Journal of Biomechanics xxx (2018) xxx–xxx

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Short communication

Material characterization of cardiovascular biomaterials using an inverse finite-element method and an explicit solver

Miriam Nightingale, Michel R. Labrosse *

Department of Mechanical Engineering, University of Ottawa, 161 Louis Pasteur, Ottawa, Ontario K1N 6N5, Canada

ARTICLE INFO

Article history:
Accepted 14 July 2018
Available online xxxx

Keywords:
Material characterization
Cardiovascular biomaterials
Inverse method
Explicit finite element solver

ABSTRACT

The ability to accurately model soft tissue behavior, such as that of heart valve tissue, is essential for developing reliable numerical simulations and determining patient-specific care options. Although several material models can predict soft tissue behavior, complications may arise when these models are implemented into finite element (FE) programs, due to the addition of an arbitrary penalty parameter for numerically enforcing material incompressibility. Herein, an inverse methodology was developed in MATLAB to use previously published stress-strain data from experimental planar equibiaxial testing of five biomaterials used in heart valve cusp replacements, in conjunction with commercial explicit FE solver LS-DYNA, to optimize the material parameters and the penalty parameter for an anisotropic hyperelastic strain energy function. A two-parameter optimization involving the scaling constant of the strain energy function and the penalty parameter proved sufficient to produce acceptable material responses when compared with experimental behaviors under the same testing conditions, as long as analytically derived material constants were available for the other non-optimized parameters and the actual tissue thickness was not much less than 1 mm. Variations in the penalty parameter had a direct effect on the accuracy of the simulated responses, with a practical range determined to be $5 \times 10^8 - 9 \times 10^8$ times the scaling constant of the strain energy function.

1. Introduction

For elective surgical procedures, such as heart valve repair, reliable computational simulation could allow for trial-and-error to happen in silico before surgery, with appropriate courses of action already whittled down from the evaluation of several surgical options. Such ability might lead to a decrease in surgery time, higher success rates and less invasive operations.

When developing finite element (FE) simulations, four main areas must be considered: geometry, boundary conditions, loading, and material behavior. This work only focuses on the latter for cardiovascular soft tissues. In such tissues, the chief load bearing components, namely collagen fibers, have a preferred direction usually along the main load path (Holzapfel et al., 2000), making the material anisotropic. In addition, large elastic deformations are allowed upon loading by the elastin components of the extra-cellular matrix until the initially crimped collagen fibers become increasingly taut and stiffen up the material response. These tissues have successfully been modeled as hyperelastic anisotropic, whereby the 2nd Piola-Kirchhoff stress tensor is obtained from the partial derivative of the strain energy function of the material with respect to the Green strain tensor (Holzapfel et al., 2000; Humphrey, 2013).

Several strain energy models have been developed, and have lent themselves to the identification of material-specific parameters by inverse methods that rely on analytical models (Holzapfel et al., 2000; Kim et al., 2012; Azadani et al., 2013; Labrosse et al., 2013), or FE simulations (Lee et al., 2014; Aggarwal and Sacks, 2016; Oomen et al., 2016; Liu et al., 2017). In the iterative process of inverse methods, a forward prediction of the material response is made based on the material parameters obtained from the previous iteration (for the first iteration, an educated guess for the material parameters might suffice). The error between the predicted and experimentally stress-strain data may then be used to drive the evolution of the material parameters such that they eventually minimize the error, within a certain tolerance.

Given that a set of material constants describing a tissue may not be unique (Liu et al., 2017), it is doubtful that material constants obtained from different inverse methods (e.g. analytical or FE), different FE solvers (e.g. static or dynamic; implicit or explicit), different FE formulations, and different methods for enforcing the quasi-incompressibility that characterizes cardiovascular soft

* Corresponding author at: University of Ottawa, Department of Mechanical Engineering, 161 Louis Pasteur, Colonel By Hall A213, Ottawa, ON K1N 6N5, Canada.
E-mail address: labrosse@eng.uottawa.ca (M.R. Labrosse).

https://doi.org/10.1016/j.jbiomech.2018.07.026
0021-9290/

tissues, might be used interchangeably without compromising the accuracy of the simulations. However, for consistency, it seems logical to use the same tools throughout, from material constants identification to ulterior simulations.

Because of the numerical difficulties involved in the structural simulation of heart valves (namely, anisotropic hyperelasticity, leaflet curvature changes, dynamics and contacts), implicit solvers, where multiple costly iterations are required until convergence for each (possibly large) time step are arguably not as effective as explicit solvers, where the solution directly marches on from one (possibly very small, to satisfy Courant's stability criterion) time step to the next (Labrosse et al., 2015). Hence the focus placed herein on an explicit solver, commercial FE software LS-Dyna R8.1.0 (LSTC, Livermore, CA, USA). LS-DYNA, as do many other FE codes, uses a penalty term P to enforce incompressibility by constraining the determinant of the right Cauchy-Green tensor of the transformation (or similar metric) to be unity. It is common practice to assume a large value for P, usually as 10^3 –10^4 times the scaling constant of the strain energy function (Bonet and Wood, 1997). Importantly, high P values significantly decrease the time step automatically determined by LS-DYNA for stability of the solution, possibly leading to impractical run times.

