MUSCLE-DRIVEN FINITE ELEMENT SIMULATION OF HUMAN FOOT MOVEMENTS

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INTRODUCTION

Human foot is of special interest because it is the primary physical interaction between the body and the ground during locomotion. Foot and lower-leg structures, such as bones, ligaments, and muscle-tendon systems, interact in order to develop various foot movements. Computer models that focus on muscle-driven human locomotion are based mainly on rigid-body dynamics and represent muscles as a series of line-segments, thus being unable to predict local tissue loadings, such as internal stresses and strains, shape changes of soft tissues, such as muscle deformation during contraction, as well as interactions of musculotendinous units with the rest surrounding soft tissues and bones during human movement.

In this work, an attempt is made to produce a natural human foot movement by incorporating realistic geometrical characteristics and material properties for muscles and tendons in a three-dimensional anatomically detailed finite element model of foot and lower-leg. The advantage of the current approach is that a simulation of foot movement is accompanied by an estimation of local internal loadings in each tissue and prediction of their shape deformation.

METHODS

CT scans with intervals of 1 mm (nearly 500 images) were chosen to represent the geometry of the right foot and lower-leg of a normal female individual of age 28 in the neutral foot position. Three-dimensional data sets are acquired and segmented using the AMIRA v4.1 software in order to describe the boundaries of skeleton, muscles, tendons and skin surface. Three-dimensional surfaces are calculated directly from reconstruction of the segmented images. Each surface is described by a set of triangles in a three-dimensional space. The desirable volumetric mesh results by filling with tetrahedra the volume enclosed by each surface. The model used in the current analysis consists of 51,113 nodes and 271,622 four-node tetrahedral first order solid elements, with a total of 153,339 degrees of freedom (Figure 1).

The tissues included in the model are assigned specific material characteristics. In the present study, skeletal muscles and tendons are considered as three-dimensional continuum composite materials that consist of fibers surrounded by connective tissue and biofluids. The constitutive model used for these materials was developed recently [3] based on an idea put forth for the mechanical behavior of muscular hydrostats [1].

The nominal longitudinal stress in a muscle fiber $\sigma_0^m$ has an active and a passive part and is written in the form

$\sigma_0^m = \sigma_0^{m(\text{act})} + \sigma_0^{m(\text{pas})}$

$\sigma_0^{m(\text{act})} = \sigma_{\text{max}} f_a \left( \varepsilon_0^m , f_a \right) f_r \left( \varepsilon_0^m \right)$

$\sigma_0^{m(\text{pas})} = \begin{cases} 
\sigma_{\text{max}} f_p \left( \varepsilon_0^m , f_a \right) & \text{for } f_a > 0 \\
\sigma_{\text{max}} f_p \left( \varepsilon_0^m , f_p \right) & \text{for } f_a = 0
\end{cases}$

In above equations $\sigma_{\text{max}}$ is the maximum isometric stress at optimum fiber length, $f_a$ is the activation state that describes the pattern of the activation signal as a function of time, $f_r$ describes the dependence of the active stress on the nominal longitudinal strain $\varepsilon_0^m = \exp(\varepsilon_0^m) - 1$ and accounts for the variation of the optimal fiber length on the activation level, $f_p$ is the function that relates the active muscle stress to the nominal longitudinal strain rate $\dot{\varepsilon}_0^m$, $f_p$ describes the dependence of the passive stress on the nominal longitudinal strain $\varepsilon_0^m$ and the activation level, and $f_p$ is the function that relates the passive muscle stress to the nominal longitudinal strain rate $\dot{\varepsilon}_0^m$. The dependence of passive force production on the fiber activation level and on the velocity of passive stretching or/and releasing are accounted for in the muscle model.

Figure 1: Finite element meshes: (a) soft tissues, (b) skeleton, (c) skeleton and muscles with tendons.

In the present model, an isotropic linear hyperelastic model is used for the non-fiber part. The constitutive model for skeletal muscles is non-linear, rate-dependent, and anisotropic. The corresponding constitutive model for the tendons is the same as that of the muscles, with the nominal fiber stress having...
only a passive part, i.e., \( \sigma_0^n = \sigma_0^{\text{pass}} (\varepsilon_0^n) \). The constitutive model for muscles and tendons is implemented in the ABAQUS general purpose finite element program via “user subroutine” (UMAT).

The bony structures are modeled as homogeneous, isotropic and linearly elastic materials and the rest soft tissues (mainly fat), including the regions of articular cartilage, are modeled as hyperelastic materials. Ligaments and plantar fascia are included in the model as “tension-only” truss elements with linear elastic material properties.

The purpose of the current simulation is to produce plantar flexion of the ankle by activating the appropriate muscles (gastrocnemius and soleus) without applying any external forces on the foot and to estimate the stress and deformation state of the contracting muscles and Achilles tendon during foot movement. In the finite element calculations, plantar flexor muscles follow one cycle of activation and deactivation. Initially, the fiber directions of the lower-leg muscles and the Achilles tendon are assumed to be parallel to the long-axis of the tibia bone (vertical with regard to the plantar surface of the foot). A quasi-static analysis that accounts for the geometry changes (“large strain” analysis) is carried out using the “implicit” version of ABAQUS. The upper bound of the lower leg is kept fixed throughout the analysis.

**RESULTS AND DISCUSSION**

Figure 2 shows the initial state and the deformed configuration of foot and lower-leg as predicted by the finite element solution. The contraction of the plantar flexor muscles causes the motion of the ankle joint and the toes to point downward. An animation of the motion can be found at [http://www.youtube.com/watch?v=ItZyJiSdVms](http://www.youtube.com/watch?v=ItZyJiSdVms).

We examine stress and strain distributions throughout the Achilles tendon as well as its shape deformation resulted from ankle plantar flexion (Figure 3). The stress that develops on tendons as a result of voluntary muscle contraction is known to be ~30% of the maximum tensile strength [2], a mean value of which is reported to be 100 MPa [5]. The calculated value of maximum principal stress \( \sigma_1 = 29.9 \) MPa agrees well with the aforementioned results. Also, the calculated axial fiber strain of \( \varepsilon_0 = 2.99 \) is in agreement with previous reports [2]. The stresses and strains are distributed non-uniformly in Achilles tendon and the maximum principal stress and maximum fiber strain appear at the lateral portion of Achilles tendon (Figures 3b and 3c), which agrees with findings in the literature [4].

**CONCLUSIONS**

A simulation of foot movement, in the context of the finite element method, based entirely on the idea of considering muscles as “active” materials and not by applying external forces on tissues was developed. The model of human foot and lower-leg is used to estimate internal stresses and strains as well as shape changes of the deformed tissues during human movement.

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**REFERENCES**