International Journal of Solids and Structures 000 (2018) 1–10

Contents lists available at ScienceDirect

International Journal of Solids and Structures

journal homepage: www.elsevier.com/locate/ijsolstr

A micro-structure based constitutive model for anisotropic stress–strain behaviors of artery tissues

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ARTICLE INFO

Article history:
Received 20 December 2017
Revised 12 January 2018
Available online xxx

Keywords:
Constitutive model
Artery tissues
Anisotropic
Soft materials

ABSTRACT

Many kinds of soft composite materials such as biological tissues exhibit strong anisotropic features. In particular, hyperelastic stress–strain behaviors of artery tissues have been long studied, since they play an important role in the cardio-vasculature. While many constitutive models were proposed, they usually failed to capture the anisotropic stress–strain behaviors under different ratios for biaxial loading. In this paper an anisotropic constitutive model for soft composite materials is presented, driven by the underlying microstructure of the fibers. A new structure-based tensor is presented to map the macroscopic deformation to the fiber stretch. One additional parameter is introduced to tune the coupling of stretch in two directions in order to capture the relatively weak anisotropic coupling. The multiaxial tensile tests with a range of anisotropic loading ratios, which was intended to capture the full spectrum of anisotropic loadings, were conducted on animal artery tissues. Model parameters were determined either directly from histological measurements or calibrated by analyzing the physical nature. Very good agreement between model and experimental results were obtained. To our best knowledge, it is the first demonstration to show the remarkable fitting results for the loading ratios that cover the full spectrum of anisotropy.

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1. Introduction

Biological tissues are often made of soft composite materials. Constitutive modeling of mechanical behaviors of soft composite materials is of great interest to biomedical research as well as to the continuum mechanics field. In biomedical research, it is desirable to have a model that can connect the stress-stretch behaviors to the underlying structures so that the mechanical behaviors of tissues can be used as a diagnostic tool. In continuum mechanics, the stress-stretch behaviors represent challenging problems of constitutive models for anisotropic soft composite materials.

Mechanical behaviors of proximal artery tissue play an important role in the cardio-vasculature. In a cardiac cycle, the pressure within the artery varies from diastole to systole, accompanied by changes of ~20%–30% in the vessel diameter (Lammers et al., 2008). The artery stiffness within the operating range from diastole to systole determines the load to the heart, contributing to hemodynamics and heart health. The stress-stretch behavior of artery tissue demonstrates two salient features. First, it has a characteristic J-shaped behavior, i.e., a small deformation resistance at low stretch followed by an increase in stiffness after some stretch. Depending on the disease state of the tissue, the transition from low to high stiffness, termed collagen engagement, can be relatively gradual or sudden. Second, the stress-stretch behavior is orthotropic, which is typically seen when the tissue is stretched in the longitudinal (or axial) and circumferential directions, respectively. Although the stress-stretch behavior in both directions can demonstrate similar J-shaped behaviors, the initial stiffness, engagement point and high-stretch stiffness can be substantially different. These two features of the stress-stretch behavior of artery tissue are due to the underlying structure.

Arterial tissue is comprised of four main components, including smooth muscle cell, elastin, collagen and non-proteinous media. The contributions from smooth muscle cells can be neglected if there is no smooth muscle cell activation, and their passive behavior can be lumped in with the matrix. Elastin forms a cross-linked network and presents mainly in the medial layer in the form of fenestrated lamellae. It is accepted that in the normal operating range of the artery, arterial elastin has almost linear stress-stretch behavior (Gosline et al., 2002; Humphrey, 1995; Humphrey, 2002; SR et al., 2008). In addition, recent studies on elastin networks from digested tissue indicate that the mechanical behavior

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https://doi.org/10.1016/j.ijsolstr.2018.01.025
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of the elastin network is also orthotropic (Lillie et al., 2010; Rezakhaniha and Stergiopulos, 2008; Sherebrin et al., 1983; Zou and Zhang, 2009). Collagen fibers in the form of undulated bundles are arranged in a loosely cross-linked mat. Collagen fibers have a wavy shape and do not carry load until they are straightened. It is generally believed that straightening collagen fiber at moderate to high stretch causes a dramatic increase in tissue stiffness. The fiber orientations of both collagen and elastin can determine the degree of anisotropy of the artery wall.

A number of constitutive models were developed to capture the mechanical behaviors of artery tissues. A widely used model was developed by Fung et al. (1979), who used an exponential type of strain energy function with parameters characterizing the different directions and the coupling between them. This approach was used in many later phenomenological models (Demiray, 1972; Holzapfel and Weizsacker, 1998; Takamizawa and Hayashi, 1987; Vaishnav et al., 1972). However, these models do not draw upon the underlying microstructure. Bischoff et al. (2002) developed an orthotropic model based on the eight-chain representation of the fiber network, but used a rectangular prism with variable side lengths for orthotropy. Holzapfel et al. (2000) and Guo et al. (2006) presented a model with two families of fibers representing the collagen fiber bundles. These fibers were modeled with two families of fibers in the circumferential-longitudinal plane at an angle to the circumferential direction and symmetric about that direction. Later, Gasser et al. (2006) presented an improved model using a distributed fiber orientation for each family of fibers, and Lopez–Pamies et. al., and co-workers (2006,2010) proposed a homogenization theory to model the mechanical response of hyperelastic solids reinforced by a random distribution of aligned cylindrical fibers, but these models do not reflect the microstructural arrangement of collagen. It has begun to be understood recently that the residual stress plays an important role in artery modeling (Merodio et al., 2013; Merodio and Ogden, 2016). The J-shaped stress-strain behavior due to collagen engagement is a characteristic feature of arterial materials. Previous works by Holzapfel et al. (2000) and Zulliger et al. (2004) to capture this behavior used a piecewise function with a predefined critical stretch ratio. However, using a piecewise function, although convenient, does not consider the energy due to bending of the collagen fibers. The stress-strain behavior of artery walls of a Marfan's syndrome patient exhibits a double curvature response rather than a J-shaped response for the healthy people. The mechanical response of the diseased artery walls shifts to the right compared with the healthy artery (Merodio and Haughton, 2010). This shift has been explored extensively to explain aneurysm formation and propagation in Alhayani et al. (2014).

