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Subtle alterations in whole body mechanics during gait following anterior cruciate ligament reconstruction

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ABSTRACT

Background: Clinically, normalization of gait following anterior cruciate ligament reconstruction (ACLr) is defined as the absence of observable deviations. However, biomechanical studies report altered knee mechanics during loading response (LR); a time of double limb support and weight transfer between limbs. It is conceivable that subtle adjustments in whole body mechanics, including center of mass (COM) velocity and ground reaction force (GRF) peaks and timing, are present.

Research question: The purpose was to compare limb and whole body mechanics during LR of gait in the surgical and non-surgical limbs post-ACLr.

Methods: Anterior and vertical COM velocity at initial contact; knee flexion range of motion, peak knee extensor moment, peak vertical and posterior GRF, minimum vertical COM position and maximum anterior and vertical COM velocity during LR were identified for twenty individuals 112 ± 17 days post-ACLr without observable gait deficits. To assess differences in timing of COM variables, coupling angles (vector coding) were calculated for multidirectional coordination of vertical and anteroposterior COM velocities and GRFs and categorized as in-phase, anti-phase, vertical phase, or anteroposterior phase coordination. Paired t-tests compared peaks between limbs; non-parametric Wilcoxon signed-rank tests compared coordination pattern frequency.

Results: Less knee range of motion (5.6 °), 30% smaller knee extensor moment, 11% smaller posterior GRF, and slower anterior COM velocity at initial contact (2%) and peak during LR (1.3%; all $p < 0.05$) were observed in the surgical compared to the non-surgical limb. For COM velocity coordination, lesser anti-phase (7.38%) and greater in-phase coordination (2.88%) were observed in the surgical limb. For GRF coordination, less in-phase coordination (1.94%) was observed in the surgical limb.

Significance: Differences in coordination patterns, suggest that individuals post-ACLr make subtle adjustments in timing of whole body mechanics; particularly in COM velocity during gait. These adjustments are consistent with reduced sagittal plane loading in the surgical knee.

1. Introduction

Clinically, normalization of gait following anterior cruciate ligament reconstruction (ACLr) is expected by 8–12 weeks post-surgery [1,2]. The absence of observable gait deviations is used as a clinical criterion for determining normalized gait; however, restoration of gait does not translate into the return of symmetrical knee mechanics for all individuals post-ACLr. This is underscored by the presence of knee extensor moments deficits (up to 50% lower) in the surgical compared to non-surgical limb during loading response of gait 3–6 months post-surgery [3–6]. Concurrently, deficits in knee kinematics, notably increased knee flexion at initial contact and limited knee flexion motion during loading response, describe a general pattern of knee stiffening [3,6,7].

Gait is a semiautomatic functional task that is performed with reciprocal patterned limb kinematics and kinetics [8]. Substantial asymmetries in limb mechanics result in observable alterations in spatiotemporal characteristics and whole body center of mass (COM) kinematics during gait. These alterations are well documented in individuals following stroke with unilateral hemiparesis [9] and lower limb amputations [10] and are necessary to maintain forward progression. Given that individuals 3 months post-ACLr demonstrate spatiotemporal symmetry with visual appearance of normalized gait [3,4,11], it may be concluded that whole body adjustments are not necessary to compensate for less substantial limb asymmetries. However, previous studies suggest that adjustments at the whole body level

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compensate for relatively minor limb asymmetries, as ground reaction forces (GRFs) asymmetries have been reported in individuals with small leg length discrepancies [12,13]. It is not known if relatively small asymmetries in mechanics at a single joint, such as those observed post-ACLr, are reflected in whole body mechanics.

