AN EMG DRIVEN MODELTO ESTIMATE ACL FORCES DURING NORMAL WALKING

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INTRODUCTION

The ACL is the most frequently injured knee ligaments. To develop better surgical procedures and rehabilitation regimens for ACL deficient patients, it is of great importance to know internal knee-ligament loading. A few numerical models have been used to calculate ACL forces during gait [1, 2]. However these models were not driven by measured muscle activations, which may vary from patient to patient.

In this paper we describe an EMG-driven model that incorporates a knee-ligament model, and we apply this approach to estimate ACL forces during normal gait.

METHODS

Five young healthy subjects who gave informed consent were included in this study. EMGs, joint positions and force plate data were collected from four walking trials. EMGs were collected from the major muscle groups of the right leg (MG, LG, Sol, TA, RF, VM/VL/VI, BFL/BFS, SM/ST) using surface electrodes. MVC trials were collected for normalization of EMG. The subjects were required to strike force plate 1 using their right foot and force plate 2 using their left foot. In this study we simulated the stance phase of the walking trials.

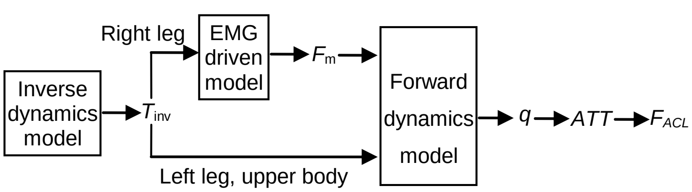

Figure 1: Flow chart of using the forward dynamics model to estimate ACL force.

We calculated ACL forces through a two-step procedure (Figure 1).

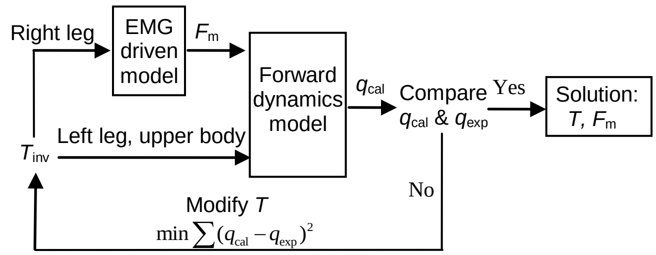

Figure 2: Flow chart of the forward dynamics model.

First, a forward dynamics model of the whole body was developed to calculate muscle forces, joint torques and joint angles, and the model was verified by successfully replicating experimental joint angles. The musculoskeletal model was constructed using SIMM (Motion Lab Systems, Inc., Baton Rouge, LA). It included 11 segments: femur, tibia/fibula, patella, talus, and calcn of both limbs, back. Here the back segment included the mass and inertial properties of the pelvis, torso, head, and arms. This model had 9 DOF in sagittal plane: horizontal and vertical position of pelvis, pelvis rotation, hip, knee and ankle extension/flexion of both limbs. The equations of motion were solved using SD/FAST (Symbolic Dynamics, Inc., Mountain View, CA). Muscle forces, Fm, calculated from an EMG-driven model [3, 4] were used to drive the model spanning the right ankle and knee joint. For the other joints, the model was driven by the joint torques calculated using inverse dynamics, T_{inv}. T_{inv} were calculated using dynamic optimization of the inverse dynamics model, with an emphasis on reducing residual forces and torques at the pelvis. F_m were calculated through our EMG-driven model to match the calculated T_{inv}. Ground reaction forces during the simulation were prescribed to the experimentally recorded values. Since T_{inv} may not necessarily drive a successful forward simulation as shown by others [5, 6], we used an optimization to find a solution of joint torques, T, that could successfully reproduce experimental kinematics during forward simulation (Figure 2). T_{inv} were used as initial values during the optimization. After solution was obtained, it could be used to drive a forward simulation and track the experimental kinematics.

Second, the forward dynamics model was incorporated with a knee-ligament model. The muscle forces and joint reaction forces/torques calculated from the previous step were used as inputs for the forward dynamics model. At each time step, we used SD/FAST to solve the translations between femur and tibia that could hold the segments in the calculated joint positions. The knee-ligament model was then used to calculate ligament forces. In the knee-ligament model, each knee ligament was composed of a number of bundles [2], and each ligament bundle was modeled as a nonlinear elastic element, described by a nonlinear force-strain relationship [7]. After the translations between femur and tibia were calculated, the ligament length was determined, and the ligament force was calculated using the force-strain relationship.

RESULTS AND DISCUSSION

The forward dynamic right knee and ankle joint moments calculated from the EMG-driven model matched the inverse dynamic joint moments (Figure 3). The calculated right hip, knee and ankle joint angles matched the measured angles (Figure 4). There were two ACL force peaks near LTO and RTO (Figure 5).

CONCLUSIONS

We have developed a forward dynamics model that uses EMGs as input to estimate ACL forces during normal stance phase. It successfully tracked measured kinematics. The model could also be implemented to calculate the translations between femur and tibia. The calculated result of two ACL force peaks near LTO and RTO was consistent with previous studies [1, 2], and could be explained by the onset of two GRF peaks during stance phase.

In future studies we will compare our simulation results with the results from *in vivo* studies, and apply this model to patients with pathologies, which could provide insight of patients' knee ligament loading to clinicians. We will also explore differences in ligament biomechanics associated with different rehabilitation protocols.

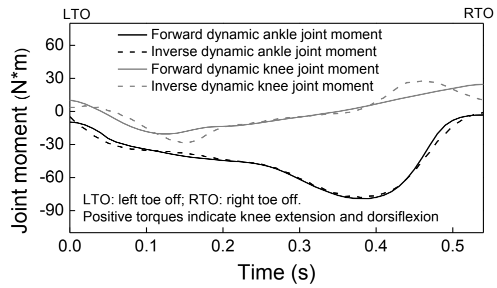

Figure 3: Forward and inverse dynamic joint moment profiles.

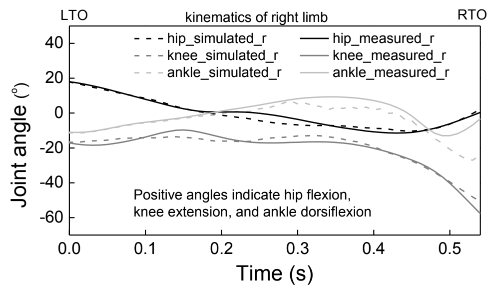

Figure 4: Simulated joint kinematics and measured kinematics.

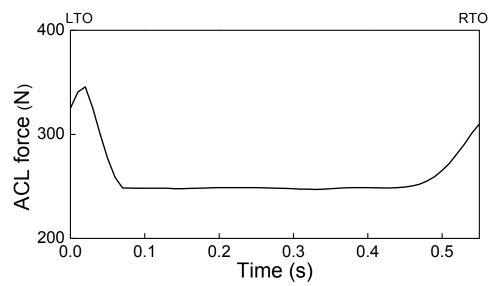

Figure 5: ACL force profile during stance phase.

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