THEORETICAL ANALYSIS OF THE FLEXED KNEE PATTERN IN ACL-DEFICIENT GAIT

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INTRODUCTION

The functional gait adaptations of anterior-cruciate-ligament-deficient (ACLD) individuals have been well documented [1-4]. Gait analysis experiments have shown that ACLD patients generally walk with a more flexed knee pattern and with an elevated or prolonged duration of hamstring muscular activity during the stance phase [1,5]. It is speculated that increased knee flexion angle is an adaptive strategy designed to increase the posterior shear force applied by the hamstrings, thereby reducing anterior tibial translation (ATT). Increased EMG levels observed in the hamstrings is also thought to be a compensatory strategy for reducing excessive ATT. It is also plausible that an increase in knee flexion angle may decrease the angle of the patellar tendon relative to the tibia, thereby reducing the anterior shear force applied by the quadriceps. While these ideas have some intuitive appeal, they have yet to be rigorously tested.

The purpose of this study was to determine how walking with a flexed knee pattern affects ATT in the ACLD knee. Previously, we used musculoskeletal modeling and computer simulation to calculate ligament forces in the intact knee during level walking [6]. Here we use similar methods to investigate the flexed knee gait compensation pattern often observed during the stance phase of ACLD gait. Our specific aims are to: (a) describe the change in ATT following removal of the ACL, (b) predict the change in ATT when knee flexion angle is increased throughout the stance phase of gait, and (c) find how increased knee flexion affects the amount of additional hamstrings force needed to bring ATT of the ACLD knee back to normal.

METHODS

The right leg was modeled using four rigid bodies: thigh, shank, hindfoot, and metatarsals (Figure 1). All joints were represented as described in [7], except the knee, which was modeled as a six degree-of-freedom joint. The geometry of the distal femur, proximal tibia, and patella was based on cadaver data reported for an average-size knee. The contacting surfaces of the femur and tibia were modeled as deformable, while those of the femur and patella were assumed to be rigid. Thirteen elastic elements were used to describe the geometric and mechanical properties of the knee ligaments and posterior capsule [8]. The model leg was actuated by thirteen musculotendinous units. A combination of forward and inverse dynamics was used to determine the relative translations of the bones during the stance phase of gait. Joint motion, ground-reaction forces, and the corresponding muscle forces obtained from a dynamic optimization solution for gait [7] were input to the leg model. The dynamical equations of motion

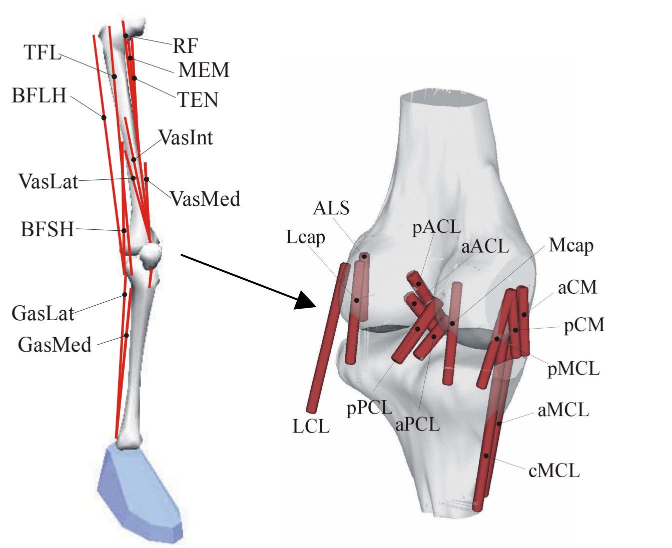

Figure 1. The muscles of the leg were modeled by a total of thirteen actuators, each consisting of a straight-line segment joining the three-dimensional attachment sites except were muscle wrapped around bone or other muscle. The passive structures of the tibiofemoral joint were modeled by a total of thirteen elastic bundles, each consisting of a straight-line segment joining the three-dimensional attachment sites.

for the three-dimensional knee model were then used to determine the ATT induced at each instant of the stance phase of gait [6]. Four simulations were performed: a) ACL-intact with a normal pattern of knee flexion, b) ACLD with a normal pattern of knee flexion, c) ACLD with 5 degrees added to the normal pattern of knee flexion throughout stance, and d) ACLD with 10 degrees added to the normal pattern of knee flexion throughout stance. The addition of 5 and 10 degrees to the normal knee flexion pattern was based on gait analysis results reported by others [1,5]. Simulations b, c, and d were repeated to determine the amount of hamstrings force needed to reduce ATT in the ACLD knee (blue line in Figure 2) to the amount calculated for the ACL-intact knee (black line in Figure 2).

