Peak knee biomechanics and limb symmetry following unilateral anterior cruciate ligament reconstruction: Associations of walking gait and jump-landing outcomes

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A R T I C L E I N F O

Keywords:
Vertical ground reaction force
Posttraumatic
Biomechanics
ACL injury

A B S T R A C T

Background: Aberrant walking-gait and jump-landing biomechanics may influence the development of post-traumatic osteoarthritis and increase the risk of a second anterior cruciate ligament injury, respectively. It remains unknown if individuals who demonstrate altered walking-gait biomechanics demonstrate similar altered biomechanics during jump-landing. Our aim was to determine associations in peak knee biomechanics and limb-symmetry indices between walking-gait and jump-landing tasks in individuals with a unilateral anterior cruciate ligament reconstruction.

Methods: Thirty-five individuals (74% women, 22.1 [3.4] years old, 25 [3.89] kg/m^2) with an anterior cruciate ligament reconstruction performed 5-trials of self-selected walking-gait and jump-landing. Peak kinetics and kinematics were extracted from the first 50% of stance phase during walking-gait and first 100 ms following ground contact for jump-landing. Pearson product-moment (r) and Spearman's Rho (ρ) analyses were used to evaluate relationships between outcome measures. Significance was set a priori ($P \leq 0.05$).

Findings: All associations between walking-gait and jump-landing for the involved limb, along with the majority of associations for limb-symmetry indices and the uninvolved limb, were negligible and non-statistically significant. There were weak significant associations for instantaneous loading rate ($\rho = 0.39$, $P = 0.02$) and peak knee abduction angle ($\rho = 0.36$, $p = 0.03$) uninvolved limb, as well as peak abduction displacement limb-symmetry indices ($\rho = -0.39$, $p = 0.02$) between walking-gait and jump-landing.

Interpretation: No systematic associations were found between walking-gait and jump-landing biomechanics for either limb or limb-symmetry indices in people with unilateral anterior cruciate ligament reconstruction. Individuals with an anterior cruciate ligament reconstruction who demonstrate high-involved limb loading or asymmetries during jump-landing may not demonstrate similar biomechanics during walking-gait.

1. Introduction

Individuals who have sustained an anterior cruciate ligament (ACL) injury and undergo surgical reconstruction (ACLR) demonstrate aberrant lower extremity biomechanics during activities of daily living such as walking gait (Blackburn et al., 2016; Butler et al., 2009; Gardinier et al., 2013; Hall et al., 2012; Pietrosimone et al., 2016b; Shi et al., 2010; Webster et al., 2012a) and dynamic movements such as jump-landing (Castanharo et al., 2011; Delahunt et al., 2012; Mohammadi et al., 2012; Paterno et al., 2007; Paterno et al., 2010) compared to healthy, uninjured individuals. Approximately 30% of these individuals sustain a second ACL injury within 24 months of the initial injury (Paterno et al., 2014), while 45–50% of individuals develop knee osteoarthritis (OA) within 10–15 years of the initial injury (Butler et al., 2009; Luc et al., 2014). Aberrant biomechanics have been implicated as contributors to the increased risk of early onset post-traumatic

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https://doi.org/10.1016/j.clinbiomech.2018.01.020
Received 23 June 2017; Accepted 22 January 2018

osteoarthritis (PTOA) (Butler et al., 2009; Hall et al., 2012; Luc et al., 2014; Palmieri-Smith & Thomas, 2009; Pietrosimone et al., 2016b; Roos, 2005) and secondary ACL injury (Di Stasi et al., 2013a; Leys et al., 2012; Paterno et al., 2010; Salmon et al., 2005; Shelbourne et al., 2009). However, it is unknown if individuals with an ACLR who demonstrate gait biomechanics that may increase the risk of PTOA are the same people who demonstrate pathomechanics that may increase the risk of sustaining a second ACL injury.

Individuals with an ACLR demonstrate biomechanical asymmetries during walking gait (Blackburn et al., 2016; Logerstedt et al., 2013; Rohman et al., 2015; Webster et al., 2012a) and this repetitive over or under-loading of articular tissues in the ACLR limb may influence deleterious joint tissue changes that may increase the risk of future PTOA onset (Butler et al., 2009; Khandha et al., 2016). Similarly, individuals with an ACLR also demonstrate greater inter-limb asymmetries during dynamic functional tests (single-leg hop, single-leg triple hop, crossover triple hop) compared to healthy controls (Rohman et al., 2015), which may increase the risk of sustaining a secondary ACL injury (Paterno et al., 2007). Previous studies have separately reported that individuals with an ACLR demonstrate greater peak vertical ground reaction force (vGRF) magnitudes and loading rates (vGRF-LR) in the injured limb compared to the uninjured limb and healthy controls during both walking gait (Blackburn et al., 2016; Noehren et al., 2013) and jump-landing (Palmieri-Smith & Thomas, 2009; Paterno et al., 2007; Paterno et al., 2010). Separate studies have reported that individuals with an ACLR also demonstrate alterations in peak sagittal and frontal plane knee kinematics compared to healthy controls during walking gait (Butler et al., 2009; Coury et al., 2006; Gokeler et al., 2010; Noehren et al., 2013) and jump-landing (Di Stasi et al., 2013b; Paterno et al., 2010; Risberg et al., 2009). There is evidence that peak biomechanics are altered and often present asymmetrically during jump-landing as well as walking in separate ACLR cohorts; yet, it remains unclear if individuals who demonstrate greater inter-limb asymmetries during walking gait are the same individuals who demonstrate greater inter-limb asymmetries during jump-landing.