Among the constitutive models for arterial tissues, those motivated by the material structures are particularly attractive. Zulliger et al. (2004) presented a model that accounted for the percentage of elastin and collagen from histology. They, however, did not consider the morphology of the constituents. The models presented by Sacks (2003) and Caner et al. (2006) use the collagen fiber orientation distribution measured from experiments. This allows an accurate model of the anisotropic behavior under multiaxial loading. Modeling of elastin anisotropy has been explored by Rezakhaniha and Stergiopulos (2008) who found that a transversely-isotropic elastin model produced a better fit for tube inflation-extension tests than models with isotropic elastin behavior. Recently, a microstructure driven constitutive model was developed by the authors to capture the anisotropic stress–strain behaviors for arterial tissue (Kao et al., 2010; Kao et al., 2011). The tissue is assumed to be composed of elastin matrix reinforced by collagen fiber bundles. The latter were modeled as a wavy elastic beam, to capture the J-shaped behavior. Anisotropic behaviors of the tissue were attributed to the anisotropic behaviors of the elastin matrix as well as the uneven distribution of collagen fiber bundles.

From the continuum mechanics point of view, it is important to develop a model that can capture the anisotropic hyperelastic behaviors under the full spectrum of anisotropic loadings. To our best knowledge, most of the constitutive models developed so far either were not tested under such experimental conditions, or failed to capture the experimental results. For example, in our previous model, if the uniaxial tests were used to fit material parameters, the equal biaxial stress would be over-predicted (Kao et al., 2011) at the intermediate stretch level. It is therefore important to more carefully investigate the stress anisotropic coupling at the intermediate stretch level. The goal of this paper is to develop an anisotropic hyperelastic constitutive model to fully capture the anisotropic behaviors of soft composite materials under a full spectrum of stress ratios. We will also use a combined approach to determine model parameters, where part of the model parameters was obtained from histology. The paper is organized as follows. In Section 2, a new structure tensor-based constitutive model is presented. In Section 3, results of the planar biaxial mechanical testing and histology measurements are presented, from which some constitutive model parameters are extracted. Section 4 presents the identification for other model parameters and the overall comparison between the proposed model and the experimental data. Finally, the conclusions are presented in Section 5.

2. Constitutive model

2.1. Preliminary

With the focus on the mechanical response, we consider the tissue as a mixture of the elastin matrix and collagen bundle network. Therefore, we assume that there exists a strain energy density function, which can be additively divided into two contributions

$$W = W^{El} + W^{Col}, \tag{1}$$

where W^{El} and W^{Col} are the strain energy density function for the elastin matrix and the collagen network, respectively. The volume fractions are lumped into the expression of strain energy density function. The deformation of the tissue can be described by the deformation gradient $\mathbf{F}$. The right Cauchy-Green deformation tensor is defined as $\mathbf{C} = \mathbf{F}^T\mathbf{F}$. The first and second invariants of $\mathbf{C}$ are defined as

$$I_1 = tr(\mathbf{C}), I_2 = \frac{1}{2}\left[tr(\mathbf{C})^2 - tr\left(\mathbf{C}^2\right)\right]. \tag{2}$$

Considering the tubular shape of the artery blood vessel, we refer to the vectors that define the circumferential and longitudinal direction as $\mathbf{a}_0$ and $\mathbf{g}_0$, respectively. Two additional invariants, typically used in anisotropic hyperelasticity are defined as

$$I_4 = \mathbf{C} : \mathbf{a}_0 \otimes \mathbf{a}_0, \quad I_6 = \mathbf{C} : \mathbf{g}_0 \otimes \mathbf{g}_0. \tag{3}$$

The strain energy density function relates the deformation gradient through the invariants $W = W(I_1, I_2, I_4, I_6)$. We assume the tissue to be incompressible. The stress, work conjugated with the deformation gradient is calculated as

$$\mathbf{s} = \left[2\frac{\partial W(\mathbf{C})}{\partial \mathbf{C}} - p\mathbf{C}^{-1}\right]\mathbf{F}^{\mathbf{T}}, \tag{4}$$

where p is the hydrostatic pressure. The Cauchy stress is calculated as $\sigma = J\mathbf{F}\mathbf{s}$, where J is the determinant of the deformation gradient. In order to describe the stress-stretch behavior of the tissue, we will construct the strain energy density function W. In the following formulation we call the circumferential direction as direction 1 and the longitudinal direction as direction 2.

Please cite this article as: T. Lu et al., A micro-structure based constitutive model for anisotropic stress–strain behaviors of artery tissues, International Journal of Solids and Structures (2018), https://doi.org/10.1016/j.ijsolstr.2018.01.025

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