The relatively small between limb differences in knee mechanics post-ACLr are observed during loading response of gait; a time of double limb support and weight transfer between limbs. At this time, sagittal plane knee mechanics ensure limb stability and allow for forward progression through the transfer of body mass [8]. In the absence of spatiotemporal asymmetries and observable gait alterations, profound alterations in COM position are not likely. However, it is conceivable that subtle alterations in COM velocity are present. Between limb differences in posterior GRF and altered knee mechanics have been identified during gait post-ACLr [14]. As alterations in GRF require changes in acceleration of the COM [8], measures of COM velocity may capture subtle whole body adjustments. COM velocity profiles during double limb support are sensitive to between limb adjustments in unilateral amputees [10]. Moreover, subtle adjustments at the whole body level may not only be reflected in peak magnitude, but also in timing of velocity and GRF magnitudes. The purpose of this study was to quantify whole body mechanics during loading response of gait post-ACLr. Based on previous literature, we hypothesized that the surgical limb would exhibit reduced knee ROM, extensor moment, posterior GRF, and anterior COM velocity compared to the non-surgical limb. In addition, analyses of COM velocity and GRF coordination in vertical and anteroposterior directions will reveal differences in timing of these variables between limbs.

2. Methods

Based on preliminary data (N = 4), sample size calculations indicated that a minimum of 10 participants were needed to detect within-subject differences with 80% power and $\alpha = 0.05$ for the primary outcome variables. Twenty individuals (13 females, 25 ± 10 years) 112 ± 17 days after primary unilateral ACLr (bone-patellar tendon-bone, hamstring, or quadriceps tendon autograft or allograft) were enrolled. Three participants reported previous ACLr to the contralateral (n = 1) and ipsilateral knee (n = 2) greater than 4 years. These individuals returned to pre-surgical levels of physical activity prior to re-injury. Participants were included if they were between 14–50 years, at least 10 weeks post-ACLr, participating in physical therapy, and walked without observable gait deviations. Participants were excluded if they had a concurrent knee pathology that limited weight bearing status of their surgical limb or current injury to the non-surgical limb.

2.1. Procedures

Testing took place at the University of Southern California's Human Performance Laboratory. All procedures were explained and an informed consent was obtained as approved by the Institutional Review Board of the University of Southern California, Health Sciences Campus. Parental consent and youth assent were obtained for participants under 18 years.

Prior to testing, participants warmed up on a stationary bike and completed self-reported measures of knee function (IKDC Subjective Knee Evaluation Form) and knee pain (VAS). Reflective markers were placed bilaterally over the following anatomical landmarks: 1 st and 5th metatarsal heads, distal end of second toe, medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanters, iliac crests, posterior superior iliac spines, acromions, and L5-S1 junction, C7 spinous process, and sternum. Tracking marker clusters, secured to rigid plates, were placed bilaterally on lateral thigh, shank, and heel of their shoe. Kinematic and GRF data were collected using a 14-camera digital motion capturing system (250 Hz) and force platforms (1000 Hz; BTS Bioengineering, Brooklyn, NY).

Participants walked at 1.4 m/s and walking velocity was determined using laser timing gates placed 5 m apart centered on force plates. Practice trials allowed participants to become familiar with the speed. Five trials per limb were collected where gait velocity fell within 5% of the 1.4 m/s.

2.2. Data analysis

Reconstructed three-dimensional marker-coordinates (BTS SMARTTracker v1.10) and GRF data were used to calculate sagittal plane joint kinematics and kinetics (Visual3D v5, C-Motion, Inc., Germantown, MD). Coordinate and GRF data were low-pass filtered using a fourth order zero-lag Butterworth filter with cut-off frequencies of 12-Hz and 40-Hz, respectively [14,15]. Local coordinate systems of body segments were derived from a standing calibration using a joint coordinate system approach [16]. Six degrees-of-freedom of each segment were calculated by transforming marker triads on clusters to the position and orientation of each segment during the static trial. Body mass was calculated by dividing average (5 s) vertical GRF during standing by 9.81 N/kg. The body's COM was estimated from trunk and lower extremity segments [17]. Kinematics, anthropometrics, and GRFs were used (inverse dynamics) to calculate internal net joint moments normalized to body mass [18].

Data were analyzed during loading response (LR) of stance, from initial contact (IC) to first peak in knee flexion (0–25% stance). IC occurred when vertical GRF became greater than 30 N. GRF data were normalized to and expressed in body weights (BW). Four trials per limb were averaged for analysis.

The following variables were identified: anterior and vertical COM velocity at IC; knee flexion range of motion (IC to peak knee flexion), peak knee extensor moment, peak vertical and posterior GRF, minimum vertical COM position and maximum anterior and vertical COM velocity during LR (Fig. 1). COM position was normalized to COM height during static trial.