PTOA development has been linked to biomechanical alterations during repetitive cyclical movements (Andriacchi & Favre, 2014), such as walking gait. Alterations in mechanical loading, as well as frontal and sagittal knee kinematics, may shift contact forces during walking to portions of the articular cartilage that are unaccustomed to such loads (Andriacchi & Mundermann, 2006), thereby potentially accelerating PTOA development. Similarly, subsequent ACL injuries occur following more frequent exposure to high velocity forces during dynamic movements (Paterno et al., 2010), such as jump-landing (Paterno et al., 2014; Paterno et al., 2010), which occur less frequently than forces experienced during gait but are associated with greater peak kinetics and different lower extremity kinematics. Therefore, it is important to know if higher peak loading, altered knee kinematics, and biomechanical asymmetries exist during both walking gait and jump-landing, as this knowledge may be essential for developing effective comprehensive screening methods that can detect the risk of sustaining an acute injury as well as developing a chronic joint condition following ACLR. Additionally, understanding the pathomechanics that transfer from one task to another will inform the development of future rehabilitation programs seeking to develop safe movement strategies to minimize the risk of future acute and chronic knee injury following ACLR. The primary purpose of this study was to determine if individuals with a unilateral ACLR, who demonstrate greater peak kinematic and kinetic magnitudes in the ACLR and uninjured limb during walking gait also demonstrate greater peak kinematic and kinetic magnitudes in each limb during jump-landing. Additionally, we will determine if those who demonstrate greater kinematic and kinetic asymmetries during walking gait also demonstrate greater asymmetries during jump-landing using limb symmetry indices (LSI). We hypothesized that individuals with greater peak kinematics and kinetics in the ACLR and uninjured limbs, as well as more asymmetric biomechanics, during walking gait would demonstrate the greater peak biomechanics during jump-landing in both the injured and uninjured limbs.

2. Methods

2.1. Study design

We conducted a cross-sectional study and all outcome measures were collected during a single testing session. Data of jump-landing biomechanics were always collected after walking gait biomechanics as to not impact walking biomechanics due to possible fatigue caused by jump-landing. An identical set of 12 biomechanical measurements were collected during the walking gait and jump-landing tasks. We evaluated 7 kinematic (peak knee flexion angle, displacement, and velocity, peak knee adduction angle and displacement, and peak knee abduction angle and displacement) and 5 kinetic measures (vGRF peak magnitude, instantaneous [INST-LR] and linear [LIN-LR] loading rates, peak internal knee extension moment, and peak internal knee abduction moments) in the involved and uninvolved limbs. LSI (ACLR limb/Uninjured Limb) for all kinetic and kinematic outcome measures were calculated for analysis. We chose these specific outcome measures as each has been reported to differ between limbs or compared to controls in either jump-landing or walking gait (Bjornaraa & Di Fabio, 2011; Blackburn et al., 2016; Butler et al., 2009; Coury et al., 2006; Delahunt et al., 2012; Di Stasi et al., 2013a; Gokeler et al., 2010; Noehren et al., 2013; Palmieri-Smith & Thomas, 2009; Paterno et al., 2007; Paterno et al., 2010; Webster et al., 2012b). Written informed consent was obtained from all participants before data collection, and the Institutional Review Board at the University of North Carolina at Chapel Hill approved all the study methods and recruitment procedures.

