduce risk factors related to ACL injury incidence (Chappell and Limpisvasti, 2008; Myer et al., 2006). Athletes properly trained in landing mechanics, such as professional dancers, have even exhibited a lack of gender disparity in lower extremity biomechanics (Orishimo et al., 2009). Therefore, it would be interesting to examine first and second landing differences between a cohort of participants before and after the completion of a neuromuscular intervention program.

Asymmetry in concomitant movements during athletic tasks has been suggested as a measure of neuromuscular control and source of gender disparity in ACL injury rates (Hewett et al., 2005). Female athletes exhibit larger side-to-side kinematic asymmetries during bilateral tasks than their male counterparts (Ford et al., 2003; Pappas and Carpes, 2012). The literature further documents that contralateral asymmetries in the lower extremities increase following an ACL injury and predispose an athlete towards further ACL ruptures (Paterno et al., 2007, 2011). Specifically, asymmetry in hip rotation has been correlated to an increase in knee abduction, which may be a primary mechanism in ACL injury (Hewett et al., 2005; Howard et al., 2011). This study found between landing differences in contralateral asymmetry for sagittal and transverse plane rotations at the hip. Unlike previous findings, the greater hip rotational asymmetry during the second landing did not affect an increase in knee abduction. However, the lack of increase in knee abduction demonstrated in this study could be justified by a lack of change in frontal plane rotational asymmetry at the hip. Decreased excursion and few asymmetrical behaviors insinuate that the frontal plane experienced enhanced neuromuscular control during the second landing than the first landing. It is possible to experience greater control under more rigorous motions as frontal plane asymmetry during landing decreases when fall height is raised from 0.2 or 0.4 m to 0.6 m (Ford et al., 2011). Meanwhile, the sagittal plane continued to demonstrate increased risk of injury as both kinematic and kinetic asymmetry increased from the first to second landing.

Potential limitations to the current study include that the first and second landings of the DVJ are not identical tasks. The first landing is a drop jump task as participants landed and proceeded to load up for a maximal vertical leap, while the second landing is a drop land task as participants landed, stabilized, and resumed a normal standing posture. Drop jumps and drop lands from equivalent heights elicit comparable vGRFs (Bates et al., 2013), but demonstrate different muscle activation patterns and place unique neuromuscular demand on the lower extremity (Ambegaonkar et al., 2011). Kinematic and kinetic differences between a drop land and drop jump remain largely uninvestigated, but it has been shown that the drop jump task is characterized by higher level of activity in the medial and lateral quadriceps, hamstrings, and lateral gastrocnemius muscles than the drop landing (Ambegaonkar et al., 2011). This variation in muscle activation can account for differences observed in several hip and knee kinetic and kinematic variables at the end of the landing phase. Also, exclusionary criteria of this study removed a large number of trials due to a failed second landing; however, 206 participants completed successful trials and provided sufficient power for analysis.

5. Conclusions

The second landing exhibited decreased flexion angles in all joints as was hypothesized. The second landing showed a lower maximum hip adduction angle and knee abduction moment and angle, but no difference in hip adduction moment. Differences between landings indicated that reduced hip and knee flexion in the second landing may provide a superior evaluation of neuromuscular control in the sagittal plane relative to the first landing of a DVJ, while greater knee abduction in the first landing may provide a superior evaluation in the frontal plane. Overall, the second landing demonstrated lower extremity mechanics that were characteristic of a higher rigor athletic task than the first landing.

