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Assessing lower extremity coordination and coordination variability in individuals with anterior cruciate ligament reconstruction during walking

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ABSTRACT

Background: Despite our knowledge of several biomechanical risk factors related to anterior cruciate ligament (ACL) injury, such as decreased knee flexion, increased knee abduction, and increased hip flexion, adduction and internal rotation during walking, jogging, and landing from a jump, the incidence of ACL tears remains high. Quantifying variability in the lower extremity provides a continuous measure of joint coordination and function that may elicit an additional aspect of ACL injury mechanisms. Research question: The aim of this study was to assess joint coordination patterns and variability in individuals following ACL reconstruction (ACLR).

Methods: Twenty participants with unilateral ACLR and twenty uninjured participants matched by sex and body mass index (BMI) walked over-ground at self-selected speed. Two force plates embedded in the walking platform recorded ground reaction forces (GRF), and a motion capture system collected kinematic data. Vector coding was used to describe coordination patterns and measure coordination variability in hip-knee and knee-ankle coupled motion. Results: Individuals with ACLR had greater variability in hip-knee coordination compared to their healthy counterparts for both the reconstructed and contralateral limbs. The individuals with ACLR also exhibited altered coordination patterns, one of which was characterized by constrained hip motion.

Significance: These results are evidence that differences in joint coordination exist between individuals with and without ACLR, even after the former are cleared to return to sport. This new insight into coordinative function after ACLR may be useful for improving rehabilitation strategies as well as identifying those at risk of injury during return to sport testing.

1. Introduction

The knee is the most prevalently injured joint in adolescent athletes [1], and the foremost cause for sport-related surgeries [2]. An estimated 200,000 anterior cruciate ligament (ACL) ruptures occur each year in the United States alone, for which surgical reconstruction (ACLR) is the standard treatment to restore joint stability and function. Unfortunately, Ardern et al. [3] reported only 82% of patients who underwent ACLR were capable of resuming their previous activities, and little more than half of those who returned to sport regained their pre-injury level of performance. There is also a significant risk of developing osteoarthritis of the knee after incurring an ACL injury [4], and the incidence of ipsilateral or contralateral injury is six times greater in individuals who have a surgically repaired ACL [5]. Further, there is a reported $2 billion annual healthcare cost associated with ACL injury [6].

Researchers have reported altered biomechanics in individuals with ACLR that include decreased knee flexion [7], increased knee abduction [7–10], increased hip adduction [11,12], flexion [8] and internal rotation [9], and increased ankle eversion [13] relative to healthy individuals during tasks like walking, running, and landing from a jump. While these differences are evidence that kinematic function may not be restored after ACLR, most variables are limited to discrete time points within a movement cycle and only consider single joint function. It is known that the lower extremity acts as a linked system, and that the mechanism(s) for ACL injury involve dynamic movement. Therefore, a continuous measure of joint coordination may improve our understanding of the biomechanical factors related to ACL injury.

An alternative method based on dynamical systems theory (DST) examines the interaction between two joints or segments. DST proposes that a healthy motor system has redundant degrees of freedom (DOF)

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that provide multiple pathways to perform a task and are controlled by coordinative structures. Accordingly, Bernstein [14] defined coordination as the process of mastering redundant DOF to produce a controllable system. Variability in coordination is a result of the infinite number of combinations employed by available DOF to complete a task, and is possibly an essential component of movement that allows for stability and flexibility. Investigators have hypothesized that there is an optimal amount of variability present in any motor system that differentiates between the ability to adjust to environmental instability and the risk for injury.