2.2. Participants

We recruited individuals with a history of a primary unilateral ACLR in the university community using word of mouth, flyers and electronic mail. All individuals included in the study had undergone an ACLR > 6 months prior to the time of data collection and had been cleared by an orthopedic surgeon for unrestricted participation in physical activity. All individuals reported engaging in a minimum of 20 min of moderate physical activity three times per week. Individuals with a history of any lower extremity orthopedic surgery other than ACLR, multi-ligament reconstruction to the ACLR knee, ACLR revision surgery, bilateral ACLR, diagnosed knee osteoarthritis, balance or neuromuscular disorders, or an orthopedic injury in either leg during the prior 6 months were excluded. All participants were asked to self-report age, sex, ACL graft type, if a concomitant surgical meniscal procedure (menisectomy or meniscal repair) was performed at the time of ACLR, and the date of ACLR. Participants completed the subjective section of the International Knee Documentation Committee (IKDC) form to determine self-reported disability at the time of testing (Ebrahimzadeh et al., 2015; Higgins et al., 2007). Additionally, participants were instructed to indicate their current activity level and activity level at the time of ACL injury via the Tegner (Briggs et al., 2009) and Marx (Negahban et al., 2011) activity rating scales at the being of the testing session. Using an exact two-tailed bivariate correlational model (G*Power v.3.1.9.2), we estimated that we would need 29 participants in order to identify a statistically significant moderate association ($r = 0.5$) between biomechanical outcomes during walking gait and jump-landing with an alpha level of 0.05, and a 1-β of 0.8. We elected to recruit and evaluate outcomes in 35 participants in order to account for potential outliers or increase variability during analysis and to account for any participant dropout during the recruitment process.

2.3. Motion capture

Three-dimensional kinematic data were sampled at 120 Hz using a ten-camera motion capture system (Vicon, Nexus, Oxford, UK) and lowpass filtered at 10 Hz (4th order recursive Butterworth) (Pietrosimone et al., 2016b). Ground reaction forces were sampled at 1200 Hz from three embedded force plates (FP406010, Bertec Corp) centered in an 891.82 m^3 motion capture volume and low-pass filtered at 75 Hz (4th order recursive Butterworth). Participants performed both walking gait and jump-landing tasks while wearing uniform tight-fitting spandex shorts supplied by the laboratory and were outfitted with 25 retroreflective markers (Pietrosimone et al., 2016b). A cluster of 3 markers was placed over the sacrum in order to estimate the anatomical positioning of the pelvic girdle. A static trial was captured while the participant stood with arms positioned at 90 degrees of abduction to estimate the location of the landmarks needed to calculate joint centers. The medial condyle and malleolus markers were removed during data collection to ensure medial knee markers would not contact each other, or influence the participants' usual movements during the walking and jump-landing trials. Knee and ankle joint centers were defined as the midpoint between the medial and lateral condyles and malleoli, respectively. The hip joint center was estimated from the coordinates of the L4-5, right anterior superior iliac spine (ASIS), and left ASIS markers using the Bell method (Cappozzo et al., 1995). Joint angles were defined based on the position of the distal segment relative to the proximal segment using the Euler method (Kidder et al., 1996) with the following planes of rotational motion: sagittal (y-axis), frontal (x-axis), and transverse (z-axis).

2.4. Walking gait analysis

The current study was part of a larger study that also sought to measure specific outcomes related to walking gait (i.e. characteristics of the heel strike transient) (Blackburn et al., 2016), which are distinguishable during barefoot walking. Therefore, all participants in this cohort completed all walking trials barefoot. During all walking gait trials, participants were instructed to walk at a self-selected speed over 2 force plates embedded in a staggered formation towards the middle of a 6 m walkway so that the entire stance phase for both limbs could be collected during a single trial (Pietrosimone et al., 2016b). Before testing, participants performed 5 practice trials through 2 infrared timing gates (TF100, TracTronix) in order to determine average walking speed which served as a baseline for consistency of walking speed over the 5 data collection trials for each limb (Pietrosimone et al., 2016b). During data collection, participants performed 5 acceptable walking gait trials that required: 1) both right and left feet individually striking and toeing off a single force plate; 2) maintaining forward eye contact and not aiming for the force plates; 3) maintaining a consistent walking gait speed within 5% of the average speed determined during the practice trials; and 4) no visibly altered walking gait during the trial (e.g., trip or stutterstep) (Pietrosimone et al., 2016a).

2.5. Jump-landing analysis

All participants wore their own athletic footwear for the jump-landing trials. Participants performed jump-landing from a 30 cm box positioned 50% of the participant's height from the front edge of the force plates (Padua et al., 2009). Participants were instructed to jump forward off the box to a double-leg landing with one foot on each force plate, and immediately jump vertically as high as possible (Padua et al., 2009). Each participant performed a minimum of three practice trials until investigators were comfortable that the participants were properly performing the jump-landing task. A minimum of 60 s separated each trial, yet participants were allowed as much rest time as needed between each jump-landing trial in order to ensure that the subsequent trials could be completed with maximal effort. After each participant vocalized that they received an adequate amount of rest after performing a trial, subsequent trials were performed. A total of five jump-landing trials were collected. A successful jump-landing trial required the participant to leave the box with both feet at the same time, land on the force plates, and perform a subsequent vertical jump. In the event of an unsuccessful trial, a subsequent trial was utilized for analysis.