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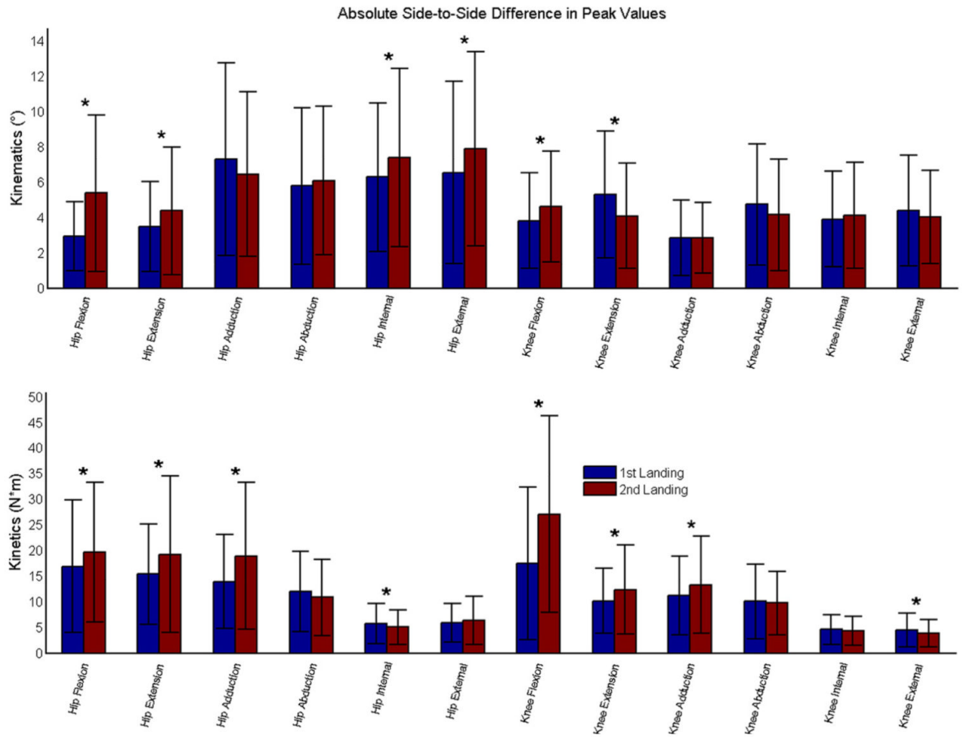

Fig. 1.

Displays the mean absolute magnitude of side-to-side differences plus standard deviation in peak values for kinematic and kinetic variables at the hip and knee. *Indicates significant differences between landings.

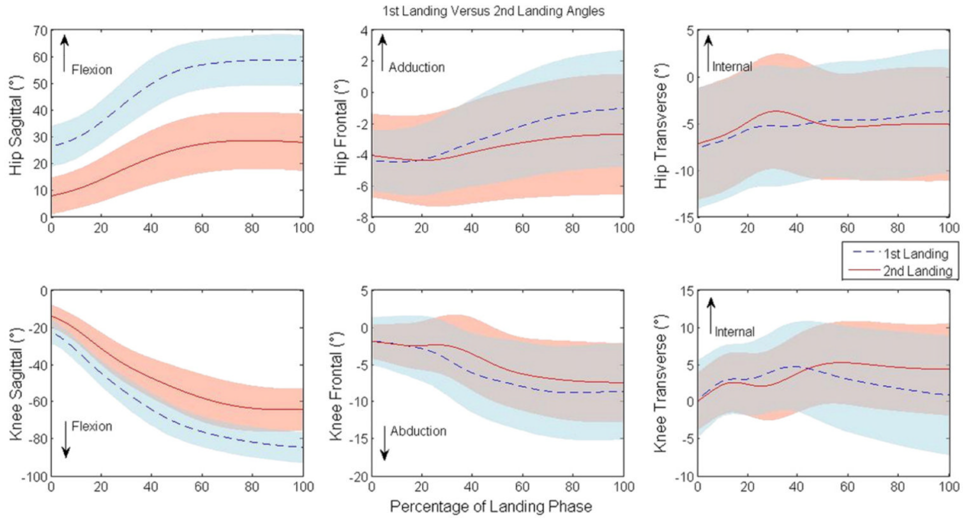

Fig. 2.

Displays the ensemble average kinematics plus standard deviation for all three planes of rotation at the hip and knee. Data is separated by first and second landings and time normalized from 0–100% of the landing phase.

