DEVELOPING AND TESTING OF AN EMG DRIVEN MODEL TO ESTIMATE ANKLE MOMENTS AND MUSCLE FORCES

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INTRODUCTION

Estimation of individual muscle force and its contribution to a joint moment has an important role in understanding injuries and disease. In previous work, we have developed EMG-driven models to estimate isometric forces in muscles about the elbow (Manal et al., 2002). Similarly, Lloyd and Besier (2003) used EMG-driven models to predict dynamic flexion and extension knee moments for a varied range of tasks. This approach has the advantage of estimating muscle force *in vivo* using EMG signals and an anatomical muscle model as input, and thereby relies on a person's particular neural control strategy rather than an assumed cost function. Developing the anatomical model is a fundamental step towards good force prediction. The more detailed the model, the more it is able to predict a variety of muscle contractions. Practical limitations, such as the ability to get EMG signals from all muscles, generally leads to a simplification of the model.

The purpose of this work is to develop and test an EMG driven model to predict dynamic muscle forces in the ankle joint. This model will be used in future studies to examine ankle muscle contributions to gait in patients with neuromuscular impairments.

METHODS

We modeled the ankle as a one degree of freedom hinge, with rotation allowed only in the sagital plane. The model includes four primary muscles: soleus and two heads of gastrocnemious as plantar flexors, and tibialis anterior as the dorsiflexor. Hill-type muscle models were used to account for each muscle's force-length and force-velocity relations. The transformation of EMG to muscle activation was calculated using the Lloyd and Besier (2003) EMG to activation model in combination with the terms introduced by the Manal & Buchanan (2003) model for the neural to muscle activation step. Delp et al,'s (1990) SIMM modeling software and the associated lower extremity model were used to calculate the muscle-tendon lengths and moment arms. As an initial step, optimal fiber length, pennation angle and tendon slack length were set to the default values specified in the SIMM lower limb model.

EMG signals were collected from healthy subjects, then rectified, filtered, and normalized relative to maximum EMGs. The tasks included: dynamometer trials and walking trials, where for the later, inverse dynamics was used to calculate ankle joint moments. Range of motion for the dynamometer trials was 5^o to 20^o, dorsiflexion to plantar flexion, which is the average range during gait.

Calibration of the model was accomplished by simulated annealing. Upper and lower bounds for the optimization process were set to keep the muscle parameters within the physiological range. Passive moments that were measured during passive trials were used to constrain passives forces in muscles.

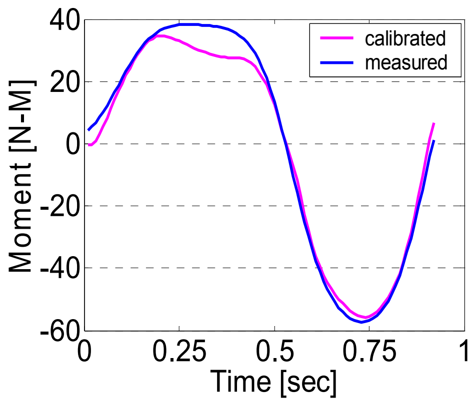

Figure 1: The model was calibrated by adjusting parameters until the estimated (calibrated) joint moments best matched the measured moments.

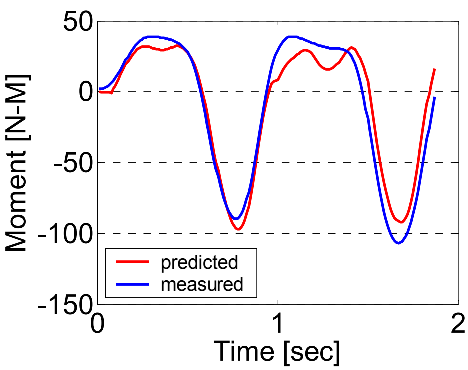

Figure 2: Prediction of ankle moments for a novel trial following calibration on the trial shown in Figure 1.

RESULTS AND DISCUSSION

Calibration of the model on a dynamometer trial is shown in Figure 1 (RMS error is 4.5 N-m and R^2=0.99). Figure 2 shows predicted moment on another dynamometer trial (RMS error is 12 N-m, R^2=0.95).

The maximum isometric muscles forces estimated by the model were much higher values than those in SIMM, with an average increase of about 200%. This is because our subjects had greater maximal joint moments when measured using the dynamometer and because we modeled the ankle using a subset of the muscles instead of the full thirteen muscles in the SIMM model.

The ankle muscles tested are characterized by relatively high pennation angles, for which values vary widely in the literature. We suggest *in vivo* measurement of the pennation angles for the specific subjects. For example, using ultrasound we found that the pennation angle for the tibialis anterior was 10^{o}, compare to the 5^{o} we used based on the model in SIMM.

Finally, only four muscles were used in this model to represent the ankle joint musculature. This simplification may require the use of "muscles" with slightly different parameters to compensate for the excluded muscles.

SUMMARY

The model presented has demonstrated its ability to estimate ankle muscle moments in static and dynamic trials. With better subject specific muscle parameters we expect more accurate predictions that will allow us to use this to better understand muscle control in people with neuromuscular impairments.

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ACKNOWLEDGEMENTS

This work was supported, in part, by NIH R01 HD38582.