2.6. Data collection and analysis

The stance phase for walking gait was defined as the interval from heel strike at ground contact (vGRF $>$ 20 N) to toe-off (vGRF $<$ 20 N). Similarly, the loading phase of jump-landing was defined as the first 100 ms following ground contact (vGRF $>$ 20 N), which has been reported to be the time period in which an ACL injury would most commonly occur (Krosshaug et al., 2007). All walking gait biomechanics were extracted from the first 50% of the stance phase of gait. Previous studies (Pietrosimone et al., 2016a; Pietrosimone et al., 2016b) have demonstrated that lower extremity biomechanics collected during the first 50% of the stance phase of gait associate with cartilage metabolism in individuals with an ACLR. Additionally, all jump-landing biomechanics were extracted during the first 100 ms following ground contact. All data were averaged across the 5 trials for the walking gait and jump-landing tasks. The instantaneous vGRF loading rate (vGRF-LR) was calculated as the peak of the first derivative of the force-time curve. Peak vGRF (BW) and vGRF-LR (BW/s) were normalized to body weight. Peak knee flexion was defined as maximum flexion angle and knee flexion excursion was calculated as the difference between the peak value and the value at ground contact. Knee joint moments were calculated using an inverse dynamics approach (Lariviere & Gagnon, 1999). The Knee Extension Moment (KEM) and knee abduction moment (KAM) were normalized to the product of height and body weight (BW $*$ m) and expressed as negative values by angular conventions. LSI was calculated for all outcomes (ACLR limb/uninjured limb) such that values $<$ 1.0 denoted lesser values on the ACLR limb while those $>$ 1.0 denoted greater values on the ACLR limb compared to the uninjured limb.

2.7. Statistical analysis

All outcomes were assessed for normality using Shapiro-Wilk tests. Associations between each primary outcome for walking gait and jump-landing were evaluated via two-tailed bivariate associations separately for the injured and uninjured limbs. We performed secondary two-tailed bivariate associations between LSI during walking gait and jump-landing. Pearson product-moment correlations (r) were used to evaluate normally distributed outcomes, while Spearman's rank order correlations (ρ) were used to evaluate associations that included one or more non-normally distributed variables. All associations were described as negligible (0.0–0.29), low (0.30–0.49), moderate (0.50–0.69), high (0.70–0.89), and very high (0.9–1.0) (Mukaka, 2012). In addition to our planned analyses, we conducted post hoc Pearson product-moment and Spearman's rank order correlation analyses to determine if primary and secondary walking gait biomechanics associated with jump-landing biomechanics through the entire loading phase (i.e. ground contact through full peak knee flexion). Furthermore, we evaluated these bivariate associations separately for males and females. Alpha levels were set a priori 0.05 for all analyses, which were performed using the Statistical Package for the Social Sciences (SPSS) software, version 21.0 (IBM Corp. Armonk, NY).

3. Results

3.1. Participant demographics

Thirty-five individuals with a unilateral ACLR (74% female, 168.4 [10.67] cm, 71.17 [16.07] kg, Table 1) completed a single data collection session during which kinetic and kinematic outcome measures for walking gait and jump-landing were acquired (Tables 2 & 3). The participants had undergone ACLR an average of 49.65 [40.62] months

Table 1
Demographics.

	Mean (SD)		
	Entire Cohort	Men	Women
Participants	35	9	26
Height (cm)	168 (11.01)	179.44 (9.7)	164.58 (8.1)
Mass (kg)	71.17 (16.07)	86.31 (21.31)	65.93 (9.74)
BMI (kg/m^2)	25 (3.89)	26.65 (4.46)	24.44 (3.6)
Tegner current	6.97 (1.9)	7.11 (1.54)	6.97 (1.9)
IKDC (%)	86.64 (9.78)	89.45 (10.75)	85.67 (9.45)
Months post-ACLR	49.65 (40.62)	53.38 (42.98)	48.5 (40.68)

(BMI: Body Mass Index, IKDC: International Knee Documentation Committee).

Table 2
Outcome measure during walking gait.

	Injured Limb	Uninjured	LSI
Peak vGRF (BW)	1.10 [0.072]	1.10 [0.08]	1.00 [0.01]
Inst. loading rate (BW/s)	55.80 [16.4]	51.27 [12.56]	1.04 (0.28)
Linear loading rate (BW/s)	6.76 (1.44)	6.92 [1.34]	1.01 [0.02]
Peak flexion angle (degrees)	10.48 [6.00]	10.93 [6.46]	0.98 (0.42)
Flexion displacement (degrees)	11.70 [3.60]	11.85 [3.74]	0.97 (0.22)
Peak flexion velocity (degrees)	149.84 [41.10]	157.84 (46.58)	0.99 (0.22)
Peak extension moment (N/m)	−0.041 [0.014]	−0.04 [0.02]	0.96 (0.39)
Peak knee adduction angle (degrees)	−0.64 [2.81]	−0.51 (4.77)	0.48 (1.56)
Peak knee abduction angle (degrees)	−0.64 [2.81]	−0.51 (4.77)	0.48 (1.56)
Peak knee adduction displacement (degrees)	1.41 [0.75]	1.44 [0.93]	1.05 (1.69)
Peak knee abduction displacement (degrees)	1.41 [0.75]	1.44 [0.93]	1.05 (1.69)
Peak knee abduction moment (N/m)	0.024 (0.01)	0.03 [0.01]	0.89 (0.60)

Values reported as mean [standard deviation] for normal distribution or median (interquartile range) for non-normal distribution.

