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Characterizing Knee Loading Asymmetry in Individuals Following Anterior Cruciate Ligament Reconstruction Using Inertial Sensors

Susan M. Sigward, Ph.D,

Human Performance Laboratory, Division of Biokinesiology and Physical Therapy, University of Southern California, 1540 Alcazar St, CHP 155, Los Angeles, CA, 90089-9006, United States.

Ming-Sheng M. Chan, MS, and

Human Performance Laboratory, Division of Biokinesiology and Physical Therapy, University of Southern California, 1540 Alcazar St, CHP 155, Los Angeles, CA, 90089-9006, United States. mingshec@usc.edu, Tel.: +1-323-442-2948

Paige E. Lin, MS

Human Performance Laboratory, Division of Biokinesiology and Physical Therapy, University of Southern California, 1540 Alcazar St, CHP 155, Los Angeles, CA, 90089-9006, United States. paigeeli@usc.edu, Tel.: +1-323-442-2948

Abstract

Limitations in the ability to identify knee extensor loading deficits during gait in individuals following anterior cruciate ligament reconstruction (ACLR) may underlie their persistence. A recent study suggested that shank angular velocity, directly output from inertial sensors, differed during gait between individuals post-ACLR and controls. However, it is not clear if this kinematic variable relates to knee moments calculated using joint kinematics and ground reaction forces. Heel rocker mechanics during loading response of gait, characterized by rapid shank rotation, require knee extensor control. Measures of shank angular velocity may be reflective of knee moments. This study investigated the relationship between shank angular velocity and knee extensor moment during gait in individuals ($n=19$) 96.7 $\pm$ 16.8 days post-ACLR. Gait was assessed concurrently using inertial sensors and a marker-based motion system with force platforms. Peak angular velocity and knee extensor moment were strongly correlated ($r=0.75$, $p<0.001$) and between limb ratios of angular velocity predicted between limb ratios of extensor moment ($r^2=0.57$, $p<0.001$) in the absence of between limb differences in spatiotemporal gait parameters. The strength of these relationships indicate that shank kinematic data offer meaningful information regarding knee loading and provide a potential alternative to full motion analysis systems for identification of altered knee loading following ACLR

Corresponding author: Dr. Susan M. Sigward will be the corresponding author of this manuscript. Sigward@usc.edu, Tel.: +1-323-442-2454.

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Keywords

ACL reconstruction; gait impairments; knee extensor moment; shank angular velocity; inertial sensors

Introduction

Clinically, gait is expected to normalize 8-12 weeks after surgery following anterior cruciate ligament reconstruction (ACLR). While individuals are typically walking without observable gait deviations, biomechanical studies using three-dimensional motion analysis and force platforms indicate that substantial knee loading deficits exist at this time [1]. Specifically, individuals exhibit reduced knee extensor moments in the surgical knee compared to the non-surgical knee as large as 30% [1]. These deficits have been found to persist up to 2 years following surgery [2-4]. This is particularly concerning as the progression of knee osteoarthritis (OA) in this already vulnerable population [5] has been attributed to alterations in knee loading [6-8]. Substantial between limb knee extensor moment asymmetries are often present along with much smaller differences in knee flexion angle (on average, 4 degrees) [2-4], making them difficult to detect without traditional motion analysis and force platforms. Difficulty identifying these asymmetries clinically may underlie their persistence. Despite the accuracy of traditional motion analysis systems in detecting asymmetrical knee loading, widespread use in clinical settings is precluded by their expense and associated time and expertise requirements. The development of methods for clinical detection of asymmetrical knee loading during gait may help direct targeted interventions to prevent long-term deficits post-ACLR.

