DEVELOPMENT OF AN ANATOMICALLY BASED COMPUTATIONAL MODEL FOR INVESTIGATING FOOT BIOMECHANICS

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SUMMARY
The proposed study is to develop a detailed, anatomically based finite element (FE) model of the foot to investigate normal foot biomechanics. Geometries of foot structures, including the skin, 20 muscles, 26 bones, 17 tendons and blood vessels were identified from the Visible Human Project (VHP) 1 dataset. FE models were iteratively fitted to the digitized data points of individual foot structure using a least-squares algorithm. To generate subject-specific FE models, MRI images of the feet from healthy subjects were obtained. The generic model was then customized to the subject’s foot geometries using a host-mesh fitting technique. For mechanics simulation, muscle fibres were embedded into a foot volume mesh, with surrounding cavity assigned as plantar soft tissue. The double-limb balanced standing phase was simulated by applying a ground reaction force (GRF) from the ground. The foot-ground interface was modeled as contact surfaces with a friction coefficient of 0.6. Potential applications include the investigation of biomechanical aspects of diabetic foot ulceration and evaluation of the effectiveness of therapeutic footwear.

METHODS
Creating a generic anatomical model
To create the generic model of the foot, realistic geometric representations of the foot structures were obtained from two-dimensional (2D) transverse cross-sectional photographic images from the VHP dataset. Boundaries of various structures such as the skin, muscles, bones, tendons and blood vessels were manually digitized using the in-house software, Zinc2, at 1 mm interval between images (Figure 1a). This resulted in a cloud of three-dimensional (3D) data points. For each foot structure, an initial linear mesh was first created with nodes selected from the data points (Figure 1b). An iterative geometric fitting procedure was then performed with cubic Hermite interpolation functions (Figure 1c) [3]. The error function, defined as the sum of distances between the data points and their orthogonal projections onto the faces of the mesh, was minimized using a least-squares algorithm [Equation 1].

\[ e(\xi_d) = \sum_{d=1}^{D} \| u(\xi_d) - z_d \|^2 \]

Equation 1: Error function, \( e(\xi_d) \), whereby \( z_d \) denotes the spatial coordinates of a data point and \( u(\xi_d) \) is its projection.

INTRODUCTION
Biomechanical factors play an important role in the cause and treatment of foot-related disorders. For example, abnormal high plantar pressure undetected by the insensate diabetic foot is thought to be one of the major mechanical causes of foot ulceration [1]. Before applying orthosis or therapeutic footwear, it is important to understand the mechanical behavior of the normal foot. Since experimental studies are generally restricted to plantar pressure measurements and overall joint movements, the FE approach provides an alternative method to determine the internal stress distribution and relative movement between the internal structures. Modeling the biomechanics of the foot is a challenging task due to the complex geometries of the foot structures, highly nonlinear material properties and often ill-defined loading and boundary conditions. The proposed study is to develop a detailed, anatomically based FE model of the foot to investigate normal foot biomechanics. Potential clinical applications include perturbing model parameters to investigate the mechanics of the diabetic foot, evaluation of various treatments and surgical planning.

1 www.nlm.nih.gov

Figure 1: (a) Zinc interface showing digitization of foot structures from the VHP dataset, (b) linear mesh of a muscle and (c) cubic Hermite mesh of a muscle after iterative fitting to digitized data points.

2 www.cmiss.org/cmgui/zinc
Creating subject-specific anatomical models

A healthy male and female subject volunteered for this study. The right foot of each subject was scanned using a MRI scanner (3T, Siemens MAGNETOM Skyra) with the foot placed in the neutral position within a specialized foot boot. The optimal images of the foot obtained in terms of scanning time and quality were 448 x 448 pixels (resolution = 0.625mm) with slice thickness of 0.6 mm. Sagittal scans were taken of the entire foot and ankle region (Figure 2).

A similar digitization procedure was performed to identify the different structures of the foot. The digitized data points were then used to customize the existing generic model to the subject's foot geometries using host-mesh fitting techniques.

Figure 2: A high resolution MRI image of the female foot scanned in the saggital plane (left); a patient-specific FE model for simulating the foot-ground interface (right) with the ground (yellow) and contact points projections (blue arrows).

Mechanics Simulations

In order to employ the detailed anatomical model for solid mechanics simulations, it was first simplified to a single soft tissue continuum that encapsulates all internal structures. The soft tissue continuum is a volume mesh created from the digitized skin data. Fibers of the 20 muscles were then fitted to the soft tissue mesh using techniques developed in [3]. Double-limb balanced standing phase was simulated. A vertical ground reaction force of 350N was applied from the inferior surface of the ground. The prescribed loads were only allowed to move in the vertical direction only. The superior nodes of the soft tissue were fixed throughout the simulation. The foot-ground interface was modeled as contact surfaces with a friction coefficient of 0.6. Both soft tissue and ground were modeled as an isotropic, neo Hookean material with material stiffness of 5MPa and 1000MPa, respectively.

RESULTS AND DISCUSSION

The generic, anatomical model of the foot developed in this study consisted of the skin, 26 bones, 20 muscles, 17 tendons, and the dorsal and plantar blood vessels networks (Figure 3). The root mean squared (RMS) error obtained for fitting the skin FE mesh to the digitized points was 0.53mm, while the maximum RMS error for fitting all other structures was 0.49mm.

From the mechanics simulation, it was found that higher stresses appeared under the 1st and 2nd metatarsals and at the heel region (Figure 4). Isotropic material properties were assigned to the soft tissue in this study for simplicity. However, increased complexity can be added to the present model to improve its predictive capability, such as embedding the bones into the model and assigning anisotropic material properties to the soft tissue to take into account the muscle fiber fields.

Figure 3: Anatomically based finite element models showing the bones (white), muscles (dark red) and tendons (purple) based upon the VHP dataset.

Figure 4: The von Mises stress distribution for a balanced standing simulation, assuming isotropic material properties for the soft tissue.

CONCLUSIONS

In this study, detailed anatomically based computational models were developed, based upon a generic dataset and subject-specific MRI data. The developed FE models form a solid framework for further investigation of biomechanical aspects of diabetic foot ulceration and evaluation of the effectiveness of therapeutic footwear.

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REFERENCES