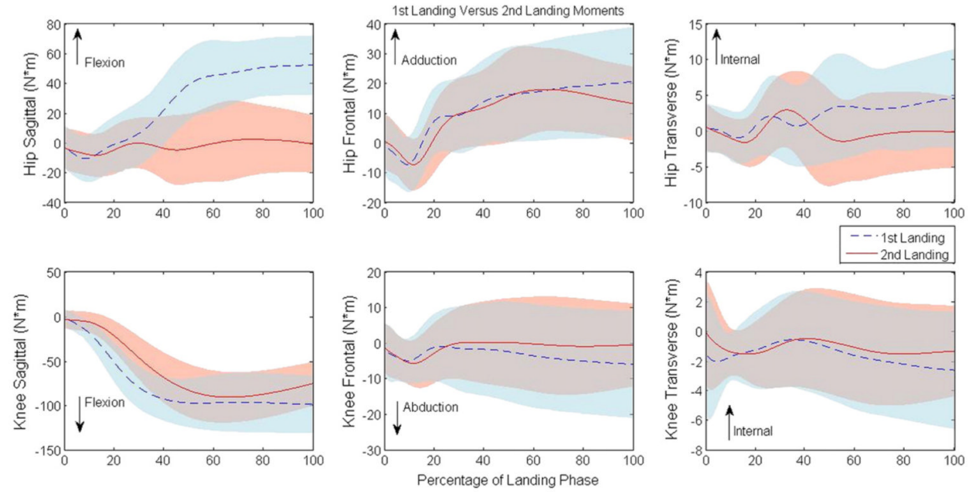

Fig. 3.

Displays the ensemble average external moments plus standard deviation for all three planes of rotation at the hip and knee. Data is separated by first and second landings and time normalized from 0–100% of the landing phase.

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Table 1

Displays the mean peak kinematic and kinetic values at the hip and knee for both legs during the first and second landings.

