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Hyperelastic compressive mechanical properties of the subcalcaneal soft tissue: An inverse finite element analysis

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ABSTRACT

Finite element (FE) foot models can provide insight into soft tissue internal stresses and allow researchers to effectively conduct parametric analyses. Accurate plantar soft tissue material properties are essential for the development of FE foot models for clinical interventions. The aim of this study was to identify the first-order and second-order Ogden hyperelastic material properties of the subcalcaneal fat using an inverse FE analysis. The cylindrical soft tissue FE model was developed based on *a* priori in vitro dynamic compression experiment. The model simulated a 1 Hz triangle wave displacement to apply a compressive strain up to 48%. The hyperelastic properties were identified by systematically varying the material parameters to minimize the difference between the model predicted force and the target experimental data. Optimal material properties were obtained ($\mu_1 = 0.0235$ kPa and $\alpha_1 = 12.07$ for the first-order Ogden model and $\mu_1 = -4.629 \times 10^{-6}$ kPa, $\alpha_1 = -16.829$; $\mu_2 = -1.613$ kPa and $\alpha_2 = -1.043$ for the second-order Ogden model). The second-order Ogden model was superior in capturing the highly nonlinear force–deformation response when compared to the first-order model (root mean square error (RMSE) 0.169 N vs. 0.570 N). The material sensitivity analysis indicated that the predicted force was strongly affected by the Poisson's ratio (12-fold increase in RMSE when reducing Poisson's ratio by 10% from the baseline) and the coefficient α_1 (3.2-fold and 32-fold increase in RMSE for both first-order and second-order Ogden models when increasing α_1 by 10% from the optimal value).

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1. Introduction

Finite element (FE) analysis of the foot has been used to predict plantar pressure (Budhabhatti et al., 2009; Chen et al., 2012; Cheung and Zhang, 2008; Isvilanonda et al., 2012), visualize internal soft tissue stress distributions (Chen et al., 2010; Gefen, 2003), perform parametric studies (Budhabhatti et al., 2009; Chen et al., 2012; Cheung et al., 2004, 2006) and design pressure relief insoles and orthoses (Chen et al., 2003; Cheung and Zhang, 2005; Erdemir et al., 2005). The accuracy of the model predictions depends greatly on physiologic material properties and anatomical data.

Hyperelastic material models, e.g., generalized polynomials (Cheung et al., 2006; Lemmon et al., 1997), Mooney–Rivlin (Miller-Young et al., 2002) and Ogden (Chen et al., 2012; Chokhandre et al., 2012; Erdemir et al., 2006; Isvilanonda et al., 2012; Spears et al., 2007), are widely used to approximate plantar soft tissue materials. The force-deformation response of the plantar soft tissues can be obtained directly from in vitro uni-axial compression tests (Ledoux and Blevins, 2007; Miller-Young et al., 2002; Pai and Ledoux, 2010). From this type of testing, hyperelastic coefficients can be identified using non-linear regression of analytical models (Miller-Young et al., 2002; Natali et al., 2010). These analytical models are often developed for simple specimen geometries and loading scenarios, and therefore require input data from specific material testing methods (e.g., frictionless uni-axial tension/compression). Alternatively, researchers have used heel-pad indentation to obtain structural force-deformation responses from intact feet (Erdemir et al., 2006; Lemmon et al., 1997). The complicated anatomy and boundary conditions associated with these tests require identification of hyperelastic properties through an inverse FE analysis (Chokhandre et al., 2012; Erdemir et al., 2006; Gu et al., 2010; Halloran and Erdemir, 2011).

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Previously, our group investigated the compressive mechanical response of the in vitro plantar fat tissue samples (Pai and Ledoux, 2010). Due to the no-slip boundary condition and large deformation imposed in these experiments, the reported stress–strain data are unsuitable for direct application to analytical model regression, therefore we explored the use of inverse FE analysis to quantifying the compressive Ogden hyperelastic material properties of the subcalcaneal fat.

