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Modeling of human artery tissue with probabilistic approach

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ABSTRACT

Accurate modeling of biological soft tissue properties is vital for realistic medical simulation. Mechanical response of biological soft tissue always exhibits a strong variability due to the complex microstructure and different loading conditions. The inhomogeneity in human artery tissue is modeled with a computational probabilistic approach by assuming that the instantaneous stress at a specific strain varies according to normal distribution. Material parameters of the artery tissue which are modeled with a combined logarithmic and polynomial energy equation are represented by a statistical function with normal distribution. Mean and standard deviation of the material parameters are determined using genetic algorithm (GA) and inverse mean-value first-order second-moment (IMVFOSM) method, respectively. This nondeterministic approach was verified using computer simulation based on the Monte-Carlo (MC) method. Cumulative distribution function (CDF) of the MC simulation corresponds well with that of the experimental stress–strain data and the probabilistic approach is further validated using data from other studies. By taking into account the inhomogeneous mechanical properties of human biological tissue, the proposed method is suitable for realistic virtual simulation as well as an accurate computational approach for medical device validation.

1. Introduction

Accurate modeling of biomechanical properties of artery tissue is important for developing realistic medical simulation systems which are commonly used in surgical training, planning and treatment, and diagnostic tools for vascular diseases. With the advancements in medical imaging technologies and mapping tools, model personalization has generated a lot of research interest [25,38]. Efficient probabilistic model personalization integrating the specific patient data to a generic model enables the application of computational approaches in clinical practice [1]. This study characterizes the patient specific material parameters and validates the models using experimental data of human iliac vessels.

Nonlinear hyperelastic property is an important attribute of the artery tissue which can contribute to modeling accuracy, and hence needs to be considered for realistic deformation simulation and haptic rendering in surgical simulation of the arterial wall [43]. Experimental procedures such as inflation tests [17], biaxial tests [42], as well as tension and indentation tests [11,26] have been performed to study the mechanical properties of arteries tissue. Those experiments revealed that the mechanical behaviour of arteries is elastic, highly non-linear and anisotropic under finite strains for slow monotonic tests, and have been adequately modeled within the framework of hyperelasticity.

Many constitutive models have been proposed for mathematical description of the mechanical behaviour of artery tissue [15,22–34]. Strain energy based models are able to provide comprehensive understanding of the inter-relationship between stress and strain. In Chui et al.'s study [5], the combined logarithmic and polynomial model was reported to perform better than the combined exponential and polynomial model [15] in modeling the stress–strain relationship of liver tissue. The combined logarithmic and polynomial model has been utilized for realistic modeling of porcine artery tissue [47]. Structural strain energy function is proposed to study the arterial tissue [48]. The proposed function includes the wavy nature of the collagen and the fraction of both elastin and collagen contained in the media, which can be determined by histology. The waviness of the collagen is assumed to be distributed log-logistically. The novel strain energy function is found to behave similarly to that of Holzapfel et al. [22], both succeed in describing the typical S-shaped pressure–radius curves with comparable quality of fit.

Uniaxial stress tests have been performed on human artery tissue to study its nonlinear biomechanical properties in this work. Due to varying micro-structural composition of tissue specimens

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and inherent noises from experimental instruments, the measured stress–strain data often comprise of a number of stress–strain curves with large deviations [36]. Material parameters of soft tissue model are usually characterized using the mean stress–strain curve and ignoring the stress–strain curve deviations. However, it is important to incorporate the inherent stiffness variations for realistic medical simulation [14]. These inherent stiffness variations of artery tissue may be modeled using probabilistic uncertainty analysis. The applicability of probabilistic uncertainty analysis has been demonstrated in evaluating structural reliability [9,10], and knee ligamentous constraint analysis [4]. A probabilistic method was proposed to model the mechanical properties of liver tissues in [14]. To our best knowledge, few reports have been published on probabilistic analysis of human artery for characterization and deformation simulation in the literature.