Recent advances in wearable technology have improved our ability to measure segment kinematics using wearable inertial sensors. Inertial sensors have advantages over traditional motion analysis systems including portability, wireless capabilities, high capture rates, ease of use and cost. However, inertial sensors are limited to the measurement of individual segment kinematics. While procedures for calculating knee joint angles using sensors on thigh and shank segments have improved, error (approximately 3 degrees) associated with integrating raw sensor data remain too large to identify small between limb differences observed during gait following ACLR [9,10]. The use of inertial sensors for calculation of knee joint moments is further precluded by the absence of ground reaction force data. However, direct outputs from accelerometers and gyroscopes have been used to determine foot contact and toe off events during gait, allowing for calculation of spatiotemporal parameters [11-13]. These methods have been validated and used to identify spatiotemporal deficits in individuals with hemiparesis [14,15], but it is not known if the procedures would be applicable in individuals following ACLR. While a previous study found no spatiotemporal deficits in individuals following ACLR compared to healthy controls, the subjects were, on average, 3 years post-surgery [13]. The use of inertial sensors for the identification of spatiotemporal differences may be more informative during early rehabilitation while individuals are still restoring gait mechanics. In addition to identification of spatiotemporal events, direct output of angular velocity has been found to be sensitive enough to identify differences during gait between individuals post-ACLR and healthy

controls [13]. Specifically, features of sagittal plane shank angular velocity during stance and swing phases of the gait cycle differed between groups. While this indicates that inertial sensors have the potential to detect subtle differences in gait attributed to injury or pathology, it is not clear if these differences relate to the relevant knee loading impairments found to persist after ACLr.

It is possible that differences in shank angular velocity in individuals following ACLr [13] are reflective of kinematic alterations made to reduce knee loading during gait. Decreased shank angular velocity has been observed in individuals following ACLr during early stance of gait just after heel strike corresponding to the time at which reduced knee extensor moments are typically observed following surgery [13]. Just after heel contact individuals transition from double-limb to single-limb support, utilizing heel rocker mechanics [16]. Heel rocker mechanics are characterized by rapid forward progression of the shank over the heel just after ground contact. This results in knee flexion which is controlled eccentrically by the quadriceps [16]. As these mechanics coincide with the time at which decreased knee extensor moments are observed in individuals following ACLr [1], it stands to reason that decreases in shank angular velocity in the surgical limb may reflect alterations in heel rocker mechanics and as a result knee extensor moments. Understanding the relationship between shank segment kinematics and knee extensor moments during gait in individuals following ACLr is the first step toward developing clinical testing procedures for detection of knee loading deficits in this population.

The purposes of this study were to 1) determine if between limb differences exist in spatiotemporal characteristics and sagittal plane shank angular velocity during loading response of gait measured using inertial sensors during the early restoration of gait mechanics (3 months post-ACLR), 2) determine if shank angular velocity during loading response of gait extracted from inertial sensors relates to peak knee extensor moments calculated with a three-dimensional motion capture system, and 3) determine if between limb ratios of shank angular velocity predict between limb ratios of knee extensor moment.

Materials and Methods

Participants

Nineteen participants (14 females, 26.8 ± 12.4 yrs) 96.7 ± 16.8 days post-ACLR were enrolled in this study. A posteriori sample size calculation was performed using the effect size of the relationship between shank angular velocity and knee extensor moment ratios. It was determined that a sample size of 17 was needed to identify a relationship between these variables with 92.4% power.

Surgical reconstruction procedures (allograft, bone-patellar tendon-bone or hamstring autograft) were performed by 7 different surgeons. Participants were included in the study if they were between the ages of 14-55, between 8 and 16 weeks status-post ACLr. They were excluded if they had a concurrent knee pathology that limited the weight bearing status of their surgical limb or a current injury to the contralateral limb that would influence gait.

Instrumentation

Inertial data were collected (Mobility Lab software, APDM Inc., Portland, OR, USA) using two calibrated and synchronized inertial sensors equipped with tri-axial accelerometers, gyroscopes, and magnetometers (Opal brand, APDM Inc., Portland, OR, USA). Angular velocity measured using the gyroscope (128 Hz) was the primary variable of interest. The gyroscope range for X- and Y-axes was ± 34.9 rad/s and Z-axis ± 26.8 rad/s; with noise density in the X- and Y-axes: 0.81 mrad/s/ $\sqrt{\text{Hz}}$; in Z-axis: 2.2 mrad/s/ $\sqrt{\text{Hz}}$, bandwidth: 50 Hz, and resolution: 14 bits. Data from inertial sensors were wirelessly streamed to a laptop through an access point and were buffered on the sensors to avoid data loss in the case of wireless interruption.