Reduced coordinative variability has been associated with orthopedic disorders [15]. Hamill et al. [16] measured variability in lower extremity motion using continuous relative phase (CRP), a method that derives the phase angle of a segment or joint from its position-velocity curve. The difference between two phase angles gives the CRP angle and is used as a measure of coordination. Both pattern and magnitude differences were observed in individuals with patellofemoral pain (PFP), with symptomatic individuals displaying less variability compared to asymptomatic individuals during treadmill running. Therefore, a decrease in variability was indicative of a pathological system. Despite different methodology, this finding was supported by Heiderscheit et al. [17] who assessed joint coordination variability using vector coding. Individuals with PFP displayed less variability during treadmill running compared to healthy participants, providing additional evidence that lower variability could suggest pathology. While both CRP and vector coding produced similar conclusions regarding reduced coordination variability in a PFP population during running, these methods are not equivalent. CRP considers velocity data while vector coding uses only the position signals, which may make vector coding more translatable for clinical interpretation.

The goal of this study was to quantify and compare joint coordination patterns and variability in individuals with ACLR and uninjured matched controls during walking using vector coding. Joint coordination was assessed as it relates to the timing of joint actions [18], which may be used to guide rehabilitation. Specifically, hip-knee couplings in the sagittal, frontal, and transverse planes related to movement patterns associated with ACL injury mechanisms and adaptive strategies post-ACLR were examined. Knee-ankle coupling in the sagittal plane was also examined to determine if the ankle contributed to gait alterations after ACLR. Based on a review of the literature that revealed decreased variability in pathological systems [16,17,19–21], within-participant variability for all hip-knee couplings was hypothesized to be lower in the ACLR group. Consistent with the idea that the healthy components of a movement system will adapt their function to compensate for dysfunction at an injured site, the ACLR group was expected to demonstrate altered coordination phase patterns compared to the control group. Gribbin et al. [22] examined differences in the hip-knee joint coupling during gait after ACLR and reported larger coupling angles in individuals with ACLR compared to healthy participants. This was interpreted as the hip having decreased contribution to joint movement relative to the knee. The authors suspected that the hip acted as a stabilizer in individuals with ACLR as a response to increased knee motion or instability. Therefore, coupling angles were hypothesized to be increased in the ACLR group compared to those in the control group such that the hip would have a smaller excursion, indicating an effort to increase stability.

2. Participants

A total of 20 participants (n = 9 female; body mass index (BMI) 25 ± 3.5 kg/m^2) who had undergone unilateral ACLR (13 right) and been cleared to return to full activity were compared to 20 control participants matched by gender, BMI (n = 9 female; BMI 22.4 ± 2.4 kg/m^2), and side of the ACL reconstruction such that the same side for the control subject was matched to the side of the ACL reconstruction in the experimental group. Type of physical therapy was unknown and type of surgery was not recorded for all participants. Participants were on average 7 years post-surgery, but this information was not available for all participants.

3. Methods

Institutional Review Board (IRB) approval was obtained prior to data collection, and all participants signed an informed consent form. Kinetic and kinematic data were collected from all 40 participants during walking in the University of Tennessee Health Science Center Motion Analysis Laboratory using an opto-electronic motion capture system (Qualisys AB, Goteborg, Sweden) synchronized with force plates embedded in a walking platform (AMTI, Watertown, MA). Retroreflective markers were placed over bony landmarks to define joint locations and attached in rigid clusters of four for segment tracking as previously published [23]. A static trial was captured for which the participant stood in a neutral position in the center of a force plate (AMTI, Watertown, MA). Subsequently, motion trials were captured as the participant walked at self-selected speed (ACLR: 1.3 ± 0.18 m/s, Control: 1.2 ± 0.07 m/s) across the walking platform. Ground reaction force (GRF) data was collected to identify specific gait events. A gait trial was considered successful if the foot made full contact with at least one force plate without the participant targeting the force plate. A minimum of three successful trials for each limb were collected for all participants.

Visual3D (C-Motion, Germantown, MD, USA) was used to create a six DOF skeletal model consisting of seven rigid segments (CODA pelvis, thighs, shanks, and feet). Kinematic data were filtered using a low-pass fourth-order Butterworth filter with a cutoff frequency of 7 Hz. Force data were filtered using a low-pass fourth-order Butterworth filter with a cutoff frequency of 15 Hz. Only the stance phase of the gait cycle was analyzed because it includes the interval during which ACL injury frequently occurs (loading response) and common gait adaptations are observed following ACLR. Stance was defined as the period from initial contact (IC) to toe off, and a 15 N threshold was assigned to the GRF curve to identify these two events. Joint angles for the hip, knee, and ankle for flexion-extension, abduction-adduction, and internal-external rotation were calculated according to the Cardan rotation sequence XYZ.