Timing of whole body parameters was assessed using vector coding analyses. These analyses allow for quantification of timing of two whole body parameters relative to each other. While traditionally used to analyze multi-planar joint kinematics [19,20], this method allows for the exploration of concurrent changes in timing and for the identification of the relative predominance of a variable in one direction (vertical) versus another (anteroposterior) [21]. Kinematic and GRF data from IC to peak knee flexion were normalized to 101 data points for further analysis. COM velocity and GRF coordination were quantified separately from profile plots with anteroposterior and vertical directions on the x- and y-axes, respectively (Fig. 2a). Coupling angles (γ) were determined by connecting consecutive time points on a profile plot and determining the counterclockwise angle of this line relative to the right horizontal (Fig. 2b) [22,23]. Coupling angles were determined using:

$$\gamma_{j,i} = \tan^{-1}((y_{j,i+1} - y_{j,i})/(x_{j,i+1} - x_{j,i}))$$

where $0^\circ \leq \gamma \leq 360^\circ$ and i is a percentage of stance of the jth trial. Coupling angles were calculated between adjacent time points, totaling 100 coupling angles, and categorized into a coordination pattern: anti-phase, in-phase, anteroposterior phase, and vertical phase (Fig. 2c). Anti-phase coordination indicates that COM velocity or GRF magnitude in anteroposterior and vertical directions are in opposite directions between data points (e.g. increasing anteroposterior and decreasing vertical motion or forces). In-phase coordination indicates that the variable is changing in the same direction for anteroposterior and vertical directions between data points (e.g. increasing anteroposterior and vertical motion or forces). Anteroposterior phase coordination (horizontal line) indicates that the variable is primarily changing in the anteroposterior direction with minimal change in the vertical direction.

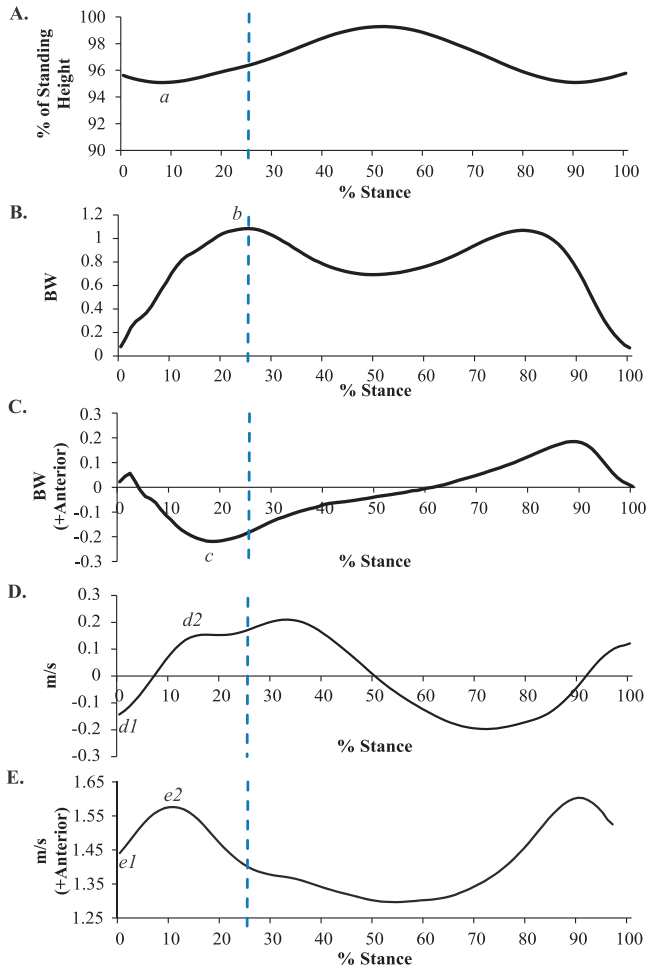

Fig. 1. Representative time series plots for (A) COM position, (B) vertical and (C) anteroposterior ground reaction force, and (D) vertical and (E) anterior COM velocity. Dashed blue line indicates peak knee flexion (not pictured; approximately 25% of stance). Lowercase letters on each plot indicate peaks where each dependent variable was identified (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