2. Methods

2.1. Material testing

The detailed preparation and testing protocol of the plantar soft tissue samples are described elsewhere (Pai and Ledoux, 2010). Briefly, each specimen (cylinder, 19.05 mm diameter) was isolated from the plantar skin and bone. The specimen was constrained between two loading platens using 220 grit sand paper and maintained under approximately physiologic temperature (35 °C) and humidity (near 100%) inside an environmental chamber. Displacement control triangle waves were used to compress the tissue to an average of 48% strain. Of the five tested frequencies (1, 2, 3, 5 and 10 Hz), the 1 Hz results from the subcalcaneal region of the non-diabetic samples ($n=9$) were used for the work presented here. Three consecutive loading and unloading force-deformation cycles were averaged to yield an estimate of purely elastic force-deformation which was used as the inverse FE analysis target data (Fig. 1).

2.2. Finite element model

An FE model of the uni-axial compression test (Fig. 2) was created using automatic hexahedral shape mesher in LS-Prepost (Livermore Software, Livermore, CA) to simulate the plantar soft tissue specimens. Linear hexahedral elements were used due to their superior convergence rate, accuracy and computation time compare to linear and quadratic tetrahedral elements (Tadepalli et al., 2011). The structure was halved due to symmetry. The model size (from the averaged specimen diameter and thickness of 19.05 mm and 9.54 mm, respectively) was assigned. Mesh refinement analysis was performed on the geometry to obtain convergence of the force-deformation output, i.e., an 8-fold increase in the total number of elements lead to only a 3.0% change in the force-deformation response (data not shown). The final soft tissue model consisted of 2160 linear hexahedron elements (constant stress solid element) (Hallquist, 2006).

The soft tissue was modeled as a homogeneous, isotropic, nearly-incompressible Ogden hyperelastic material (material type 77 in LS-DYNA) (Hallquist, 2006). The strain energy function is represented by Eqs. (1)–(5):

$$W(\lambda_1,\lambda_2,\lambda_3)=\sum_{m=1}^{n}\frac{\mu_m}{\alpha_m}\left(\tilde{\lambda}_1^{\alpha_m}+\tilde{\lambda}_2^{\alpha_m}+\tilde{\lambda}_3^{\alpha_m}-3\right)+\frac{1}{2}K(J-1)^2 \quad (1)$$

$$K=\frac{(1+\nu)}{3(1-2\nu)}\sum_{m=1}^{n}\mu_m\alpha_m \quad (2)$$

$$J=\lambda_1\lambda_2\lambda_3 \quad (3)$$

$$\tilde{\lambda}_i=J^{-1/3}\lambda_i \quad (4)$$

$$\mu_m\alpha_m>0 \text{ (for each } m=1,\ldots,n \text{ and no sum over } m) \quad (5)$$

where $\lambda_1,\lambda_2,\lambda_3$ are the three principal stretches, the Jacobian, J, denotes the relative volume change, and $\tilde{\lambda}_i$ are the deviatoric principal stretches. Density ($\rho=1.00$ g/cm^3), Poisson's ratio (ν) and hyperelastic coefficients (μ_m and α_m) are the material parameters. Bulk modulus, K, can be derived from Eq. (2). Coefficients μ_m and α_m are the design variables that will be optimized. The product of μ_m and α_m is constrained according to Eq. (5) to ensure physiologic response and material stability (Ogden et al., 2004), i.e., this product must be positive. The first-order ($n=1$) and second-order ($n=2$) Ogden hyperelastic models were analyzed. Poisson's ratio ($\nu=0.4999$) was estimated from the longitudinal wave velocity in subcutaneous fat ($C_L=1476$ m/s) (Duck, 1990), mass density (ρ) and elastic modulus ($E=674$ kPa) (Pai and Ledoux, 2010) by solving Eq. (6) (Duck, 1990).

$$C_L=\left[\frac{E(1-\nu)}{\rho(1-2\nu)(1+\nu)}\right]^{\frac{1}{2}} \quad (6)$$

The sand paper-soft tissue interface was reproduced in the FE model by rigidly constraining the superior and inferior soft tissue surfaces to the rigid loading platens, i.e., a no-slip boundary condition. Nodes along the plane of symmetry were constrained to prevent movement in the normal direction. Potential model instabilities include non-physiologic deformation and high hourglass energy larger than 10% of the internal energy (Miller, 2011). To avoid such instabilities, we implemented the following LS-DYNA parameters: global damping (damping constant, 35; i.e., 10% of the critical damping determined from the lowest natural frequency), hourglass controlling (IHQ=4; QM=0.002–0.025) and automatic time stepping. Compression was applied by translating the inferior platen toward the fixed superior platen using a linear ramp from 0 to 4.60 mm (48% strain) in 0.5 s (comparable to a 1 Hz triangle wave). The reaction force on the upper platen from the half-symmetric model was doubled to represent the full geometry. The model deformation and force-deformation response were validated with analytical solutions (i.e., mathematical equations that describe force and deformation of a fixed-end cylinder under axial compression) up to 30% strain (Klingbeil and Shield, 1966; Miller, 2005).