	Flexion	Extension	Abduction	Adduction	Internal	External
Hip angle						
Right 1st landing (°)	59.8 (9.4)	25.8 (8.1)	−5.9 (4.3)	0.2 (5.4)	−1.2 (7.3)	−11.7 (7.5)
Left 1st landing (°)	60.1 (9.5)	26.8 (7.9)	−5.9 (4.7)	0.1 (5.7)	0.8 (7.1)	−9.3 (6.9)
Mean 1st landing (°)	59.9 (9.3)	26.3 (7.7)	−5.9 (2.6)	0.2 (3.2)	−0.2 (6.2)	−10.5 (6.0)
Right 2nd landing (°)	29.6 (10.9)[a]	7.1 (7.3)[a]	−5.8 (4.9)	−0.8 (4.9)[a]	−1.6 (6.7)	−12.1 (6.8)
Left 2nd landing (°)	31.8 (11.3)[a]	8.7 (7.4)[a]	−6.1 (5.0)	−1.2 (5.2)[a]	1.6 (7.3)	−8.6 (7.4)
Mean 2nd landing (°)	30.7 (10.7)[a]	7.9 (6.9)[a]	−5.9 (3.4)	−1.0 (3.2)[a]	0.0 (5.7)	−10.3 (5.6)
Hip moment						
Right 1st landing (N * m)	69.8 (24.1)	−23.9 (19.1)	−11.3 (11.2)	34.7 (18.2)	12.1 (6.8)	−9.3 (5.6)
Left 1st landing (N * m)	68.0 (22.6)	−25.8 (19.1)	−16.6 (10.7)	28.6 (19.4)	12.3 (6.3)	−7.7 (6.3)
Mean 1st landing (N * m)	68.9 (20.9)	−24.8 (17.1)	−13.9 (9.2)	31.6 (17.2)	12.2 (5.7)	−8.5 (5.0)
Right 2nd landing (N * m)	27.8 (20.2)[a]	−35.8 (23.9)[a]	−12.3 (9.9)	31.6 (17.4)	10.4 (5.5)[a]	−11.1 (6.2)[a]
Left 2nd landing (N * m)	29.9 (20.6)[a]	−35.2 (23.7)[a]	−16.5 (9.8)	27.4 (19.1)	10.6 (5.9)[a]	−9.7 (6.7)[a]
Mean 2nd landing (N * m)	28.9 (17.2)[a]	−35.5 (20.8)[a]	−14.4 (8.2)	29.5 (14.8)	10.5 (5.0)[a]	−10.4 (5.4)[a]
Knee angle						
Right 1st landing (°)	−83.8 (8.5)	−21.6 (7.2)	−11.7 (6.7)	−1.1 (4.6)	7.5 (6.1)	−3.3 (6.8)
Left 1st landing (°)	−85.1 (8.6)	−23.6 (7.2)	−10.3 (6.1)	−0.5 (4.4)	7.2 (5.9)	−2.9 (6.4)
Mean 1st landing (°)	−84.4 (8.3)	−22.6 (6.6)	−11.0 (5.8)	−0.8 (4.2)	7.4 (5.5)	−3.1 (6.0)
Right 2nd landing (°)	−65.2 (11.5)[a]	−13.4 (6.6)[a]	−9.7 (5.8)[a]	−0.6 (3.9)	7.8 (5.7)	−2.4 (4.9)
Left 2nd landing (°)	−66.6 (11.4)[a]	−14.9 (6.9)[a]	−8.6 (5.1)[a]	−0.2 (3.7)	8.2 (5.2)	−1.5 (4.7)[a]
Mean 2nd landing (°)	−65.9 (11.2)[a]	−14.2 (6.3)[a]	−9.2 (4.9)[a]	−0.4 (3.4)	8.0 (4.9)	−2.0 (4.2)[a]
Knee Moment						
Right 1st landing (N * m)	−111.6 (31.8)	1.2 (10.8)	−15.4 (11.1)	12.2 (11.3)	4.5 (3.9)	−6.3 (3.8)
Left 1st landing (N * m)	−114.0 (32.4)	−1.3 (10.6)	−17.8 (10.7)	8.1 (12.1)	2.5 (3.5)	−7.7 (4.0)
Mean 1st landing (N * m)	−112.8 (30.1)	−0.1 (9.4)	−16.6 (9.4)	10.1 (10.1)	3.5 (2.9)	−7.0 (3.1)
Right 2nd landing (N * m)	−96.5 (31.2)[a]	2.3 (12.6)	−13.3 (9.2)[a]	12.8 (12.0)	4.5 (3.8)	−5.1 (3.1)[a]
Left 2nd landing (N * m)	−99.4 (33.0)[a]	0.9 (11.0)[a]	−16.2 (9.4)	10.2 (12.3)	3.1 (3.7)	−6.0 (3.1)[a]

	Flexion	Extension	Abduction	Adduction	Internal	External
Mean 2nd landing (N * m)	−98 (28.4)[a]	1.6 (9.7)	−14.7 (7.9)[a]	11.5 (9.7)	3.8 (3.1)	−5.5 (2.4)[a]

[a]Indicates a statistically significant difference in 2nd landing value from corresponding value during 1st landing.

Table 2

Displays the mean times when peak kinematic and kinetic values occurred during landing phase. Times are displayed in seconds relative to IC and reported separately by leg side and first and second landings.

