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Estimating the material properties of heel pad sub-layers using inverse Finite Element Analysis

Nafiseh Ahanchian[a,*], Christopher J. Nester[a], David Howard[b], Lei Ren[c], Daniel Parker[a]

[a] School of Health Sciences, University of Salford, Salford, UK
[b] School of Computing, Science & Engineering, University of Salford, Salford, UK
[c] School of Mechanical, Aerospace and Civil Engineering, University of Manchester, UK

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ABSTRACT

Detailed information about the biomechanical behaviour of plantar heel pad tissue contributes to our understanding of load transfer when the foot impacts the ground. The objective of this work was to obtain the hyperelastic and viscoelastic material properties of heel pad sub-layers (skin, micro-chamber and macro-chamber layers) *in-vivo*.

An anatomically detailed 3D Finite Element model of the human heel was used to derive the sub-layer material properties. A combined ultrasound imaging and motorised platform system was used to compress heel pad and to create input data for the Finite Element model. The force–strain responses of the heel pad and its sub-layers under slow compression (5 mm/s) and rapid loading-hold-unloading cycles (225 mm/s), were measured and hyperelastic and viscoelastic properties of the three heel pad sub-layers were estimated by the model.

The loaded (under ~315 N) thickness of the heel pad was measured from MR images and used for hyperelastic model validation. The capability of the model to predict peak plantar pressure was used for further validation. Experimental responses of the heel pad under different dynamic loading scenarios (loading-hold-unloading cycles at 141 mm/s and sinusoidal loading with maximum velocity of 300 mm/s) were used to validate the viscoelastic model.

Good agreement was achieved between the predicted and experimental results for both hyperelastic (<6.4% unloaded thickness, 4.4% maximum peak plantar pressure) and viscoelastic (Root Mean Square errors for loading and unloading periods <14.7%, 5.8% maximum force) simulations. This paper provides the first definition of material properties for heel pad sub-layers by using *in-vivo* experimental force–strain data and an anatomically detailed 3D Finite Element model of the heel.

1. Introduction

The behaviour of the plantar heel pad has been the topic of considerable research because it forms a critical interface with the supporting surface. It is affected by aging and disease and is the site of pain [1–3]. Study of heel pad behaviour has been achieved through experimental [4–6] and numerical methods, particularly Finite Element Analysis (FEA) [7–9]. The latter provides data such as the distribution of internal tissue stress that cannot be experimentally measured. However, for FEA models to prove effective they should be based on geometric and material properties that ensure the model behaviour is sufficiently close to *in-vivo* heel pad behaviour, as seen during human gait.

In most Finite Element (FE) models, hyperelastic rather than viscoelastic material models were used to simulate nonlinear behaviour of the heel pad [7–10]. Results from these studies were limited to static or fixed loading rates due to the absence of a dynamic *in-vivo* system that allows compression of plantar tissues at various high speeds, whilst also providing the data required for estimation of viscoelastic parameters and validation.

In addition, the heel pad is typically modelled as a homogeneous single-layer material rather than an *in-vivo* tri-layer biological structure (macro, micro and skin layers) [7,10,11]. In a few cases, the heel pad was modelled as a dual-layer composite structure (fat and skin), but this ignores the different behaviours and interactions between micro and macro layers [8,9,12]. This may compromise the ability of FEA to predict internal stresses.

A further issue with some of the models reported thus far is that experimental data were obtained *ex-vivo* [12–15]. Tissue dissection disrupts the normal *in-vivo* tissue constraints and the

* Corresponding author.
E-mail addresses: nahanchian@yahoo.co.uk, N.Ahanchian@salford.ac.uk (N. Ahanchian).

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effects of time and loss of vascular supply are not fully understood [16]. Clearly, *in-vivo* methods at appropriate loading rates are preferred over *ex-vivo* approaches.

In summary, most of heel pad models are limited by excluding viscoelastic effects and/or using less than three layers. Moreover, some approaches to validation may not test models with sufficient rigour. Hence, the objective of this work was to estimate hyperelastic and viscoelastic material properties of "macro-chamber", "micro-chamber" and "skin" layers using inverse FEA and *in-vivo* experimental data.

2. Methods

2.1. Finite Element model

An anatomically detailed model of the right heel of a healthy female volunteer (34 years old, height 164 cm, weight 63 kg, shoe size 5 UK) was constructed based on unloaded MRI images. MRI data were T1 weighted with a flip angle of 25, taken in coronal view using 3D fast field echo (Philips 1.5T Acheiva), with pixel size = 0.29mm × 0.29 mm (2.4% resolution), and slice intervals = 1.25 mm. The images were segmented to identify the plantar fascia, muscle tissue, macro-chamber, micro-chamber and skin layers and create corresponding 3D surface geometries using ScanIP v3.1 (Simpleware Ltd, Exeter, UK). Different segmentation algorithms including thresholding, confidence connected region-growing, floodfill and paint were used for identifying the corresponding tissues. 3D surface geometries were imported into SolidWorks 2010 (Dassault Systemes, USA) to generate 3D solid geometries and the complete assembly. Since MRI slices were out of the plane of boundaries between soft tissue layers, the effect on structural modelling will be minimal. Also, the 0.29 mm between slices is a small percentage of the anterior/posterior length of the structured modelled. A full description of the development of the heel region structures can be found elsewhere [17].