A vector coding technique [24,25] was used to quantify joint coordination variability. Using a custom script in Matlab (Mathworks, Massachusetts, USA) an angle-angle curve was created such that the angular displacements of the proximal and distal joints were plotted on the x- and y-axes, respectively. A coupling angle (CA) was calculated as the angle between two adjacent data points on the angle-angle curve relative to the right horizontal, expressed by Eq. (1). The CA was calculated at each percentage of stance (i) for each trial (m).

$$CA = tan^{-1}\left(\frac{y_{m,i+1} - y_{m,i}}{x_{m,i+1} - x_{m,i}}\right) \quad (1)$$

The CA was calculated for each joint coupling in both limbs, and averaged in four periods of stance (described later) across three trials for each participant using circular statistics (Eqs. (2)–(4)). The circular average of the CA ranged between 0° and 360°, and was used to preserve the direction of joint movement that would be lost if the data were compressed to 0°-180° [26].

$$\bar{x}_i = \frac{1}{n}\Sigma\, cos\,(CA_{m,i}) \quad (2)$$

$$\bar{y}_i = \frac{1}{n}\Sigma\, sin\,(CA_{m,i}) \quad (3)$$

$$\bar{CA} = \begin{cases} arctan\left(\frac{\bar{y}_i}{\bar{x}_i}\right) \\ arctan\left(\frac{\bar{y}_i}{\bar{x}_i}\right) + 180°,\ when\ \bar{x}_i < 0 \end{cases} \quad (4)$$

Table 1
Joint coordination variability in the reconstructed and matched limb for ACLR versus Control.

Coupling	Period	Variability (SD) (Mean ± Standard Deviation)		p-Value	Cohen's d
		ACLR	Control		
HA/KR	1	9.2 ± 5.4	6.3 ± 3.0	0.083	0.68
HA/KA	1	7.3 ± 7.0	4.5 ± 2.8	0.123	0.52
HR/KR	1	17.0 ± 9.5	10.8 ± 6.9	0.017	0.76
HA/KR	2	28.4 ± 17.9	17.6 ± 17.6	0.048	0.61
HA/KA	2	24.7 ± 18.4	15.1 ± 12.1	0.114	0.62
HA/KA	3	8.0 ± 6.3	4.8 ± 3.2	0.097	0.64
HR/KA	3	21.5 ± 20.2	9.0 ± 6.0	0.017	0.84
HR/KR	3	22.5 ± 17.9	14.1 ± 12.0	0.114	0.56

Note: P1: HS to initial GRF peak; P2: initial GRF peak to second GRF peak; P3: Second GRF peak to half the time to reach TO.

The circular standard deviation of the mean CA was calculated in each stance period and represented within-participant variability. Participant standard deviations were averaged across the ACLR and control groups for each joint coupling in both limbs. Cohen's d was also evaluated and defined as small (d < 0.3), moderate (d = 0.3-0.5), or large (d < 0.5).

Six joint couplings were selected for analysis based on kinematic differences between ACLR and uninjured populations reported in the literature: hip abduction-adduction/knee abduction-adduction (HA/KA), hip abduction-adduction/knee rotation (HA/KR), hip flexion-extension/knee flexion-extension (HF/KF), hip rotation/knee abduction-adduction (HR/KA), hip rotation/knee rotation (HR/KR), and knee flexion-extension/ankle dorsiflexion-plantarflexion (KF/ADF). The stance phase was further divided into four periods based on discrete events in the vertical GRF (vGRF) curve [27]. This was done to account for the changing functional demands (loading, weight acceptance, and propulsion) throughout this portion of the gait cycle. Initial loading (P1) was defined from IC to the first force peak to represent the loading due to impact force. Mid-stance (P2) was the interval between the first and second (maximum) force peaks, which corresponded to full body-weight acceptance. Terminal stance (P3) was defined as the end of P2 to half the time to reach toe off (TO). The peak propulsive force typically occurs at the end of P3. Pre-swing (P4) began at the end of P3 and terminated at TO, which represented the unloading phase before swing occurs.