Kinematic and ground reaction force data were collected concurrently with inertial data using marker-based motion capture systems. Due to a transition to a new motion capture systems during the study, data were collected using either a 11-camera motion capturing system (Qualysis Inc., Gothenberg, Sweden; 250 Hz) and force platforms (Advanced Mechanical Technologies, Inc., Newton, MA, USA; 1500 Hz) or a 14-camera motion capturing system (BTS Smart-DX motion capture system, BTS Bioengineering Corp., Milan, Italy; 340 Hz) and force platforms (BTS P-6000 DX, BTS Bioengineering Corp., Milan, Italy; 1360 Hz).

Procedures

Testing took place at the Human Performance Laboratory within the Competitive Athlete Training Zone. All procedures were explained to each participant 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 all participants under the age of 18 years.

To define body segments, reflective markers were placed bilaterally over the following landmarks: 1st and 5th metatarsal heads, medial and lateral malleoli and femoral epicondyles, greater trochanters, iliac crests, and posterior superior iliac spines, and the joint space between the fifth lumbar and first sacral spinous processes. Marker clusters, secured to rigid plates were placed bilaterally on the lateral surfaces of the thigh, shank and heel counter of the shoe. Clusters, distal toe, iliac crest, posterior superior iliac spine and lumbar markers remained on the subject during gait analysis; all others were removed following a calibration trial.

Inertial sensors were placed bilaterally on the lateral shanks with the X-axis aligned in the superior-inferior direction and the Z-axis pointing perpendicular from the segment to correspond with sagittal plane motion of the shank (Figure 1). Care was taken to align the X-axis of the shank sensors with the lateral epicondyle and lateral malleoli. For testing, the position of the inertial sensors coincided with the position of the tracking marker clusters; therefore, they were firmly affixed to the rigid plates using elastic Velcro straps and tape.

Participants were instructed to walk at their comfortable pace across a 10-meter path. Self-selected walking velocity was determined using laser timing gates during practice trials, allowing participants to become familiar with the instrumentation. Three acceptable trials for

each limb were collected. A trial was considered acceptable if the participant made full contact with their foot on the force platform and if gait velocity fell within 5% of their self-selected velocity.

Data Processing - Marker based three-dimensional motion analysis

Three-dimensional marker-coordinates were reconstructed (Qualysis Tracking Manager or BTS SMART Tracker) and in combination with ground reaction force data were used to calculate sagittal plane joint kinematics and kinetics (Visual 3D, Version 4.8, C-Motion, Inc., Rockville, MD, USA). Coordinate data was low-pass filtered using a fourth order zero-lag Butterworth filter with a 12-Hz cut-off frequency. Local coordinate systems of the body segments were derived from the standing calibration using a joint coordinate system approach [17]. Lower extremity segments were modeled as a frustra of cones, and the pelvis was modeled as a cylinder. Six degrees-of-freedom of each segment were calculated by transforming the triad of markers on the marker clusters to the position and orientation of each segment during the standing calibration trial. Kinematics, anthropometrics and ground reaction forces were used in standard inverse dynamics equations to calculate internal knee extensor net joint moment [18]. Data obtained were exported and analyzed using a custom MATLAB program (Version R2014b, The MathWorks, Natick, MA, USA).

Peak knee extensor moments were identified during loading response defined as initial contact to the first peak in knee flexion angle (approximately 20% of stance). To characterize symmetry between limbs with respect to knee extensor moment, a between limb ratio of peak knee extensor moments (knee extensor moment surgical / knee extensor moment non-surgical) was calculated. A ratio below 1 indicates a smaller extensor moment in the surgical knee than the non-surgical knee. Three trials were averaged for analysis.

Data Processing - Inertial sensor data

Angular velocity in the Z-axis of the gyroscope was output from the inertial sensors. Prior to processing, signals from the left shank sensor were negated to coincide with the sign convention on the right shank; positive represented counterclockwise rotation. Heel strike and toe off were determined using a previous validated algorithm [13,19]. Raw angular velocity data were filtered using a high-pass IIR filter, with a cut-off frequency of 0.25-Hz to remove drift. A low-pass filter with a cut-off frequency of 35-Hz was applied to remove spurious peaks. From filtered data, peak positive shank angular velocity, signifying mid-swing, was identified. The negative peaks just prior to and after the peak positive angular velocity were used to identify toe off and heel strike, respectively (Figure 2). Stance time was calculated as the time between heel strike and toe off of the same limb; swing time was calculated as the time between toe off and heel strike of the same limb.