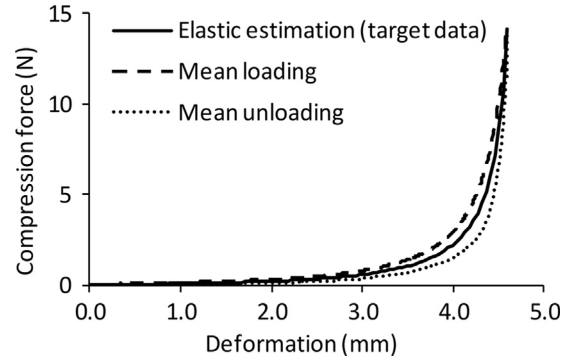

Fig. 1. The mean elastic compressive force-deformation response (solid line) approximated by averaging the loading (dash line) and unloading (dotted line) cycles of the 1 Hz cyclic test of the plantar soft tissue (Pai and Ledoux, 2010). The peak force and displacement were 14.12 N and 4.60 mm, respectively.

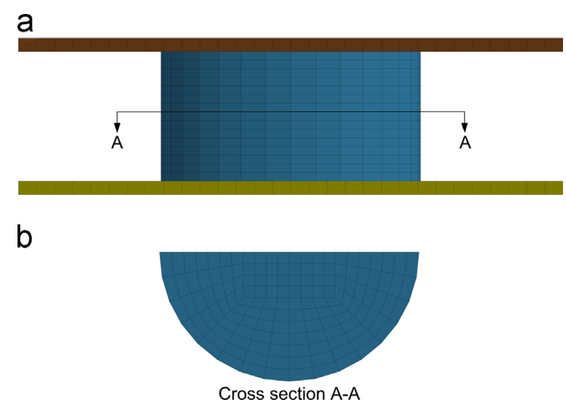

Fig. 2. Finite element model of the uni-axial compression test of the plantar soft tissue cylinder (halved due to symmetry). Concerning boundary conditions: no slip (fixed) was permit at the soft tissue-platen interfaces to emulate the experimental set up.

2.3. Material parameter identification

The Ogden hyperelastic material properties (μ_m and α_m) were identified by minimizing the difference between the model predicted force and the target experimental data using LS-OPT (Livermore Software, Livermore, CA). A sequential response surface method (SRSM) optimization technique (Stander et al., 2010) was used. The design space ($0<|\mu_m|<0.05$ MPa, $0.1<|\alpha_m|<70$) was estimated based on prior literature (Erdemir et al., 2006; Gu et al., 2010; Spears et al., 2007) and pilot simulations. The inverse analysis began by randomly selecting initial guesses for the material parameters (Tables 1a and b) in MATLAB (MathWorks, Natick, MA). For the first-order model, five random positive μ_m and α_m initial values were chosen, and then the negatives of the same values were calculated, resulting in 10 initial guesses. Similarly, for the second-order model, five random positive μ_m and α_m initial values were chosen, and then the corresponding double positive and double negative combinations were used, resulting in 20 initial guesses. For each initial guess, a D-optimal point selection algorithm (Stander et al., 2010) in LS-OPT chose a subset of μ_m and α_m from the region of interest; the number of experiments (i.e., the number of unique μ_m and α_m pairs) in each subset ranged from 8 to 42. In the first iteration, the region of interest included the entire design space. Next, a series of FE model inputs were generated and solved in LS-DYNA (version 971d R5.1.1 explicit analysis). The computations were performed on two Dell PowerEdge R610 rack servers (each has two–six core 3.46 GHz Intel® Xeon® processors). The optimization objective function was defined as a normalized mean square error