The CA magnitude for each coupling in each period of stance was classified into one of four coordination phases based on the scheme developed by Chang et al. [28]. A proximal coordination pattern indicated more proximal joint (hip) contribution to motion relative to the distal joint (knee), whereas a distal coordination pattern indicated the inverse. In other words, the hip had a larger excursion than the knee if the CA magnitude fell within the proximal phase range, and a smaller excursion if the CA magnitude was in the distal phase range. An in-phase coordination pattern reflected motion in which both the proximal and distal joints rotated a similar amount in either a positive or negative direction.

A Wilcoxon signed-rank test was used to compare the average variability of the reconstructed lower extremity between the ACLR and control groups for each joint coupling in each period. A second analysis was performed to consider both the reconstructed and contralateral limbs simultaneously. This was done by a multivariate analysis of variance (MANOVA) to compare the average variability between the ACLR and control groups for each coupling in each period. Cohen's d was also calculated to evaluate the effect size of the observed differences.

4. Results

Mean coordination variability ranged between 17 ± 10° and 28 ± 20° (mean ± standard deviation) in the ACL-reconstructed limb compared to 9 ± 6° and 18 ± 18° in the control-matched limb for hip-knee coupled motion during walking (Table 1). Specifically, greater variability was found in the ACL-reconstructed limb compared to the control-matched limb for HR/KR coupled motion during the loading phase (p = 0.017, d = 0.76), HA/KR during mid-stance (p = 0.048, d = 0.61), and HR/KA during terminal stance (p = 0.017, d = 0.84). HR/KA relative motion plots for one matched pair of participants are shown in Figs. 1 and 2, and Fig. 3 shows an ensemble curve of coupling angle variability during the stance phase of walking for each group.

Post-hoc tests further revealed a higher mean coordination variability in the ACL-contralateral limb (range between 10 ± 7° and 19 ± 19°) compared to the control-matched limb (6 ± 3° and