Next, raw angular velocity data were filtered using a high-pass IIR filter with a cut-off frequency of 0.25-Hz and a low-pass filtered with a cut-off frequency of 6-Hz to determine peak shank angular velocity following heel strike. A 6-Hz cut-off was determined from power spectrum analysis of the raw angular velocity data. Peak shank angular velocity during loading response was identified as the first negative peak in the Z-axis (Figure 3) in the surgical and non-surgical limbs. A between limb ratio of peak shank angular velocity

(shank angular velocity surgical / shank angular velocity non-surgical) was calculated. Peak shank angular velocity was calculated for each limb from three consecutive steps of each trial. Data from three trials were averaged together for analysis.

Statistical analysis

Paired t-tests were performed to determine between limbs differences in stance time, swing time, peak knee extensor moment and shank angular velocity during loading response of gait. Pearson's product-moment correlation was used to investigate the relationship between knee extensor moment and shank angular velocity during loading response of gait; data from both limbs were considered for analysis. Simple linear regression was used to determine if between limb ratios of shank angular velocity predicted between limb ratios of knee extensor moment ratio. Significance level was set at $\alpha = .05$ (PASW Statistics, Version 18, IBM Corp., Chicago, IL, USA). Statistical power for the linear regression analysis and posteriori power analyses were performed using G*power 3.1 (Dusseldorf, Germany).

Results

No differences were noted between limbs with respect to stance ($p = 0.132$) and swing ($p = 0.840$) times (Table 1). During loading response, peak knee extensor moment was significantly smaller in the surgical knee compared to the non-surgical knee ($p < 0.001$; Table 1) and peak shank angular velocity was smaller in the surgical knee compared to the non-surgical knee ($p < 0.001$; Table 1).

Knee extensor moment and shank angular velocity ratios were, on average, 0.71 ± 0.33 and 0.83 ± 0.12 and ranged from -0.01 to 1.43 and 0.62 to 1.10 , respectively. A strong positive correlation ($r = 0.756$, $p < 0.001$; power = 0.924) between knee extensor moment and shank angular velocity. Greater shank angular velocity was related to greater knee extensor moment. Linear regression indicated that between limb ratios of shank angular velocity were predictive of between limb ratios of knee extensor moment. Shank angular velocity ratio explained 57.5% ($p < 0.001$) of the variance in knee extensor moment ratio.

Discussion

Direct outputs from inertial sensors provide information regarding segment kinematics that have been successfully used to determine spatiotemporal characteristics of gait and to differentiate kinematic differences between healthy individuals and those who had previously undergone ACLr [13]. The data presented in this study establish an important link between the kinematic differences detected using inertial sensors during gait and knee loading impairments calculated using marker-based motion analysis and force platforms. This is an important step toward developing clinical gait analysis procedures capable of providing relevant information regarding loading deficits in individuals who have recently undergone ACLr and have normalized gait.

The normalization of gait following ACLr typically occurs 8-12 weeks following surgery and is used as a benchmark for rehabilitation progress and a criteria for advancing to more demanding exercises [20,21]. It has become clear that restoration of gait without observable

deviations does not translate into the return of symmetrical knee loading mechanics for all individuals following ACLr. This is underscored in the current study as knee extensor moments were, on average, 31% smaller in the surgical compared to the non-surgical knee, while no between limbs differences in stance or swing times were detected. Normalization of the timing of stance and swing phases likely contributes to the appearance of unimpaired gait. Use of these procedures to detect spatiotemporal differences in this population may not be relevant. However, significant differences were observed between limbs in shank angular velocity with smaller peak angular velocity in the surgical limb. These data are consistent with previous analyses of individuals approximately 3 years post-ACLR [13] suggesting that shank angular velocity may be an important indicator of altered gait mechanics in this population.

Knee extensor moments were correlated with shank angular velocities measured with inertial sensor gyroscopes. Peak shank angular velocity represents the rate of anterior shank rotation during loading response, a critical feature of heel rocker mechanics. Rapid anterior rotation of the shank occurs as the foot is being lowered to the floor; the shank rotates faster than the femur can advance, resulting in knee flexion, which coincides with a peak in knee extensor moment shortly after heel contact (Figure 4). Faster shank rotation is related to a greater knee extensor moment. The strength of the correlation between knee extensor moment and shank angular velocity in this study supports the relationship between heel rocker mechanics and knee extensor moment.