(range = 9–161 months) prior to testing. The majority of participants reported their ACL to have been reconstructed with a patellar tendon autograft (n = 26, 74%) and 9 participants reported that a semitendinosus/gracilis autograft was used for ACLR. Sixteen participants (45.7%) reported a history of a concomitant meniscal injury, whereas 17 participants (48.6%) reported no concomitant meniscal injury and 2 participants (5.7%) were uncertain if they sustained a concomitant meniscal injury.

3.2. Associations for injured and uninjured limbs

There were no significant associations found between any walking gait or jump-landing biomechanics for the injured limb (Table 4). For the uninjured limb, individuals with higher instantaneous loading rate (INST-LR) during walking gait demonstrated higher INST-LR during jump-landing (ρ = 0.39, P = 0.02) and individuals with a greater peak knee abduction angle during walking gait demonstrated a greater peak knee abduction angle during jump-landing (ρ = 0.36, P = 0.03). There were no other associations found in any of the other primary outcome measures for the uninjured limb. The strength of the positive correlations for the non-significant associations ranged from negligible to low (66.7% negligible, 33.3% low, Table 4). The negative correlations for the non-significant associations were all classified as negligible.

3.3. Associations for Limb Symmetry Index

Individuals with greater peak knee abduction displacement LSI during walking gait demonstrated lesser peak knee abduction displacement LSI during jump-landing (ρ = − 0.39, p = 0.02). There were no other significant associations found in any of the other secondary (LSI) outcome measures. The strength of the non-significant correlations between biomechanical outcomes for walking gait and jump-landing ranged from negligible to low (75% negligible, 25% low, Table 4). The strength of the non-statistically significant correlations between biomechanical outcomes for walking gait and jump-landing ranged from negligible to low (75% negligible, 25% low, Table 4).

3.4. Primary post hoc analyses: Associations between gait and jump–landing biomechanics throughout the entire loading phase

Greater INST-LR (ρ = 0.39, P = 0.02) and peak knee abduction angle (ρ = 0.35, P = 0.04) during walking gait was associated with INST-LR and peak knee abduction angle during jump-landing when measured through the entire loading phase of jump-landing. There was not a significant association for peak knee abduction displacement LSI (ρ = −0.08, P = 0.64) between walking gait and jump-landing in this post hoc analysis. All other outcomes showed no significant associations and the strength of the associations, between walking gait and jump-landing, were low or negligible (27.8% low, 72.2% negligible, Table 4).

3.5. Secondary post hoc analyses: Difference in associations between gait and jump–landing biomechanics separated by gender

For females, greater uninjured INST-LR (ρ = 0.435, P = 0.026), uninjured LIN-LR (ρ = 0.547, P = 0.004), and lesser INST-LR LSI (ρ = −0.404, P = 0.04) during walking gait was significantly associated with INST-LR, LIN-LR, and INST-LR LSI during jump-landing. For males, greater uninjured peak knee extension moment (r = 0.753,

Table 3
Outcome measure during first 100 ms of jump-landing.

	Injured Limb	Uninjured	LSI
Peak vGRF (BW)	2.18 (0.97)	2.49 (0.838)	0.88 [0.22]
Inst. loading rate (BW)	182.87 [56.31]	180.12 (119.27)	0.90 [0.23]
Linear loading rate (BW)	61.22 (34.47)	62.16 (35.05)	1.01 [0.33]
Peak flexion angle (degrees)	76.48 (15.69)	75.72 [11.91]	1.06 [0.21]
Flexion displacement (degrees)	57.48 [13.77]	55.80 [11.17]	1.10 (0.61)
Peak flexion velocity (degrees)	620.91 [49.03]	624.64 [55.39]	0.99 [0.08]
Peak extension moment (N/m)	−0.16 [0.03]	−0.18 [0.031]	0.88 (0.21)
Peak knee adduction angle (degrees)	3.55 [8.23]	4.18 [7.86]	0.28 (1.55)
Peak knee abduction angle (degrees)	1.71 [4.61]	1.31 [4.62]	0.67 (3.67)
Peak knee adduction displacement (degrees)	2.03 (6.70)	2.21 (5.68)	0.07 (1.45)
Peak knee abduction displacement (degrees)	−0.10 (1.56)	0.00 (0.53)	0.85 (1.12)
Peak knee abduction moment (N/m)	−0.03 (0.02)	−0.02 (0.04)	0.67 (3.67)

Values reported as mean [standard deviation] for normal distribution or median (interquartile range) for non-normal distribution.