	Flexion	Extension	Abduction	Adduction	Internal	External
Hip angle						
Right 1st land	0.166 (0.053)	0.003 (0.003)	0.059 (0.063)	0.136 (0.071)	0.117 (0.066)	0.069 (0.057)
Left 1st land	0.168 (0.050)	0.003 (0.003)	0.057 (0.056)	0.142 (0.070)	0.121 (0.064)	0.068 (0.060)
Mean 1st land	0.167 (0.051)	0.003 (0.003)	0.058 (0.038)	0.139 (0.051)	0.119 (0.054)	0.069 (0.044)
Right 2nd land	0.132 (0.030)[a]	0.003 (0.007)	0.062 (0.051)	0.098 (0.058)[a]	0.090 (0.041)[a]	0.085 (0.055)[a]
Left 2nd land	0.131 (0.029)	0.004 (0.015)	0.065 (0.052)	0.093 (0.058)	0.094 (0.042)	0.075 (0.058)
Mean 2nd land	0.132 (0.028)[a]	0.004 (0.009)	0.064 (0.033)	0.096 (0.039)[a]	0.092 (0.031)[a]	0.080 (0.045)[a]
Hip moment						
Right 1st land	0.143 (0.050)	0.034 (0.022)	0.033 (0.039)	0.106 (0.049)	0.105 (0.052)	0.072 (0.040)
Left 1st land	0.142 (0.048)	0.036 (0.024)	0.036 (0.044)	0.118 (0.048)	0.112 (0.051)	0.069 (0.047)
Mean 1st land	0.142 (0.044)	0.035 (0.018)	0.034 (0.033)	0.112 (0.038)	0.109 (0.043)	0.070 (0.033)
Right 2nd land	0.083 (0.042)[a]	0.072 (0.038)[a]	0.032 (0.029)	0.090 (0.029)[a]	0.073 (0.037)[a]	0.082 (0.034)[a]
Left 2nd land	0.093 (0.039)	0.067 (0.035)	0.029 (0.025)	0.093 (0.034)	0.077 (0.031)	0.079 (0.037)
Mean 2nd land	0.088 (0.033)[a]	0.070 (0.030)[a]	0.031 (0.020)	0.092 (0.024)[a]	0.075 (0.025)[a]	0.081 (0.027)[a]
Knee angle						
Right 1st land	0.198 (0.043)	0.002 (0.000)	0.146 (0.037)	0.035 (0.051)	0.076 (0.045)	0.089 (0.087)
Left 1st land	0.196 (0.043)	0.002 (0.000)	0.146 (0.033)	0.036 (0.048)	0.079 (0.045)	0.080 (0.084)
Mean 1st land	0.197 (0.042)	0.002 (0.000)	0.146 (0.029)	0.035 (0.042)	0.078 (0.038)	0.085 (0.074)
Right 2nd land	0.148 (0.030)[a]	0.003 (0.011)	0.135 (0.042)[a]	0.036 (0.033)	0.089 (0.039)[a]	0.049 (0.059)[a]
Left 2nd land	0.147 (0.028)[a]	0.002 (0.011)	0.131 (0.043)[a]	0.040 (0.032)	0.097 (0.043)[a]	0.042 (0.050)[a]
Mean 2nd land	0.147 (0.028)[a]	0.002 (0.011)	0.133 (0.038)[a]	0.038 (0.027)	0.093 (0.035)[a]	0.046 (0.046)[a]
Knee moment						
Right 1st land	0.121 (0.038)	0.005 (0.005)	0.090 (0.072)	0.055 (0.051)	0.062 (0.057)	0.091 (0.068)
Left 1st land	0.126 (0.041)	0.004 (0.004)	0.079 (0.069)	0.064 (0.055)	0.069 (0.055)	0.079 (0.069)
Mean 1st land	0.124 (0.033)	0.004 (0.004)	0.085 (0.056)	0.059 (0.045)	0.066 (0.044)	0.085 (0.051)
Right 2nd land	0.109 (0.019)[a]	0.011 (0.009)	0.059 (0.047)[a]	0.054 (0.041)	0.046 (0.037)[a]	0.072 (0.051)[a]

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	Flexion	Extension	Abduction	Adduction	Internal	External
Left 2nd land	0.109 (0.019)[a]	0.012 (0.009)	0.051 (0.044)[a]	0.069 (0.049)	0.062 (0.048)	0.059 (0.047)[a]
Mean 2nd land	0.109 (0.017)[a]	0.011 (0.008)[a]	0.055 (0.036)[a]	0.061 (0.035)	0.054 (0.033)[a]	0.066 (0.039)[a]

[a]ndicates a statistically significant difference in 2nd landing value from corresponding value during 1st landing.

Table 3

Displays mean kinematic and kinetic values at the hip and knee at IC separated by leg side and landing.