We developed an optimization approach to identify the material constants for five representative cardiovascular biomaterials using an anisotropic hyperelastic strain energy model implemented within LS-DYNA. To improve the robustness and performance of the optimization, we treated P as an additional material constant and determined a range of practical values for the material constants to be used for simulation of these biomaterials in LS-DYNA.

2. Methods

2.1. Experimental data

The experimental stress-strain data used herein (Fig. 1) were obtained from five heart valve cusp replacement and native biomaterials that were recently tested in our lab using biaxial tensile testing: (1) glutaraldehyde fixed porcine pericardium (GPP), (2) fresh autologous porcine pericardium (APP), (3) CardioCel patch (CC), (4) fresh porcine aortic valve leaflets (PC), and (5) St Jude Medical pericardial patch (SJM). The experimental details are available in (Labrosse et al., 2016). Briefly, nine samples of each biomaterial were tested; 10 preloading cycles were used before stress-strain curves were collected. The materials' strains were computed from digital image correlation (DIC) over the central region of the samples.

2.2. Forward calculation

An input file was created for LS-DYNA to mesh a $5 \times 5 \times 1$ mm^3 solid volume with $10 \times 10 \times 2$ 8-noded hexahedral elements. The FE model represented one-quarter of the physical sample. The nodes on the symmetry plane perpendicular to the fiber (respectively cross-fiber) direction were only fixed in the fiber (respectively cross-fiber) direction. Only the bottom nodes of the model were fixed in the thickness direction, letting the model experience isochoric deformations. Displacements corresponding to the experimental Green strains measured from DIC were imposed on the appropriate sides of the model as ramps over a period of 0.01 s.

An adapted 3D Fung model, referred to as Guccione et al.'s model, was chosen as strain energy function for its ability to represent cardiovascular materials (Guccione et al., 1991; Labrosse et al., 2013) and availability in LS-DYNA (*MAT_HEART_TISSUE +). The strain energy function W for Guccione et al.'s material model is expressed in terms of the Green strain components E_{ij} such that

$$W = \frac{c}{2}(e^Q - 1) + \frac{P}{2}(I_3 - 1), \tag{1}$$

where

$$Q = b_1 E_{11}^2 + b_2\left(E_{22}^2 + E_{33}^2 + 2E_{23}^2\right) + 2b_3\left(E_{12}^2 + E_{13}^2\right) \tag{2}$$

and I_3 is the third invariant of the right Cauchy-Green tensor. The Green strain components are modified to eliminate any effects of volumetric work (Guccione et al., 1991). Indices 1, 2 and 3 refer to the fiber, cross-fiber and thickness directions of the sample, respectively. As can be seen, if P is included as a material constant, five material parameters need to be defined: c, b_1, b_2, b_3 and P. Equivalently, we looked to identify c, b_1, b_2, b_3 and K, where $P = cK$. The computations were run on a workstation with two Intel Xeon E5640 2.67 GHz 4-core processors and 6 GB of RAM.

2.3. Error minimization

The Green strains and Cauchy stresses (t_{ij}) outputted by LS-DYNA were collected and arranged with the experimental data

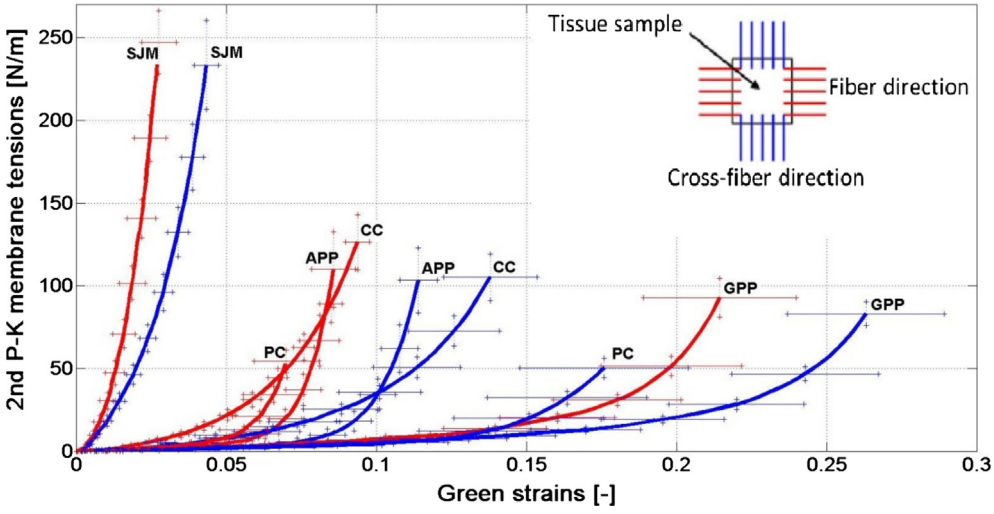

Fig. 1. Experimental material responses for five biomaterials under equibiaxial tension planar testing: (1) glutaraldehyde fixed porcine pericardium (GPP), (2) fresh autologous porcine pericardium (APP), (3) CardioCel patch (CC), (4) fresh porcine AV leaflets (PC), and (5) St Jude Medical pericardial patch (SJM) (Labrosse et al., 2016).

Please cite this article in press as: Nightingale, M., Labrosse, M.R. Material characterization of cardiovascular biomaterials using an inverse finite-element method and an explicit solver. J. Biomech. (2018), https://doi.org/10.1016/j.jbiomech.2018.07.026