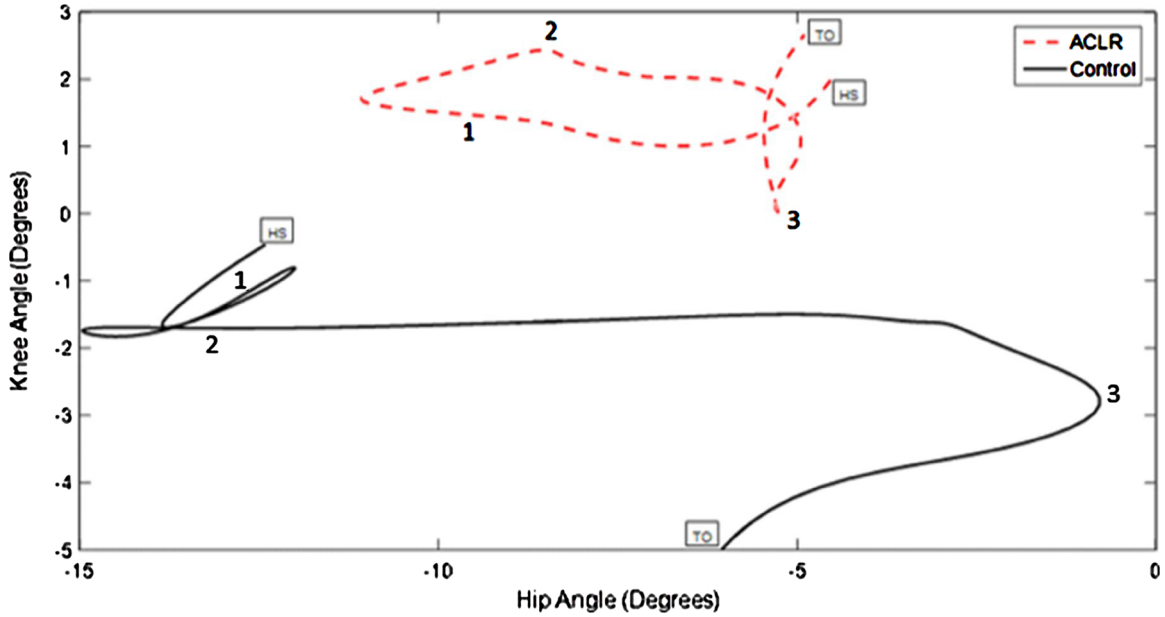

Fig. 1. Angle-angle plot of the hip internal rotation (+)/knee adduction (+) coupling for one matched pair across the stance phase (HS: heel strike; 1: end of P1; 2: end of P2; 3: end of P3; TO: toe off).

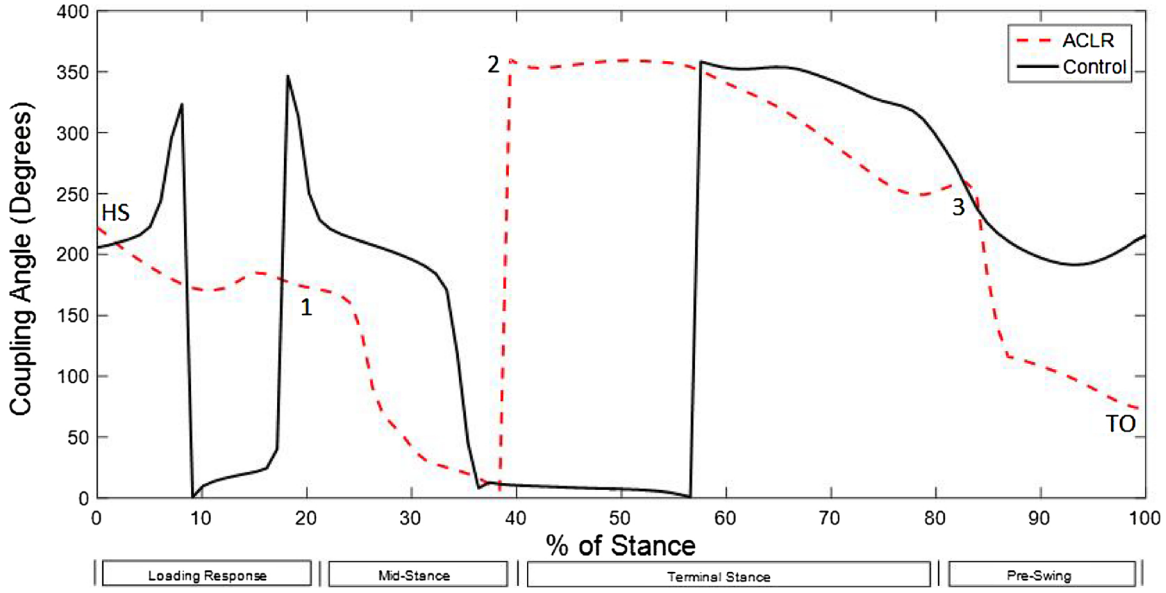

Fig. 2. Coupling angle for hip rotation/knee abduction-adduction across the stance phase (HS: heel strike; 1: end of P1; 2: end of P2; 3: end of P3; TO: toe off).