	Flex/ext	Ab/add	Int/ex
Hip angle			
Right 1st landing (°)	25.9 (8.1)	−4.4 (3.3)	−8.8 (7.9)
Left 1st landing (°)	26.8 (7.9)	−4.4 (3.6)	−6.6 (7.5)
Mean 1st landing (°)	26.4 (7.7)	−4.4 (2.0)	−7.7 (6.4)
Right 2nd landing (°)	7.1 (7.3)[a]	−3.8 (4.0)	−8.4 (7.4)
Left 2nd landing (°)	8.7 (7.4)[a]	−4.3 (4.2)	−6.0 (7.5)
Mean 2nd landing (°)	7.9 (6.9)[a]	−4.1 (2.7)	−7.2 (5.9)
Hip moment			
Right 1st landing (N * m)	−1.4 (17.3)	2.5 (12.8)	0.7 (4.4)
Left 1st landing (N * m)	−3.1 (16.7)	−3.9 (11.2)	0.1 (4.0)
Mean 1st landing (N * m)	−2.3 (14.5)	−0.7 (9.7)	0.4 (3.4)
Right 2nd landing (N * m)	−2.1 (17.6)	3.9 (11.4)	1.3 (4.4)
Left 2nd landing (N * m)	−3.8 (16.3)	−2.5 (11.4)	−0.1 (3.9)
Mean 2nd landing (N * m)	−2.9 (13.7)	0.7 (9.0)	0.6 (3.2)
Knee angle			
Right 1st landing (°)	−21.6 (7.2)	−2.2 (3.6)	0.5 (5.9)
Left 1st landing (°)	−23.6 (7.2)	−1.7 (3.4)	0.4 (5.7)
Mean 1st landing (°)	−22.6 (6.6)	−1.9 (3.2)	0.4 (5.2)
Right 2nd landing (°)	−13.4 (6.5)[a]	−2.0 (2.6)	−0.3 (4.5)
Left 2nd landing (°)	−14.9 (6.8)[a]	−1.8 (2.4)	0.5 (4.7)
Mean 2nd landing (°)	−14.2 (6.3)[a]	−1.9 (2.3)	0.1 (3.9)
Knee moment			
Right 1st landing (N * m)	−0.2 (12.1)	1.4 (10.0)	−0.2 (5.6)
Left 1st landing (N * m)	−2.2 (11.3)	−4.6 (8.4)	−2.5 (4.6)
Mean 1st landing (N * m)	−1.2 (9.9)	−1.6 (7.1)	−1.3 (4.0)
Right 2nd landing (N * m)	−2.1 (13.5)	1.7 (9.2)	1.0 (4.7)[a]
Left 2nd landing (N * m)	−3.9 (12.2)	−4.2 (8.7)	−1.1 (4.4)[a]
Mean 2nd landing (N * m)	−3.0 (10.2)	−1.2 (6.8)	0.0 (3.5)[a]

[a]Indicates a statistically significant difference in 2nd landing value from corresponding value during 1st landing.

Table 4

Displays mean kinematic and kinetic values at the hip and knee at maximum vGRF separated by leg side and landing.