To reduce the computation time only a portion of the foot was modelled. Planes at 92.5 mm from the back and 45 mm from the bottom of the heel were chosen to be flat face boundaries of the model. The solid model was meshed with 11,504 hexahedral elements (type C3D8R) using ABAQUS v6.10 (Dessault Systemes, USA). The number of elements was obtained by performing a mesh convergence study. The selected mesh density was based on the change in the peak force for a subsequent doubling of mesh density being less than 3%. The meshed model was exported to Ls-Dyna v2.2 (Livermore Software Technology Corporation, Livermore, USA) for inverse FEA. Effects of stiff tissues (foot bones and Achilles tendon) on the biomechanical behaviour of the heel pad were simulated by applying zero-displacement constraints to all nodes forming the soft tissue-stiff tissue interface. The Achilles was modelled as stiff since under tension it will be far stiffer than the fat pad and far from it too, acting as a rigid attachment to the heel bone which is thereafter attached to the heel pad. All nodes at the superior and anterior boundaries (flat faces) of the model were fully constrained. The model was tilted by 17° to replicate the position of the foot during subsequent experiments performed with a Soft Tissue Response Imaging Device (STRIDE) (Fig. 1) [18]. In Ls-Dyna, the flat indentation plate of the STRIDE was modelled as a rigid structure (Fig. 1). Tied contact was defined between the parts of the heel model and frictionless surface-to-surface contact was defined between the indentation plate and heel skin.

The macro-chamber, micro-chamber and skin layers were modelled as nonlinear viscoelastic materials (Fig. 1). The first-order Ogden model was used to represent the hyperelastic behaviour of heel pad tissues as done previously [7–9]. The corresponding material properties appear in the strain energy function as follows

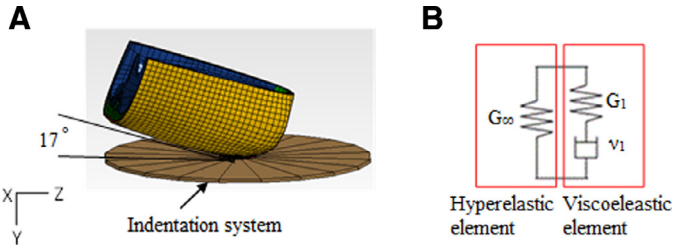

Fig. 1. (A) The complete meshed model of the heel region; (B) the behaviour of the tissues making up the three layers was modelled using a combination of an Ogden hyperelastic model and a Maxwell element.

$$W = \frac{\mu}{\alpha}(\lambda_1^{\alpha} + \lambda_2^{\alpha} + \lambda_3^{\alpha} - 3) \quad (1)$$

where λ_{1-3} are the principal stretches in the x, y and z directions respectively, μ is the shear modulus, and α is the deviatoric exponent (μ and α being the hyperelastic material parameters). Viscoelastic tissue behaviour was modelled using one generalised Maxwell element for the viscoelastic overstress in the Ogden model. The Maxwell viscoelastic element consists of a linear spring with stiffness G_1 and a linear damper with viscosity v_1 in series. The relaxation time (a measure of the time taken for the stress to relax) for the Maxwell unit is $\tau_1 = v_1 / G_1$. Its inverse is the decay constant $\beta_1 = 1/\tau_1$. The stiffness G_1 (the shear relaxation modulus) and decay constant β_1 are the viscoelastic material parameters of the model in Ls-Dyna. The corresponding material properties appear in the relaxation function, $G(t)$, written as a first-order Prony series, representing the combined hyperelastic and viscoelastic model as follows

$$G(t) = G_{\infty} + G_1 e^{-\beta_1 t} \quad (2)$$

where G_{∞} is the long term shear modulus (Fig. 1).

The focus of the reported work is to identify the properties of and model the heel pad. However, the surrounding tissues that constrain the heel pad must also be modelled adequately enough to provide realistic boundary conditions. Therefore, to simplify the FE model, the plantar fascia and muscle tissues were modelled as linear elastic materials. However, the literature contains limited reports concerning the material properties of muscle tissues and plantar fascia and, in most other FE studies, the foot muscles have been merged with the heel pad tissue and assigned the same material properties [19–21]. Moreover, the plantar fascia has previously been modelled with tension-only truss elements with Young's modulus determined from tensile tests [19,20,22]. Since there is poor agreement between studies, a series of parametric studies was conducted to assess the sensitivity of the FEA results to the material properties used for the plantar fascia and muscle tissue. Different material properties, derived from published data [21–25], were assigned to the plantar fascia and muscle tissue and this revealed only a small effect on the force–strain behaviour of the heel pad (Root Mean Square (RMS) error <1.5% and <0.67% max force for the plantar fascia and muscle tissue respectively). The initial material properties derived from published literature were therefore used to start the FEA (Table 1).

2.2. Experimental acquisition of force and tissue displacement data

The aim of this stage was to perform a series of slow and rapid compression tests on the same heel used to generate the geometric model and obtain the force–strain responses of the heel pad and its sub-layers. Ethical approval was granted by the University of Salford ethical committee.

STRIDE applies controlled vertical compression cycles of various speeds and load profiles to the heel pad *in-vivo*. It simultaneously

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