A NUMERICAL-EXPERIMENTAL APPROACH FOR CHARACTERIZING SUBJECT SPECIFIC HYPERELASTIC PROPERTIES OF THE HEEL PAD

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INTRODUCTION

Many investigators have examined the mechanical properties of the heel pad since such data are essential for realistic numerical models of the foot and for accurate comparison of normal and diseased tissue behavior. Traditionally, structural properties of the heel have been determined from *in vivo* impact tests [1,2] during which load-deformation characteristics were measured and stress-deformation curves were derived. However, the calculation of true stress in these studies was not possible due to unknown boundary conditions. Lately, the material properties of the heel have been determined from *in vitro* testing of tissue samples with uniform geometry [3,4]. Such an approach is not possible for living subjects, therefore limiting the applicability of cadaver experimentation techniques to the determination of individualized material models *in vivo*.

Numerical-experimental approaches have been successfully used to determine hyperelastic material properties of tendons and muscles under compression [5,6]. The methodology uses the load-deformation data of the tissue of interest as inputs to a finite element model that mimics the geometry and boundary conditions of the experiment. Combined with imaging techniques, this approach allows *in vivo* determination of material properties.

The aims of this study were to develop a combined methodology of finite element modeling and ultrasound imaging for *in vivo* identification of heel pad properties and to determine subject-specific hyperelastic material parameters for calcaneal soft tissue of healthy individuals.

METHODS

The right heels of 8 healthy subjects (3 males, 5 females, age: 19-75 yrs) were indented by using a cylindrical indenter (dia. 25.4 mm) to a depth of 37%-60% of the initial thickness of the heel pad. The indentation was carried out by a linear actuator at a rate of 12.7 mm/sec. An ultrasound machine (SSD-500, ALOKA Co., Ltd.) recorded the displacement of the indenter tip with respect to the calcaneus and a serially connected load cell (Kistler Instrument, Corp.) measured the indentation force.

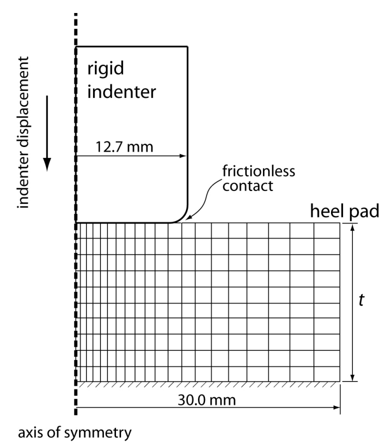

Figure 1. Axisymmetric model of the heel.

A displacement driven axisymmetric finite element model of the heel was developed to determine material properties of the soft tissue (Figure 1). The initial thickness of the heel pad (t) was adjusted for each subject (range: 14.1–23.4 mm) whereas the diameter was set at 60 mm. The indenter was assumed to be rigid and the soft tissue was modeled as an incompressible hyperelastic material with the strain-energy function (U) represented by a first order Ogden form as

$$U = \frac{2\mu}{\alpha}\left(\lambda_1^{\alpha} + \lambda_2^{\alpha} + \lambda_3^{\alpha} - 3\right) \qquad (1)$$

where $\lambda_{1\text{-}3}$ represent the deviatoric principal stretches and μ and α are material parameters [7]. μ is equal to the initial shear modulus and therefore linearly related to initial elastic modulus [7]. α, is a measure of the increase in material stiffness as a result of loading. Contact between the indenter and the soft tissue was assumed to be frictionless. The material parameters shown in equation 1 were derived by an iterative process involving a nonlinear least squares formulation [8].

During each step, a finite element model was solved using the current material parameters and the force–displacement characteristics of the heel tissue were calculated. The material parameters were then changed to minimize the error between the experimental load-displacement curve and the one predicted by the model. ABAQUS (Hibbitt, Karlsson & Sorensen, Inc.) was used as the finite element solver together with a freeware implementation of the Levenberg-Marquardt algorithm [9].

RESULTS

All heel pads exhibited nonlinear deformation during indentation (Figure 2). The iterative process described above using an Ogden material model was capable of capturing the hyperelasticity of the heel with the root-mean-square (RMS) error between finite element model predictions and experiment results being smaller than 4 N (5% maximal loading) for all subjects (Table 1).

DISCUSSION

An *in vivo* methodology was developed to measure material properties of the heel pad. The technique integrates the influence of boundary conditions using finite element analysis to improve upon previous *in vivo* measurements [1,2].

The measurements made in this study considered only constant rate loading of the heel. Therefore, hysteresis and viscoelastic effects were neglected. The parameters calculated using this approach should be viewed with care since the expected loading conditions in prospective simulations should approximate those of *in vivo* measurements. Nevertheless, the methodology can be updated easily to include different material models as well as viscoelastic effects, when such data are available. Among other limitations of the present approach are that the geometry of the heel was simplified, all tissues between the skin and bone were lumped, and the contact between indenter and the heel was assumed to be frictionless. A sensitivity analysis revealed that material properties were not overly influenced by the selected diameter of the heel. The curvature of the heel pad, on the other hand, was found to be influential, especially on the values of μ; increasing the convexity of the plantar surface elevated μ. The inclusion of a detailed geometry may improve the accuracy of our results with the disadvantage of increased pre-processing effort and complexity of the finite element model.

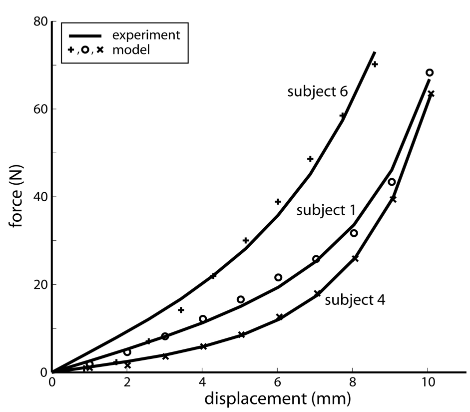

Figure 2. Experimental and finite element model predicted indentation forces for three subjects.

Comparison of the present results with those of other studies is difficult due to variations in measurement approaches and because of the diversity of material definitions [4]. A direct comparison can be made between heel pad properties obtained in this study and compression characteristics of the supraspinatus tendon properties reported by Zobitz et al. [6] who also used a first order Ogden formulation. The mean initial shear modulus (μ) of the heel pad in the current study was found to be similar to that of supraspinatus tendon, whereas α was much smaller. The high variability of μ in the current study can be attributed to the diverse age and mixed gender of our subject sample, and also to the lack of fit at low loading levels in some subjects (see subject 6, Figure 2).

Table 1. Material properties and force prediction errors.

Subject	μ (kPa)	α	RMS (N)	RMS (% Max Force)
1	13.00	5.16	1.5	2.2
2	6.86	7.52	1.6	2.7
3	12.31	7.63	2.0	2.7
4	7.97	10.79	0.4	0.6
5	12.91	6.64	1.4	2.0
6	31.60	8.46	3.3	4.6
7	9.40	6.42	1.2	2.4
8	20.40	5.50	1.6	2.2
Mean (SD)	14.3 (8.1)	7.3 (1.8)	1.6 (0.8)	2.4 (1.1)

The numerical-experimental approach described in this study provides the basis for development of computational foot models with personalized soft tissue representations. This methodology can also be used for detailed comparison of normal and diseased tissue properties.

ACKNOWLEDGMENTS

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