RESULTS

Consistent with trends reported previously by others, ATT increased throughout the stance phase of gait when the ACL was removed (blue line in Figure 2). Increasing knee flexion significantly reduced ATT of the ACLD knee only at heel-strike (red and green lines in Figure 2). An increase in knee flexion decreased the amount of hamstrings force needed to keep ATT the same as that calculated for the intact knee (Figure 3, red and green lines).

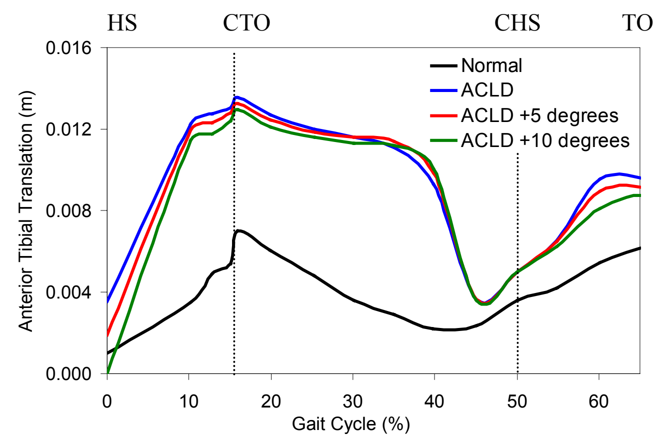

Figure 2. ATT during level walking for four conditions: ACL-intact, ACL-deficient, ACL-deficient with 5 degrees added to knee flexion, and ACL-deficient with 10 degrees added to knee flexion (HS – heel strike, CTO – contralateral toe off, CHS – contralateral heel strike, TO – toe off).

CONCLUSIONS

The results of Figure 2 show that increasing knee angle alone has little effect as a compensatory strategy to limit ATT in ACLD walking. This prediction of the model opposes the idea that a flexed knee pattern during gait causes a reduction in the anterior shear force applied to the tibia by the patellar tendon. For instance, although quadriceps force was highest at CTO [6], there is little change in net ATT in Figure 2 (compare blue, red, and green lines at CTO). When the ACL was removed in the model, the hamstrings force generated during normal walking (Fig. 3, black line) was insufficient to keep ATT the same as that calculated in normal gait. The amount of hamstrings force needed to keep ATT the same as that in the intact knee was much larger in the ACLD knee (Fig. 3, blue line). Increasing knee flexion reduced this force significantly (Fig. 3, red and green lines). Thus, our modeling results support the view that ACLD patients who walk with flexed knees do so because less hamstrings force is then needed to stabilize the ACLD knee in stance.

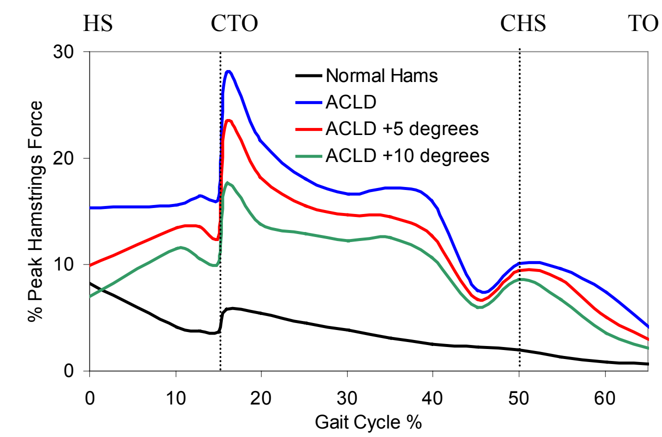

Figure 3. The percent of peak isometric hamstrings force for the ACL-intact knee during normal gait (black line) and that required to bring ATT of the ACL-deficient knee (blue line in Figure 2) back to normal (black line in Figure 2) with increasing levels of knee flexion.

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