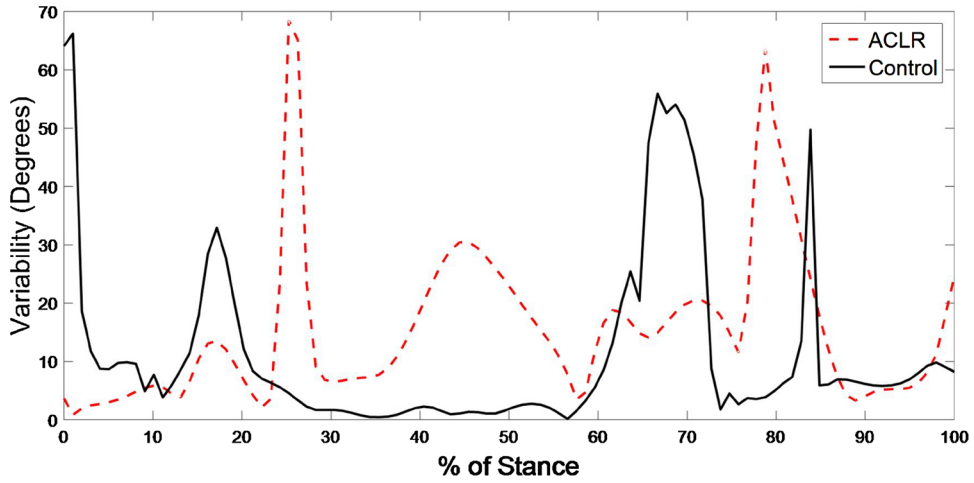

Fig. 3. Ensemble average coupling angle variability in hip rotation/knee abduction-adduction across the stance phase.

11 ± 7°) for hip-knee coupled motion during walking (Table 2). Specifically, greater variability in HA/KR and HR/KR motions during the loading phase ($p = 0.021$ and $p = 0.044$, respectively) and HR/KA during terminal stance ($p = 0.03$) was found in the ACL-contralateral limb compared to the control-matched limb.

There were no significant findings for coordination variability in knee-ankle coupled motion. The coordination patterns varied within and between groups for hip-knee and knee-ankle coupled motion during different periods of stance (Table 3). A common adaptation observed in individuals with ACLR was reduced hip motion compared to controls. Fig. 1 shows HR/KA coupled motion for one matched pair of participants and highlights a reduced hip rotation excursion in the ACL-reconstructed limb compared to the control. Fig. 2 shows the CA magnitude for the same HR/KA coupling and highlights a difference in coordination timing between the ACL-reconstructed and control limbs, most notably during terminal stance. Fig. 3 shows the ensemble curves of average coupling angle variability in HR/KA for each group.

Table 2
Joint coordination variability in the contralateral and reconstructed-match limb for ACLR versus Control.

Coupling	Period	Variability (SD) (Mean ± Standard Deviation)		p-Value	Cohen's d
		ACLR	Control		
HA/KR	1	9.7 ± 6.5	6.3 ± 3.0	0.021	0.67
HR/KR	1	15.5 ± 9.3	10.8 ± 6.9	0.044	0.58
HR/KA	3	18.7 ± 19.1	9.0 ± 6.0	0.030	0.69

Note: P1: HS to initial GRF peak; P3: Second GRF peak to half the time to reach TO.

5. Discussion

The purpose of this study was to assess joint coordination patterns and variability in individuals after ACLR and uninjured controls using vector coding. Contrary to our first hypothesis and previous studies [16,17] that reported decreased variability in a pathologic group, the present study found increased coordination variability in individuals with ACLR. According to Stergiou et al. [29], movement variability that falls outside an optimal range is representative of an unhealthy system; less variability describes a rigid state in which available DOF are constrained, and greater than optimal variability is characteristic of an unstable state. Individuals with PFP who had less variability reported

Table 3
Joint coordination phase magnitude (°) and classification for the reconstructed and matched limbs in ACLR versus Control.