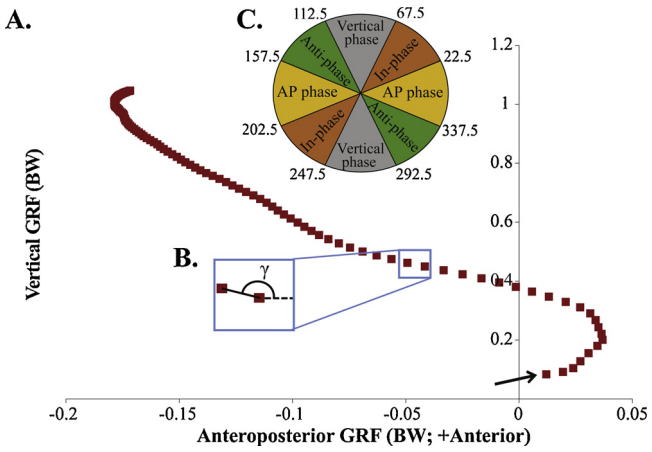

Fig. 2. (A) Representative plot of vertical and anteroposterior GRFs with black arrow denoting heel strike and direction of motion. (B) Blue box inset demonstrates how coupling angle (γ) was determined based on the vector orientation between two consecutive data points relative to the right horizontal. (C) Categorization of coupling angles into coordination patterns (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

Vertical phase coordination (vertical line) indicates that the variable is primarily changing in the vertical direction with minimal change in the anteroposterior direction. Because coupling angles are directional, circular statistics were used to calculate mean coupling angles and standard deviations [23,24]. For each trial and subject, coupling angle was first calculated to identify coordination patterns during loading response. Then frequency of each coordination pattern was averaged across five trials for each limb and subject to calculate mean coordination pattern frequency for anti-phase, in-phase, vertical phase, or anteroposterior phase coordination.

2.3. Statistical analysis

Dependent variables were compared between surgical and non-surgical limbs: anterior and vertical COM velocity at IC; knee range of motion, knee extensor moment, vertical and posterior GRF, minimum COM position, and maximum anterior and vertical COM velocity during LR. To account for multiple comparisons, one-way MANOVA was performed to determine the effect of limb. In the case of a significant effect of limb, paired t-tests were performed for post hoc testing.

For coordination analyses, the dependent variable, coordination pattern frequency, is a discrete variable. Therefore, non-parametric Wilcoxon signed-rank tests were used to compare mean rank differences in coordination pattern (anti-phase, in-phase, vertical phase, or anteroposterior phase) frequency between surgical and non-surgical limbs for COM velocity and GRF couplings. The assumption of symmetrical distribution of differences between limbs was assessed by histogram and met for each coordination pattern. Significance level was set at $\alpha = 0.05$ for all analyses (SPSS Statistics v22, Chicago, IL).

3. Results

Data were reported as mean (SD) unless otherwise stated. Self-reported IKDC and VAS were 61 (10.5) and 0.4 (0.5) cm. The MANOVA indicated a significant effect of limb ($p = 0.001$). Post hoc analyses revealed that, on average, the surgical limb exhibited less knee range of motion (-5.57 (4.46) degrees; $p < 0.001$), knee extensor moment (-0.185 (0.250) Nm/kg; $p = 0.004$), posterior GRF (-0.023 (0.033) BW; $p = 0.006$), and anterior COM velocity at IC (-0.032 (0.054) m/s; $p = 0.015$) and peak during LR (-0.024 (0.047) m/s; $p = 0.034$) compared to the non-surgical limb (Table 1). No between limb differences in vertical GRF ($p = 0.06$), minimum COM position ($p = 0.428$), or vertical COM velocity at IC ($p = 0.953$) or peak during LR ($p = 0.902$) were observed.

Coordination data were reported as median differences between limbs (Table 2). For COM velocity coordination, surgical limb exhibited less anti-phase ($p = 0.035$) and greater in-phase coordination ($p = 0.046$) compared to non-surgical limb (Fig. 3a & c). No differences in COM velocity vertical phase ($p = 0.550$) and anteroposterior phase coordination ($p = 0.322$) were observed. For GRF coordination, surgical limb exhibited less in-phase coordination ($p = 0.026$) compared to non-surgical limb (Fig. 3b & d). No differences in GRF anti-phase ($p = 0.489$), vertical phase ($p = 0.341$), and anteroposterior phase ($p = 0.823$) coordination were observed.

