Three-dimensional Muscle Models of the Human Triceps Surae Muscle-Tendon Complex

D. Shin[1], S-W. Chi[1], J-S. Chen[1], J. Hodgson[1], V. Edgerton[1], and S. Sinha[2]
[1]University of California, Los Angeles, CA, United States, [2]University of California, San Diego, CA, United States

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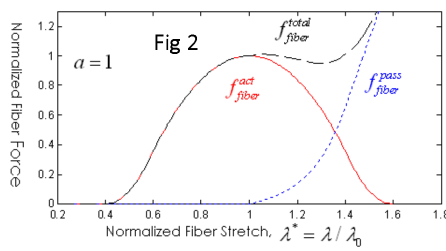
Introduction:
There is a controversy regarding the nature of strain distribution along the aponeurosis-tendon complex. Different investigators have reported quite different results. Our group has reported inhomogeneous stain distribution in the human lower leg as experimentally recorded by the phase contrast MRI technique. We hypothesize that a good numerical model can provide additional insights regarding the complex underlying mechanics that underpin this phenomenon. Last year, we have reported a validation study that compared the load and time displacement response of the human triceps surae aponeurosis measured with PC-MRI and the FE model and reported good agreement between the two[1]. However, the study was confined to the free tendon region where muscle fibers were not attached since the FE model only accounted for the aponeurosis. In our improved model, we have added soleus muscle as an additional mesh to the FE model as well as fibers at 25 degrees from the longitudinal axis. Our FE simulation results have shown similarly complex yet heterogeneous strain distribution along the aponeurosis at different MVC levels.

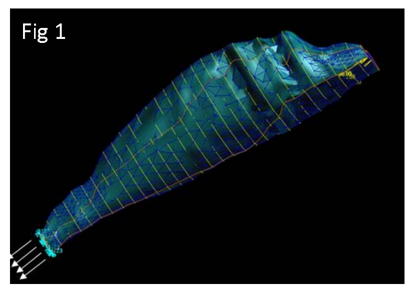
Materials and Methods:
The FE model consists of two pieces which simulate the aponeurosis (the shaded part) and soleus (Fig. 1). Since the muscle-tendon complex undergoes large deformations, we implement hyperelastic models to describe the stress-strain behavior of the aponeurosis and soleus. We assumed that material behavior of the aponeurosis is isotropic and homogenous and the constitutive relationship can be characterized by a cubic form of strain energy function, given in Eq. (1). The cubic model is capable to simulate the well known force-length behavior of the aponeurosis, which shows that it is more compliant at low-force levels and stiffer at the high-force levels. In the muscle portion, we consider the muscle active and passive force-length characteristics, plotted in Fig. 2, and add them into a transversely isotropic hyperelastic model, given in Eq. (2). The voluntary contraction can be modeled by rising up the activation levels.

$$W = G_{10}\left(\bar{I}_1 - 3\right) + G_{20}\left(\bar{I}_1 - 3\right)^2 + G_{30}\left(\bar{I}_1 - 3\right)^3 + \frac{K}{2}\left(J-1\right)^2 \quad (1)$$

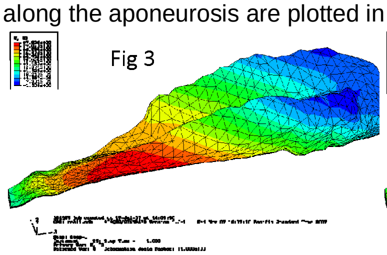
$$W = C_{10}\left(\bar{I}_1 - 3\right) + C_{01}\left(\bar{I}_2 - 3\right) + \bar{W}_{fiber}\left(\lambda\right) + \frac{K}{2}\left(\ln\left(J\right)\right)^2 \quad (2)$$

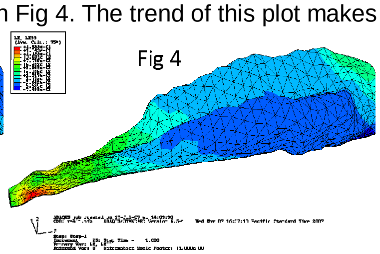
The 3D geometry of the aponeurosis (the shaded part) and soleus was generated using a stack of axial slices, 7 mm think, acquired with a T2 weighted TSE sequence across the entire length of the leg after manually segmenting the aponeurosis boundaries and soleus boundaries from each slice and smoothing the curves . The FE mesh was generated in MSC.PATRAN and both distal and proximal ends of this model were fixed to simulate the isometric plantar flexion. We assume muscle fibers are arranged in parallel at 25 degrees form the longitudinal axis and have the same activation level.

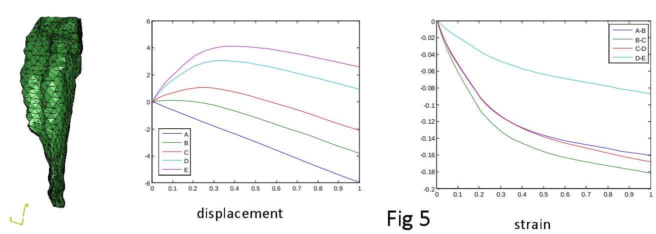




Results and Discussion:
To simulate the voluntary contraction under isometric condition, the two ends of the tendons are fixed to keep the total length of this model unchanged. Muscle fibers contract and generate forces when the activation level rises. The heterogeneous strain distribution can be understood from the longitudinal displacement contour (Fig. 3) and the longitudinal strain contour (Fig. 4). We also examined the hypothesis that during isometric contractions, the aponeurosis length change is non-uniform by numerical simulation. We introduced 2 mm movement at the distal end of this model to mimic slight angle movement. The longitudinal displacements and strains at points along the aponeurosis are plotted in Fig 4. The trend of this plot makes an agreement with experimental observation.







Conclusions:
Non-uniform strain distribution was displayed from the output of the FE mode, which is in agreement with our previously reported PC-MRI data. Based on this computational model, we hypothesize that the combination of active and passive behavior, geometrical factor as well as ankle motion contributes to non-uniform strain distribution even when the material properties are assumed to be constant.

References: 1. D Shin et al. Presented at Fifteenth ISMRM Meeting in 2007 (Abstract #3991)