Coupling	Period	Coordination Phase (°) (Mean/Classification)	
		ACLR	Control
KF/ADF	1	17/Proximal	30/In-phase
HA/KA	2	239/In-phase	190/Proximal
HA/KR	2	230/In-phase	202/Proximal
HR/KA	2	334/Anti-phase	350/Proximal
KF/ADF	2	112/Distal	122/Anti-phase
HR/KA	3	194/Proximal	205/In-phase
HR/KA	4	146/Anti-phase	176/Proximal

Note: P1: HS to initial GRF peak; P2: initial GRF peak to second GRF peak; P3: Second GRF peak to half the time to reach TO; P4: end of P3 to TO.

pain, which may have elicited an avoidance of painful movement strategies and thus a more rigid system. Conversely, the individuals with ACLR who had greater variability did not have pain, but may have persistent neuromuscular deficits which result in instability or the inability to produce optimal coordination patterns. Pollard et al. [30] evaluated lower extremity coupling in female soccer players with ACLR during a side-step cutting maneuver using vector coding, and reported increased variability in an ACLR group compared to controls in HR/KA, HF/KA, KA/KF, and KA/KR couplings. These findings are consistent with the current study, and were attributed to altered neuromuscular control.

For the HA/KA coupling during P2, the ACLR group showed an in-phase relationship but controls demonstrated increased hip contribution. In normal gait the hip adducts in early stance and undergoes more frontal plane motion than the knee, which is consistent with the proximally dominant coordination pattern seen in the control group. However, the observed in-phase pattern in the ACLR group could support the idea offered by Gribbin et al. that the hip is acting as a stabilizer, and thus has constrained motion. The same coordination patterns were also found in the HA/KR coupling during P2, and a similar conclusion can be drawn. Since the knee is mostly extended during mid-stance, it does not experience much internal-external rotation. Therefore, the hip should contribute more frontal plane motion in this coupling. However, the ACLR group again exhibited decreased hip motion, indicative of a stabilizing role.

For the HR/KA coupling during P2, the control group maintained a proximally dominant coordination pattern but the ACLR group displayed anti-phase motion. The hip normally reaches peak internal rotation during mid-stance, and the knee exhibits minimal frontal plane motion throughout the gait cycle, which is reflected in the control coordination pattern. Conversely, the knee appeared to abduct slightly in the ACLR group while the hip was internally rotated, which is reflective of their anti-phase pattern. Interestingly, this coupling transitioned to a proximally dominant pattern in individuals with ACLR during terminal stance (P3) while controls exhibited in-phase motion. During terminal stance, the hip progresses towards maximal extension, and this is accompanied by internal rotation. Individuals with ACLR seemed to depend primarily on the hip during this stance period, avoiding frontal plane knee motion. On the contrary, the control group demonstrated both hip internal rotation and knee adduction. We think the more unstable position that occurs as the heel lifts off the ground during terminal stance possibly triggers individuals with ACLR to limit knee motion as a protective mechanism. Finally in P4, HR/KA was in anti-phase coordination in the ACLR group but controls demonstrated hip dominance. In this portion of stance, the hip begins to flex, and this is accompanied by external rotation. Results suggest that individuals with ACLR may be unable to control frontal motion at the knee prior to the swing phase. It is also possible that the hip has reduced rotational motion to help stabilize the knee.

Our results related to those reported by Gribbin et al. [22] who studied the effect of ACLR on the hip-knee joint coupling during walking and jogging using vector coding. The magnitude of joint excursion for the hip transverse-knee frontal coupling was increased in the ACLR group during late swing, but decreased in all other couplings (hip frontal-knee frontal, hip frontal-knee sagittal, hip frontal-knee transverse, hip sagittal-knee frontal, and hip sagittal-knee transverse) during all sub-phases in walking compared to controls. CA values were increased in the ACLR group during all sub-phases of walking, and variability was also increased in the ACLR group during mid- and late stance phases of walking for all couplings except the hip sagittal-knee frontal and hip sagittal-knee transverse couples during mid-stance. Findings of decreased magnitude of joint excursion in the ACLR group suggested that the ACLR knee may experience constrained motion as a strategy for stabilizing the joint. Further, the authors proposed that increased variability in the ACLR group could represent a struggle to find an optimal movement pattern in the ACLR limb.