4. Discussion

These data indicate that alterations in whole body mechanics exist along with impairments in sagittal plane knee mechanics during loading response of gait in individuals post-ACLr. Self-reported IKDC scores and knee pain during testing suggest that participants were progressing typically [25]. Altered knee mechanics observed in this study are consistent with previous studies [3–7,14]. Participants walked without observable gait alterations, which was supported by an absence of differences in COM position. The surgical limb exhibited less knee range of motion (5.6 °) and extensor moments (30%) compared to

Table 1
Between limb comparisons in whole body mechanics during loading response.

	Surgical limb	Non-surgical limb	Significance	Between Limb Difference	
				95% Confidence Interval	Effect Size
Knee flexion ROM (degrees)	11.31 ± 4.31	16.88 ± 3.15	< 0.001[a]	[-7.66, -3.48]	1.25
Knee extensor moment (Nm/kg)	0.435 ± 0.214	0.620 ± 0.191	0.004[a]	[0.068, 0.302]	0.74
Vertical GRF (BW)	1.00 ± 0.06	1.04 ± 0.08	0.060	[-0.083, 0.002]	0.43
Posterior GRF (BW)	0.187 ± 0.030	0.210 ± 0.024	0.006[a]	[-0.038, -0.007]	0.69
Minimum COM position (% Standing height)	95.9 ± 2.1	95.5 ± 0.75	0.428	[-0.006, 0.013]	0.19
Vertical COM velocity (m/s)					
At initial contact	−0.151 ± 0.034	−0.151 ± 0.034	0.953	[-0.013, 0.012]	< 0.01
Maximum during loading response	0.173 ± 0.030	0.174 ± 0.025	0.902	[-0.012, 0.010]	0.04
Anterior COM velocity (m/s)					
At initial contact	1.42 ± 0.06	1.45 ± 0.01	0.015[a]	[-0.057, -0.007]	0.54
Maximum during loading response	1.57 ± 0.05	1.59 ± 0.05	0.034[a]	[-0.046, -0.002]	0.44

Note: Values presented as mean ± standard deviation. [a]Significant differences between limbs ($p < 0.05$).

Table 2
Between limb comparisons in COM velocity and GRF coordination during loading response.

	Surgical limb	Non-surgical limb	Difference	Significance
COM Velocity Coordination (%)				
Anti-phase	10.38	21.25	−7.38	0.035[a]
In-phase	38.38	29.38	2.88	0.046[a]
Vertical phase	22.50	22.13	2.25	0.550
Anteroposterior phase	23.25	25.25	−1.25	0.322
GRF Coordination (%)				
Anti-phase	21.75	20.00	1.38	0.489
In-phase	7.13	9.75	−1.94	0.026[a]
Vertical phase	64.50	62.88	2.88	0.341
Anteroposterior phase	5.50	5.25	−0.75	0.823

Note: Values presented as medians and median differences. [a]Significant differences between limbs ($p < 0.05$).

the non-surgical limb. Between limb differences in posterior GRF and anterior COM velocity suggest that whole body mechanics are altered. Posterior GRFs were 11% smaller in the surgical limb, which is similar to previous work [14]. While the magnitude of differences in anterior COM velocity were small (< 2%), smaller COM velocity in the surgical limb are consistent with smaller posterior GRFs. However, the presence of smaller posterior GRFs along with limited knee flexion is counter-intuitive, as knee flexion during loading response should reduce impact acceleration and decrease GRFs. Limited knee flexion typifies a more rigid strategy of loading usually associated with greater impact acceleration and GRFs. The combination of reduced knee flexion with less anterior COM velocity and posterior GRFs suggests that adjustments in whole body mechanics are not the product of limited knee motion but are modulated separately from knee motion to limit GRFs in the surgical limb. This supports the premise that individuals post-ACLr alter COM velocity during double limb support and make subtle alterations in whole body mechanics to reduce demands on the surgical knee.