Table 4
Correlations between walking gait and jump-landing at 100 ms.

	Injured Limb	Uninjured	LSI
Peak vGRF	ρ = −0.22 (0.21)	ρ = −0.04 (0.84)	*r* = −0.01 (0.99)
Inst. loading rate	*r* = 0.14 (0.26)	ρ = 0.39 (0.02)*	ρ = −0.27 (0.11)
Linear loading rate	ρ = 0.05 (0.76)	ρ = 0.23 (0.21)	*r* = −0.15 (0.38)
Peak flexion angle	ρ = 0.25 (0.20)	*r* = 0.19 (0.23)	ρ= −0.08 (0.63)
Flexion displacement	ρ = 0.06 (0.73)	ρ = 0.13 (0.26)	ρ = 0.40 (0.15)
Peak flexion velocity	*r* = 0.34 (0.17)	ρ = 0.06 (0.32)	*r* = −0.05(0.72)
Peak extension moment	*r* = −0.02(0.92)	r = 0.34 (0.17)	ρ = 0.19 (0.23)
Peak knee adduction angle (degrees)	ρ= −0.07 (0.69)	ρ = −0.10 (0.58)	ρ= −0.15 (0.40)
Peak knee abduction angle (degrees)	ρ = 0.26 (0.13)	ρ = 0.36 (0.03)*	ρ = 0.21 (0.22)
Peak knee adduction displacement (degrees)	ρ = 0.21 (0.24)	ρ = −0.20 (0.25)	ρ = 0.01 (0.98)
Peak knee abduction displacement (degrees)	ρ = 0.25 (0.09)	ρ = 0.22 (0.21)	ρ = −0.39 (0.02)*
Peak knee abduction moment (N/m)	ρ = 0.15 (0.38)	ρ = 0.22 (0.21)	ρ = −0.17 (0.33)

Spearman's rho: ρ, Pearson Product Moment Correlation: r, (*P* value), * indicates significance.

Table 5
Correlations between walking gait and jump-landing at 100 ms for female and male participants.

	Injured Limb	Uninjured Limb	LSI
Female participants (n = 26)			
Peak vGRF	ρ= −0.16 (0.43)	ρ = −0.01 (0.99)	r = −0.02 (0.91)
Inst. loading rate	*r* = 0.37 (0.07)	ρ = 0.44 (0.03)*	ρ = −0.40 (0.04)*
Linear loading rate	ρ = 0.24 (0.23)	ρ = 0.55 (0.01)*	*r* = −0.14 (0.48)
Peak flexion angle	ρ = 0.19 (0.35)	*r* = 0.28 (0.17)	ρ= −0.17 (0.40)
Flexion displacement	ρ = 0.14 (0.50)	ρ = 0.15 (0.47)	ρ = 0.02 (0.94)
Peak flexion velocity	*r* = 0.12 (0.55)	ρ = 0.38 (0.06)	*r* = −0.19 (0.35)
Peak extension moment	r = −0.19 (0.34)	*r* = −0.13 (0.54)	ρ = 0.34 (0.09)
Peak knee adduction angle (degrees)	ρ= −0.04 (0.86)	ρ = −0.03 (0.88)	ρ = 0.05 (0.80)
Peak knee abduction angle (degrees)	ρ = 0.37 (0.06)	ρ = 0.36 (0.07)	ρ = 0.05 (0.83)
Peak knee adduction displacement (degrees)	ρ = 0.19 (0.34)	ρ = −0.26 (0.20)	ρ = −0.18 (0.48)
Peak knee abduction displacement (degrees)	ρ = 0.26 (0.21)	ρ = 0.29 (0.15)	ρ = −0.47 (0.06)
Peak knee abduction moment (N/m)	ρ = −0.09 (0.67)	ρ = 0.35 (0.08)	ρ = −0.33 (0.10)
Male participants (*n* = 9)			
Peak vGRF	ρ = −0.38 (0.31)	ρ = −0.17 (0.67)	*r* = −0.22 (0.57)
Inst. loading rate	*r* = 0.07 (0.85)	ρ = 0.32 (0.41)	ρ = 0.27 (0.49)
Linear loading rate	ρ = −0.12 (0.77)	ρ = −0.10 (0.80)	*r* = −0.18 (0.65)
Peak flexion angle	ρ = 0.17 (0.67)	r = 0.12 (0.77)	ρ = 0.07 (0.87)
Flexion displacement	ρ = −0.18 (0.64)	ρ = 0.55 (0.13)	ρ = 0.43 (0.24)
Peak flexion velocity	*r* = 0.29 (0.44)	ρ = −0.05 (0.90)	*r* = 0.52 (0.15)
Peak extension moment	r = −0.22(0.57)	r = 0.75 (0.02)*	ρ = −0.03 (0.93)
Peak knee adduction angle (degrees)	ρ = −0.05 (0.90)	ρ = −0.37 (0.33)	ρ = −0.55 (0.13)
Peak knee abduction angle (degrees)	ρ = −0.40 (0.29)	ρ = −0.32 (0.41)	ρ = 0.28 (0.46)
Peak knee adduction displacement (degrees)	ρ = 0.35 (0.36)	ρ = −0.15 (0.70)	ρ = 0.31 (0.54)
Peak knee abduction displacement (degrees)	ρ = 0.33 (0.39)	ρ = 0.12 (0.76)	ρ = −0.17 (0.67)
Peak knee abduction moment (N/m)	ρ = 0.15 (0.70)	ρ = −0.05 (0.90)	ρ = −0.42 (0.27)