	Flex/ext	Ab/add	Int/ex
Hip angle			
Right 1st landing (°)	38.8 (9.7)	−3.7 (4.3)	−7.0 (7.5)
Left 1st landing (°)	41.2 (10.3)	−3.9 (4.9)	−4.9 (7.0)
Mean 1st landing (°)	40.8 (9.9)	−3.8 (2.8)	−5.8 (6.2)
Right 2nd landing (°)	16.5 (10.8)[a]	−4.0 (4.5)	−6.0 (7.2)
Left 2nd landing (°)	19.4 (11.7)[a]	−4.3 (5.1)	−2.8 (7.9)[a]
Mean 2nd landing (°)	18.2 (10.9)[a]	−4.1 (3.2)	−4.5 (5.9)[a]
Hip moment			
Right 1st landing (N * m)	−0.2 (26.0)	14.6 (16.7)	1.3 (6.9)
Left 1st landing (N * m)	1.6 (26.5)	5.7 (16.4)	2.3 (6.0)
Mean 1st landing (N * m)	3.1 (25.5)	10.7 (13.9)	1.8 (5.1)
Right 2nd landing (N * m)	−4.6 (20.5)	8.5 (14.7)[a]	0.8 (5.9)
Left 2nd landing (N * m)	−0.6 (21.1)	6.8 (18.8)	1.3 (6.7)
Mean 2nd landing (N * m)	−3.0 (19.1)[a]	7.8 (14.9)[a]	'0.8 (5.7)
Knee angle			
Right 1st landing (°)	−49.9 (11.4)	−4.5 (6.0)	3.6 (6.1)
Left 1st landing (°)	−53.7 (12.2)	−4.0 (5.4)	3.8 (6.1)
Mean 1st landing (°)	−53 (11.0)	−4.3 (5.3)	3.7 (5.6)
Right 2nd landing (°)	−37.3 (13.4)[a]	−3.3 (5.0)[a]	2.4 (5.4)[a]
Left 2nd landing (°)	−40.7 (14.2)[a]	−2.9 (4.7)[a]	2.8 (5.4)
Mean 2nd landing (°)	−39.7 (13.3)[a]	−3.2 (4.4)[a]	2.7 (4.9)
Knee moment			
Right 1st landing (N * m)	−73.6 (32.1)	−0.2 (14.5)	0.0 (4.2)
Left 1st landing (N * m)	−80.1 (33.6)	−3.7 (15.1)	−1.7 (4.4)
Mean 1st landing (N * m)	−78.8 (29.5)	−2.0 (12.6)	−0.9 (3.5)
Right 2nd landing (N * m)	−40.3 (31.4)[a]	−0.2 (12.0)	−0.7 (3.2)
Left 2nd landing (N * m)	−45.9 (35.4)[a]	−2.1 (12.3)	−1.5 (3.6)
Mean 2nd landing (N * m)	−45.1 (33.6)[a]	−1.4 (10)	−1.2 (2.8)

[a]Indicates a statistically significant difference in 2nd landing value from corresponding value during 1st landing.

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Table 5

Displays the mean magnitude of kinematic and kinetic side-to-side asymmetries for both the first and second landings.

	1st landing	2nd landing
Joint angle		
Hip min flexion (°)	2.9 (2.0)	5.4 (4.4)[a]
Hip max flexion (°)	3.5 (2.5)	4.4 (3.6)[a]
Hip adduction (°)	7.3 (5.5)	6.5 (4.7)
Hip abduction (°)	5.8 (4.4)	6.1 (4.2)
Hip internal (°)	6.3 (4.2)	7.4 (5.1)[a]
Hip external (°)	6.6 (5.2)	7.9 (5.5)[a]
Knee min flexion (°)	3.8 (2.7)	4.6 (3.1)[a]
Knee max flexion (°)	5.3 (3.6)	4.1 (3.0)[a]
Knee adduction (°)	2.9 (2.2)	2.8 (2.0)
Knee abduction (°)	4.8 (3.4)	4.2 (3.2)
Knee internal (°)	3.9 (2.7)	4.1 (3.0)
Knee external (°)	4.4 (3.2)	4.0 (2.6)
Joint moment		
Hip extension (N * m)	16.9 (12.9)	19.7 (13.7)[a]
Hip flexion (N * m)	15.4 (9.8)	19.3 (15.3)[a]
Hip adduction (N * m)	13.9 (9.2)	18.9 (14.4)[a]
Hip abduction (N * m)	12.0 (7.8)	10.8 (7.5)
Hip internal (N * m)	5.7 (3.9)	5.0 (3.3)[a]
Hip external (N * m)	5.9 (3.8)	6.4 (4.7)
Knee extension (N * m)	17.4 (14.9)	14.9 (19.2)[a]
Knee flexion (N * m)	10.2 (6.4)	6.4 (8.7)[a]
Knee adduction (N * m)	11.1 (7.7)	7.7 (9.5)[a]
Knee abduction (N * m)	10.1 (7.3)	7.3 (6.2)
Knee internal (N * m)	4.6 (2.9)	2.9 (2.8)
Knee external (N * m)	4.5 (3.3)	3.3 (2.6)[a]

[a]Indicates a statistically significant difference in 2nd landing value from corresponding value during 1st landing.