Vertical GRF and COM velocity at initial contact and during loading response were not different between limbs, suggesting that these features of whole body mechanics are not altered during gait post-ACLr. Preservation of symmetry in one direction may help give the appearance of normalized gait. The absence of differences in vertical GRF is consistent with previous studies [14] and may be attributed to between limb comparisons, as vertical GRFs are modulated by walking velocity, which is the same between limbs [8].

Vector coding techniques were able to identify subtle alterations in relative timing of whole body mechanics that were not captured in comparisons of peak GRF and COM velocity. Differences between limbs were more apparent in timing of vertical and anteroposterior COM velocity. In general, vertical COM velocity is negative at initial contact and increases throughout loading response; this is accompanied by an increase then decrease in anterior COM velocity (Figs. 1 & 3). Areas reflecting greater in-phase and less anti-phase coordination in the surgical limb are highlighted in Fig. 3c. Differences in coordination patterns observed just after foot contact account for the greater proportion of in-phase coordination in the surgical limb. Anterior COM velocity is lower at initial contact and increases proportionally with vertical COM velocity, resulting in in-phase coordination. In contrast, in the non-surgical limb, increases in vertical COM velocity are observed with relatively faster and smaller changes in anterior COM velocity, resulting in a greater portion of vertical phase coordination. It is not clear if more gradual increases in anterior COM velocity (resulting in greater in-phase coordination) in the surgical limb are used to modulate posterior GRFs or if this is product of lower anterior COM velocity at initial contact. The appearance of prolonged increases in anterior COM velocity past the juncture when it slows down in the non-surgical limb suggests that adjustments in relative timing of COM velocity may be necessary to compensate for lower initial anterior COM velocity. In essence, it appears that the prolonged increase in anterior COM velocity may be required to catch up and maintain somewhat similar velocities between limbs. This difference in timing is reflected in the larger proportion of anti-phase coordination in the non-surgical limb. Anterior COM velocity decreases while vertical COM velocity continues to increase during the later portion of loading response, resulting in a larger proportion of anti-phase coordination. In the surgical limb, a more abrupt decrease in anterior COM velocity occurs at the end of loading response with little to no change in vertical COM velocity.

A small difference between limbs in in-phase coordination was observed in GRF coordination, as in-phase coordination accounted for less than 10% of loading response. Less in-phase coordination in the surgical limb indicates that the surgical limb exhibited less concurrent increases in vertical and anteroposterior GRFs. These between limb differences were subtle and only representative of a fraction of loading response, appearing at the junction just after peak posterior GRF (Fig. 3d). While these differences were statistically significant, it is unlikely that they are clinically meaningful. No additional alterations in GRF coordination were observed, suggesting that the magnitude of GRFs is modulated without altering timing.

During loading response, the goals of impact absorption and forward progression are successfully achieved to maintain visual appearance of unaltered gait. Alterations in posterior GRF, anterior COM velocity, and COM velocity and GRF timing demonstrate adjustments in whole body mechanics during gait. Smaller posterior GRF and anterior COM velocity suggests that these alterations are made to facilitate decreased loading in the surgical limb. Alterations in timing of

