Comparison of Ligament Loading and Anterior Tibial Translation in Healthy and ACL-Deficient Knees During Gait

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INTRODUCTION:
The ACL is the most frequently injured ligament of the knee. To develop better surgical procedures and rehabilitation regimens for ACL-Deficient (ACLD) patients, it is of great importance to know internal knee-ligament loading. A few numerical models have been used to calculate ligament loading in the knee joint during gait (Shelburne et al., 2004). However these models were not driven by measured muscle activations, which may vary from patient to patient and are difficult to predict using these models.

In this report we describe for the first time an EMG-driven model that incorporates a knee-ligament model, and we apply this approach in a pilot study to estimate anterior tibial translation (ATT), anterior shear forces and ligament loading in the knee joint during gait of a patient without an ACL and a healthy control.

DATA COLLECTION:
One healthy subject and one ACLD patient who gave informed consent were included in this study. The experimental protocol was approved by the Human Subjects Review Board of University of Delaware. EMG, joint position and force plate data were collected from four walking trials. EMGs were collected from nine muscles of the right leg (MG, LG, Sol, TA, RF, VM, VL, BFL, SM/ST) using surface electrodes. Maximum voluntary contraction trials were collected for normalization of EMG. In this study we simulated the stance phase of the right leg during the walking trials.

METHODS:
We calculated knee-ligament forces through a two-step procedure. First, an EMG-driven model (Buchanan et al., 2004) was used to estimate the muscle forces of the right leg to match the inverse dynamics calculated joint moments from heel strike (HS) to toe off (TO).

Second, another knee model that incorporated a knee-ligament model was developed to calculate the ligament loading. The model included the tibiofemoral and patellofemoral joints in the sagittal plane, and three segments: the femur, tibia and patella. The muscle forces and joint reaction forces calculated from the previous step were used as input. The knee-ligament model was used to calculate ligament forces. In the knee-ligament model, each knee ligament was composed of a number of bundles (Shelburne et al., 2004), and each ligament bundle was modeled as a nonlinear elastic element, described by a nonlinear force-strain relationship. The approach used to model the patellofemoral joint in this study was similar to that used by Liu and Maitland (2000), through which patella ligament force was calculated (Fig. 1A). The contact and geometric compatibility conditions were required to be satisfied at the tibiofemoral joint. The translations between femur and tibia, knee joint contact force and ligament forces were solved through iterations until the force equilibrium of the tibia were satisfied (Fig. 1B).

![Figure 1. The patellofemoral and tibiofemoral joint models.](Image)

RESULTS:
The ACLD knee had increased ATT compared to the healthy knee, with a maximal difference of 9.1 mm near HS (Fig. 2). The ACL of the healthy knee contributed to balance the anterior shear force (Fig. 3), while the MCL of the ACLD knee contributed to balance the anterior shear force instead of the ruptured ACL (Fig. 4).

DISCUSSION:
We have developed a biomechanical model that uses EMGs as input to estimate the translations between femur and tibia, shear forces and ligament loading. ATT increased throughout the stance phase for the ACLD patient compared with the healthy control, which is consistent with previous in vivo and in vitro studies. The MCL was the main passive restraint to anterior shear force in the ACLD knee, and this finding is consistent with previous simulation and in vitro studies.

This is the first time that an EMG-driven model has been used to estimate ligament loading in ACLD patient, and the results provided insight on how ACLD patient compensate for the loss of ACL. In future studies we will recruit more subjects, compare the calculated results with those from previous in vivo experiments, and explore differences in ligament biomechanics associated with different rehabilitation protocols.

REFERENCES:

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