These results suggest that despite meeting the criteria for return to sport, indicating a restoration of knee function and stability, individuals with ACLR exhibit altered coordinative function in their reconstructed lower extremity compared to non-injured controls. Coordination is a product of the dynamic interactions between the nervous system, musculoskeletal system, and environment, and the self-organized relations among these constituents are what allow for flexibility in movement patterns [31]. Therefore, deviations from normal coordination variability observed in ACLR individuals in the present study may be evidence of sensorimotor deficits relating to neuromuscular control, and although we did not follow the participants over time we examined their joint coordination variability at a time point as long as 7 years after rehabilitation.

One of the limitations of the study is that many of the subjects only had at most three acceptable trials with clear camera views of markers and full foot contact on force plates. More trials would be desirable, but not clinically practical due to the time required to collect data including EMG on a variety of activities of daily living, as well as proprioception, strength and return to sport assessments. We used 20 subjects in our study, for which it has been estimated that three trials are sufficient to compare within-person variability between groups with an effect size of 0.80 [32]. Additionally, two vector coding methods have been used to calculate the variability between angle-angle cyclograms in coordination studies. One uses the vector coupling angle [17] and the other uses both the vector coupling angle and the length [33]. It has been proposed that the coefficient of correspondence, which accounts for both vector magnitude and vector coupling angle, may provide a better measure of variability than vector angle alone [33]. Both vector coding methods have been shown to potentially be affected by statistical artefacts associated with the application of circular statistics. This may artificially increase the estimated coordination variability, especially when the vectors are short [34], something that is expected to occur when running. Besides vector coding, a variety of other methods have also been proposed, each one often mentioning the shortcomings of the others. These include geometric moments [35]; area [36]; a normalized root-mean-squared error technique [37]; functional data analysis, which is the application of principal component analysis techniques to functional data [38]; a bivariate approach [39]; and one based on the bivariate method that defines an ellipse from the coupling vector end points and calculates its area [34]. We used the vector coding method and focused on the vector angle for our analysis. Although we could have used any of the other methods it is not clear at present what the optimal approach should be as it would appear that each method has some limitation or drawback and a test to objectively compare methods is not available.

The goal of this study was to quantify and compare lower extremity joint coordination between ACLR and healthy individuals, and current literature pertaining to this objective is limited with respect to the use of vector coding for evaluation of gait in an ACLR participant group.

Therefore, this broad investigation of joint coordination variability provides a framework for future studies that should focus on hip-knee couplings in both frontal and transverse planes from heel strike to terminal stance. The inclusion of Cohen's d further suggested a medium to large effect of ACLR on select measures of variability, and provided additional support for the pursuit of a better understanding of joint coordination in the ACLR population, specifically for HR/KA and HR/KR.

6. Conclusion

We evaluated joint coordination variability in an ACLR population, and our findings suggest that individuals with ACLR do not display normal coordinative function. The increased variability observed in this group during walking relative to uninjured individuals may be an indicator for the risk of injury in more dynamic tasks like jumping and pivoting. Existence of an optimal range of coordination variability has been supported in the literature and related to the ability to perform functional movement [40]. A deviation greater than the optimal state is associated with instability and reduced adaptability. The present study was retrospective so it remains unknown whether increased coordination variability is a cause for, or effect of ACL injury and subsequent repair. Overall, individuals with ACLR exhibited increased joint coordination variability in several hip-knee couplings compared to matched controls during walking. Individuals with ACLR also displayed altered coordination patterns, suggesting the use or availability of different movement strategies compared to controls. Thus, coordinative function may not be fully restored in individuals with ACLR following rehabilitation and/or return to sport, or abnormal coordinative function may be an implication for ACL injury. Inclusion of electromyography may add insight to the interpretation of coordination patterns. A prospective investigation of coordination focusing on hip-knee coupled motion in both frontal and transverse planes may contribute to improving ACL injury prevention and rehabilitation, and supplement research on the development of osteoarthritis.

Declarations of interest

None.

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