Between limb ratios of shank angular velocity were predictive of between limb ratios of knee extensor moment, explaining 57.5% of the variance in knee extensor moment ratio. These data suggest that modulation of heel rocker mechanics, specifically rotation of the tibia over the foot, contributes to the reduced knee extensor moments observed during gait given that other ground reaction forces, used to calculate knee joint moments, were not considered. It is likely that ground reaction forces contribute to the 42% of the variance not explained by shank angular velocity.

Using between limb ratios, referencing the non-surgical limb, for assessment of gait mechanics is important given that the magnitude of both extensor moments and angular velocities during gait are related to walking velocity and likely influenced by other factors such as shoe wear or walking surface. This limits our ability to use a gold standard or reference of shank angular velocity to indicate the restoration of heel rocker mechanics. While the assumption that the non-surgical limb demonstrates normal gait mechanics may not be accurate, it likely provides the best frame of reference. Between limb asymmetries in knee extensor moments have been identified at 6, 12 and 24 months post-ACLR [2-4]; the persistence of these deficits may be due to the inability to detect them clinically or the assumption that normal gait has been restored. Despite the small sample size in the present study, we found that relationship between knee extensor moment and shank angular velocity ratios (observed power = .924) is promising for the translation of these procedures to be used for clinical identification of impaired knee loading in this population. Early detection of these between limb deficits using inertial sensors may help clinicians target their interventions accordingly.

Conclusions

Together, these data indicate that kinematic features of the shank can provide information relevant to knee loading in individuals following ACLr as they restore gait. Between limb differences in shank angular velocity are related to between limb differences in knee extensor moment at a time during which spatiotemporal asymmetries are not detected. Clinicians rely on the observation of gait deviations to determine normalization of gait following ACLr. The use of inertial sensors for quantification of subtle differences in shank angular velocity may improve their ability to detect and address mechanical loading deficits. Determination of the diagnostic accuracy of such procedures in a larger subject population that includes individuals across time periods of recovery is needed for translation of these findings for clinical use.

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Highlights

Difficulty identifying knee loading deficit may underline their persistence.

Altered knee loading is related to altered heel rocker mechanics during gait.

Shank angular velocity plays an important role in calculating knee extensor moment.

Angular velocity between-limb ratios predict knee extensor moment deficits.

Wearable sensor gyroscopes detect gait impairments following ACL reconstruction.

Figure 1. Placement of inertial sensors. Sensor axes indicated by red arrows. Orientation of axes depicted on the bottom right.

Figure 2.

Spatiotemporal parameters identified from shank angular velocity filtered at 35-Hz. Peak positive shank angular velocity signifies mid-swing. Negative peak shank angular velocity prior to and following mid-swing were used to identify toe off (red star) and heel strike (red circle), respectively.

Figure 3.

Shank angular velocity around the Z-axis corresponding to rotation of the shank in the sagittal plane. Asterisks indicate the time of heel strike in non-surgical (solid red line) and surgical (dashed red line) limbs. Peaks after heel strike (black arrows) were identified from shank angular velocity filtered at 6-Hz.

Figure 4. Illustration of heel rocker mechanics which are characterized by rapid forward rotation of the shank over the heel following ground contact; resulting in knee flexion controlled eccentrically by the quadriceps.

Table 1

Between limb comparisons.

	Surgical Limb	Non-surgical Limb	Sig.
Stance phase (s)	0.573 $\pm$ 0.050	0.580 $\pm$ 0.060	0.132
Swing phase (s)	0.447 $\pm$ 0.034	0.446 $\pm$ 0.035	0.840
Knee extensor moment (Nm/kg)	-0.646 $\pm$ 0.400	-0.926 $\pm$ 0.370	<0.001 *
Shank angular velocity ($^{\circ}$ /s)	-155.56 $\pm$ 26.79	-177.85 $\pm$ 23.89	<0.001 *

Note:

* significant differences between limbs. All values reported are mean $\pm$ SD.