Spearman's rho: ρ, Pearson Product Moment Correlation: r, (P value), * indicates significance.

$P = 0.019$) significantly associated with greater uninjured peak knee extension moment during jump-landing. All other associations between walking gait and jump-landing, for both females and males, were not statistically significant (Table 5).

4. Discussion

The primary finding of this study was that there was no consistent association for peak knee kinematics and kinetics, as well as kinematic and kinetic asymmetries between walking gait and jump-landing tasks in individuals with a unilateral ACLR. We found 3 significant correlations suggesting only a weak positive association between walking gait and jump-landing for uninjured INST-LR, uninjured peak knee abduction angle, and peak knee abduction displacement-LSI during walking gait and during jump-landing. No other significant associations were found for knee biomechanics between jump-landing and gait. The results of this study suggest that higher magnitude of loading, as well as asymmetrical loading of the knee, is task dependent following unilateral ACLR. Contrary to our hypothesis, it should not be assumed that individuals with an ACLR who demonstrate biomechanics that may increase risk of a future ACL during a jump – landing tasks will be the same people that may demonstrate biomechanics that may increase the risk for developing PTOA, and vise versa.

Jump-landing and walking gait are movements that are organized differently by the central nervous system, which may be one possible explanation for why magnitudes and asymmetries in knee biomechanics and lower extremity loading may not be similar. Rhythmic movements, such as walking gait, are traditionally hypothesized to be governed by central pattern generators (CPG), which are neural networks located in the spinal cord (MacKay-Lyons, 2002). Conversely, jump-landing is a non-repetitive movement that requires more supra-spinal or cortical-level control (Winters & Crago, 2000). During jump-landing, feed-forward neuromuscular control is needed to activate musculature to prepare for contact with the ground (Winters & Crago, 2000). ACL injury and ACLR may have different effects on neuromuscular pathways that coordinate movements that preferentially rely on supra-spinal control (jump-landing) compared to CPG movements (walking). Walking gait and jump-landing also place uniquely different demands on the nervous system and the neuromuscular effects of ACL injury and ACLR may be unique for different individuals. The differing patterns of altered biomechanics between these two activities, as seen in this study, suggest that a single test is insufficient to screen for generalized movement deficiencies or aberrant mechanics that may be present in all movements. Therefore, assessments of various types of movements, both dynamic and rhythmical, may be necessary for determining individuals who are at risk for developing PTOA and sustaining subsequent acute lower extremity injuries.

Movements such as walking gait and jump-landing exert stress to different tissues within the ACLR limb. Repetitive rhythmic movements such as walking gait apply compressive force on the articular cartilage of the tibiofemoral joint (Radin et al., 1991). Following ACLR, alterations in the magnitudes and rates of loading in the ACLR limb during walking gait may lead to greater cumulative compressive force and increased stress on the articular cartilage (Radin et al., 1991). Individuals with an ACLR demonstrate greater knee adduction moments (KAM) during walking gait compared to healthy controls (Butler et al., 2009), which increases the mechanical loads on the articular cartilage of the medial compartment and may hasten the development of PTOA. Dynamic movements such as jump-landing exert stress upon the articular cartilage (Souza et al., 2012) as well as greater loads on the ACL (Berns et al., 1992; Markolf et al., 1995). Aberrant landing mechanics observed following ACLR (Bell et al., 2014; Escamilla et al., 2012; Gokeler et al., 2010) may lead to increased tensile forces placed upon the reconstructed ACL and a heightened risk for subsequent injuries. While alterations in both walking gait and jump-landing are commonly observed in individuals with an ACLR, the it is possible that the results from our study suggest different individuals may seek to reduce the stresses placed upon different knee joint tissues by individual altering their movement strategies in individual tasks. Future research should seek to determine the mechanisms that lead to changes in movements of multiple tasks.

