Measurement of In Vivo ACL Elongation During Gait

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Introduction

Over 200,000 ACL injuries occur in the United States every year, half of which are experienced by young athletes between 15 and 25 years of age. The consequences of ACL deficiency include pain, instability, damage to the menisci and early-onset osteoarthritis (OA). Furthermore, it has been suggested that current reconstructive techniques do not decrease the probability of developing OA when compared to non-operative treatment. An understanding of normal ACL function would be beneficial to solving this clinical issue by providing insight on how daily activities load the ligament. Several cadaveric studies have previously characterized ACL mechanics, but they are likely limited by an inability to replicate the complex loading environments experienced in vivo. Furthermore, there are limited data on in vivo elongation of the ACL during overground walking. Therefore, the objective of this study was to characterize the in vivo ACL elongation during normal gait using a new method developed by our lab.

Methods

Five healthy male subjects (mean age, 26.2 years; range, 21-32 years) without previous history of knee injury were included in this study. MR images of each subject’s knee were taken using a 3T magnet (Trio Tim, Siemens), a double-echo steady state sequence (flip angle: 25°, TR: 17 ms, TE: 6 ms) with a 15x15cm field of view, and an 8 channel receive-only knee coil (Invivo, Orlando, FL). Image resolution was 512x512 pixels at a 1mm slice thickness. From these images, the cortical bone and attachment site of the ACL were outlined using solid-modeling software (Rhinoceros, Robert McNeel and Associates, Seattle, WA) as described and validated previously. 3D models of each tested knee were created from these collective tracings.

3D motion analysis using an 8-camera motion capture system (240 Hz) (Motion Analysis Corporation, Santa Rosa, CA) and three embedded force plates (2400 Hz) (AMTI, Boston, MA) was next performed to collect each subject’s normal gait. Reflective markers were placed unilaterally on anatomic landmarks and in non-symmetric clusters across the leg until a total of 44 markers were positioned. A successful trial was classified when a subject’s marked leg stepped cleanly onto a single force plate without interference from the other foot. Immediately after gait analysis, biplanar fluoroscopic images were taken with the markers still in place. This allowed for the position of the markers to be registered to the underlying bone using a shape-matching technique. Numerical optimization was utilized to first register the fluoroscopic marker data to that of a static standing trial obtained during motion analysis by rigidly transforming one bone at a time. After this initialization, similar optimizations using subsets of the total marker count which define each individual bone were performed. This step minimized error associated with soft tissue motion, similar to techniques shown by other investigators. 3D dynamic in vivo knee models were created by superimposing the optimized gait kinematics onto the knee model. ACL length was defined as the distance between the area centroids of each attachment site. ACL length, knee flexion, and ground reaction force (GRF) were measured and averaged across a gait cycle. During a lunge activity, this system measured ACL length to within an average of 0.5mm compared to that measured using biplanar fluoroscopy.

Results

The results of the normal gait trials are depicted in Figure 1. Gait cycle was normalized for all subjects as the motions between subsequent heel strikes on the same leg. During the gait cycle, flexion increased initially to midstance where GRF reached a local maximum with weight acceptance. Flexion then decreased with the knee extending until the start of terminal extension. During this time, the contralateral leg is in its swing phase. To finish the cycle, flexion increases then decreases quickly as the leg pushes off the ground through toe off and enters swing phase. In general, the length of the ACL during gait decreased with increasing flexion angle. ACL length peaked at 13.5% elongation approximately 60% of the way through stance phase, defined as the time between heel strike and toe off.

Discussion

In this study, the observed flexion angle and ACL elongation trends were consistent with previously described patterns during the stance phase of treadmill gait. However, the present study provides additional data on in vivo ACL deformations during the swing phase. Most interesting was the timing of peak ACL lengthening, which occurred halfway through the stance phase despite non-minimal flexion angles. We hypothesize that this pattern is due to an increase in quadriceps activation coinciding with a time when single leg stance begins and the opposite leg is in swing phase. Quadriceps contractions are known to load the ACL at low flexion angles due to the anterior shear component of the patellar tendon translating the tibia forward. A future study with EMG data could confirm these findings. This novel approach provides a new means of measuring in vivo kinematic patterns and ACL strains during unrestricted dynamic movements. Future studies utilizing this technique will provide new insights about ACL and other soft tissue function during normal dynamic activities.

Figure 1. Data were described over a single gait cycle between subsequent foot strikes. ACL length peaked around 13% elongation during the single-legged portion of stance phase. Stance phase was defined as the time between heel strike and toe off.


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