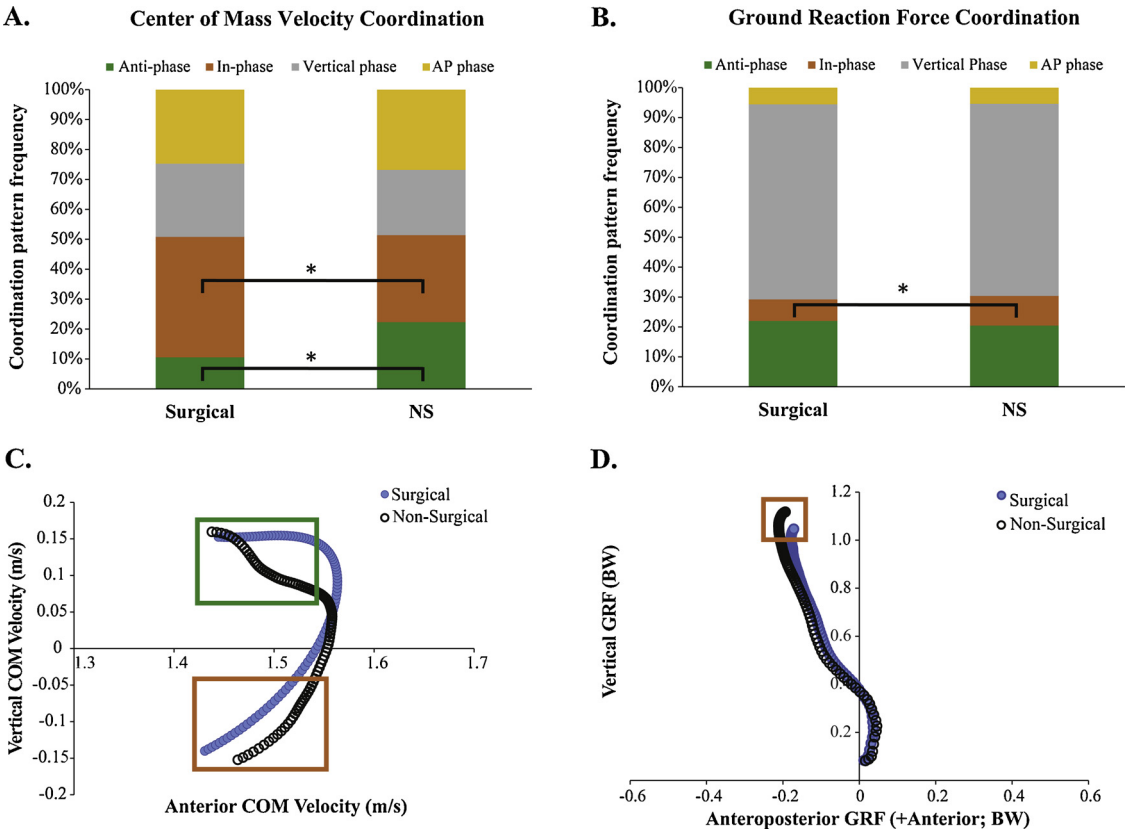

Fig. 3. Coordination of (A) COM velocity and (B) GRF and average time series plot of vertical and anteroposterior (C) COM velocities and (D) GRFs during loading response in the surgical and non-surgical limbs. Boxed regions indicate regions of altered coordination; *p < 0.05.

anteroposterior and vertical COM velocity and the magnitude of posterior GRFs appear to be a mechanism to compensate for these differences. Moreover, the absence of alterations in vertical COM position and velocity and vertical GRF may work to maintain the presence of visually unaltered gait post-ACLr. Future studies are needed to determine how this information can be used to inform rehabilitation strategies.

This study has several limitations. Interpretation of these data assumes that the non-surgical limb exhibits normal gait mechanics. While it provides the best frame of reference for within-subject comparisons, this may not be accurate. Our aim was to examine whole body mechanics in individuals with unilateral knee deficits following ACLr. However, when interpreting these data, one must consider that participants with concomitant meniscal injuries and previous ACLr were included. They had no additional weight-bearing restrictions following surgery and were progressing typically. Only one participant had reconstructive surgery on the contralateral limb. As results were similar when analyzed without this participant, these data were not removed. The COM model utilized accounted for 82% of body mass, but did not include estimations of the arms and head. Previous work suggests that this model is an accurate representation of the COM compared to full body models for running and cutting [17]. Furthermore, small differences in COM velocity should be interpreted with caution due to their reliance on affixed markers and motion capture, which may have greater error. However, larger differences observed in relative timing are characterized across consecutive time points and thus are less likely to be influenced by marker error.

5. Conclusions

Altered knee mechanics are present at a time when clinical normalization of gait is expected 3 months post-ACLr. Deficits in posterior GRF and anterior COM velocity are consistent with strategies to reduce loading in the surgical knee. Differences in coordination patterns suggest that individuals may be able to compensate for initial changes in whole body control during gait by altering timing of COM velocity and GRF during stance on the surgical limb.

Conflicts of interest

None.

This study was approved by the Institutional Review Board of the Health Sciences Campus of the University of Southern California.

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