Dynamic movements, including various hopping tasks, are utilized to make progression decisions through a rehabilitation program and determine criterion for returning to sport (Adams et al., 2012; Rudolph et al., 2000). It is hypothesized that hopping or jumping related tasks could be used to monitor landing kinematics and visualize aberrant movement strategies that could increase the risk for a subsequent ACL injury. However, there is not a standardized assessment of repetitive movements, such as walking gait, within the rehabilitation process and these movements are not used in determining if the individual has passed the criteria for returning to sport. The lack of evaluation of walking gait during rehabilitation may prevent the detection of altered walking biomechanics that may lead to early development of PTOA. Therefore, there may be a need for a battery of clinical tests evaluating dynamic and repetitive movements to simultaneously determine the multifactorial risk of both PTOA and subsequent ACL injuries following ACLR. In order to reduce the risk of subsequent acute ACL injuries and the development of PTOA, rehabilitation that is focused on addressing altered walking gait and jump-landing biomechanics separately is required. Traditional rehabilitation protocols focus on returning individuals to physical activity following ACLR by correcting aberrant biomechanics that manifest in movements that occur during both activities of daily living (i.e. walking) and dynamic sport specific movements (i.e. jump-landing). Multiple forms of rehabilitation specifically designed to address alterations in dynamic voluntary movements, as well as rhythmical movements may be needed in order to improve the deleterious biomechanical patterns of these separate types of tasks, following ACLR surgery.

4.1. Study limitations

While the results of this study provide insight into future research needed in this area, our findings should be examined within the context of some limitations. The participants in this study demonstrated a wide range of time after surgery (9 months to 161 months). This large range may have allowed for participants to be at different stages of pre-radiographic PTOA development, which may contribute to alterations in biomechanical movement patterns. Future studies should strive for a narrower range of post-surgical time periods within participants. Specific jump-landing LSI outcomes, such as peak adduction angle LSI and peak adduction displacement LSI, were found to be non-normally distributed and have median LSI values that were significantly lower than 1.00. The magnitudes of these measures are often small (Table 3) and a one or two-degree difference between limbs may influence the LSI to a greater extent than other biomechanical outcomes with greater absolute magnitudes. All individuals in this study sustained a unilateral ACL injury and subsequent ACLR, but future studies with larger sample sizes should seek to examine patients with bilateral ACL injury, as well as how graft type and concomitant meniscal injury influence the associations between walking and jump-landing biomechanics. Additionally, we performed post-hoc analyses to investigate how gender influenced associations between walking gait and jump-landing biomechanics. The majority of our participants (n = 26, 74.2%) were female, therefore future studies with larger sample sizes may be needed to comprehensively determine how gender influences the associations between gait and jump-landing biomechanics following ACLR. The primary focus of this study was on understanding knee-related biomechanics as well as measures related to the vGRF during jump-landing and walking. Future studies that also evaluate the hip and ankle joints biomechanics during both tasks may provide more insight into how individuals adapt to performing dynamic movements and activities of daily living following ACLR. The cross-sectional nature of this study limited our ability to directly relate specific peak kinetic and kinematic variables to the risk of developing PTOA or a subsequent ACL injury within the cohort that was studied. However, similar gait biomechanics as measured in this study (i.e. peak vGRF, INST-LR, peak knee adduction moment), are associated with deleterious metabolic outcomes that that may be related to cartilage breakdown (Miyazaki et al., 2002; Pietrosimone et al., 2016a; Pietrosimone et al., 2017; Pietrosimone et al., 2016b). Similarly, Paterno et al (Paterno et al., 2010) identified peak knee abduction angle and sagittal-plane knee moment LSI as biomechanical predictors for a subsequent ACL injury. Longitudinal studies are needed to better understand how altered peak knee biomechanics during walking and jump-landing following ACLR influence the risk for radiographic PTOA development and a second ACL injury.

5. Conclusion

We did not find a consistent association for peak knee biomechanics and inter-limb asymmetries between walking gait and jump-landing tasks in individuals with a unilateral ACLR. There were only three weak significant associations for INST-LR and peak knee abduction angle in the uninjured limb and peak knee abduction displacement-LSI between walking gait and jump-landing. The current study provides evidence that biomechanical alterations may be different between tasks in people with unilateral ACLR. These findings suggest that biomechanics of walking gait and jump-landing need to be assessed separately to evaluate biomechanical alterations that may increase the risk of developing PTOA or sustaining a second ACL injury. Additionally, rehabilitation of movement may need to be more task specific to simultaneously decrease the risk of PTOA and a secondary ACL injury.

Acknowledgements

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

Conflict of interest

No authors have any conflicts of interest to report.

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