In this study, we introduce a probabilistic approach to model the inhomogeneity, specifically the stiffness variations of human artery tissue. Cyclic tension tests in longitudinal and circumferential directions were performed on human arteries. The arteries are highly inhomogeneous with large variances in stress–strain curves. In order to study the anisotropic mechanical properties of arterial tissue, a combined logarithmic and polynomial constitutive equation [5,47] is employed to model the average stress–strain relationship of human artery tissue in circumferential and longitudinal directions. The material parameters are expressed as a statistical function with normal distribution. The mean values of the material parameters are identified using GA while the standard deviation of the material parameters are determined using direct calculation method (DC) and inverse mean-value first-order second-moment (IMVFOSM) method, respectively. The probabilistic approach is then verified using Monte-Carlo (MC) method and demonstrates good correspondence with cumulative distribution function (CDF) between simulated and experimental stress–strain data. Validation of the material parameters are carried out with experimental human artery data.

In Section 2, the probabilistic approach is introduced and material parameters are determined from the scattered experimental stress–strain data. Section 3 presents the verification of derived material parameters through MC simulation and historical experimental data. Section 4 concludes the study with a short summary.

2. Materials and methods

2.1. Elongation tests on artery samples

There are three types of arteries: elastic, mixed and muscular. Abdominal aorta, femoral artery and popliteal artery are examples of elastic, mixed and muscular arteries respectively [6]. Uniaxial elongation tests were performed on 20 samples of human femoral arteries from 5 donors in circumferential and longitudinal directions, average 4 samples from each donor. The experiments were carried out with a mechanical property measurement system which was designed to meet the requirements of automated environment control, testing, and data collection for biological tissue [47], as shown in Fig. 1. Load cells (LCM UF series) were employed to measure force imposed on specimen. Human femoral arteries were obtained from human donors and stored in an ice box with Histidine Tryptophan Ketoglutarate (HTK) solution before experiment for about 2 weeks' time. The arteries were intended for organ transplant, and the donors' identities were protected in according to an approved Institutional Review Board protocol. All arterial specimens were excised from the distal end of the abdominal aorta, with lengths between 40 and 50 mm and

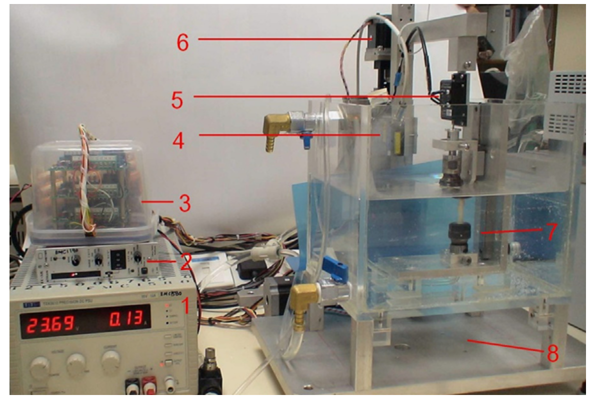

Fig. 1. The mechanical testing system; (1) Power source, (2) strain gauge amplified for load cell and pressure transducer (not shown), (3) stepper motor control, (4) distance laser sensor, (5) load cell, (6) translational stage with stepper motor, (7) clamping feature and fixture, (8) base.

diameters between 6 and 11 mm. During experiment, the artery specimen was immersed in Krebs Ringer solution at 37 °C throughout the experiment, with pH value maintained at 7.3–7.4 using carbon dioxide, which is designed to improve the test accuracy by mimicking the inner environment of human body.

Specimens were tested in both circumferential and longitudinal directions. A ring segment was sliced off from one end of each arterial specimen and tested in circumferential direction. The remaining section of the specimen was used to perform tests in the longitudinal direction. All specimens were preconditioned before data collection over a course of at least five cycles, allowing the specimens to reach a steady state as no further changes occurred on the stress–strain curves [16,44]. During the experiments, force and displacement data were measured and recorded. Stretch ratio λ was calculated by dividing instantaneous gauge length by its original length while engineering stress T was calculated by dividing the instantaneous load by the original cross-sectional area. Strain ε can be expressed as $\lambda - 1$. The tension tests were performed until the stretch ratio reached 1.55 for both directions at a ramping speed of 2.5 mm/s.

Two sets of scattered stress–stain curves which represent the data in circumferential and longitudinal directions were plotted together with its mean curve in Fig. 2. Each set includes twenty curves. The experimental stress with respect to a specific stretch ratio is assumed to vary according to normal distribution. Mean value and standard deviation of stress is calculated as:

$$\mu_T = \frac{1}{N}\sum_{i=1}^{N} T_i, \tag{1}$$

$$\sigma_{T_i} = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(T_i - \mu_T)^2}, \tag{2}$$

where T_i denotes the experimental stress value and N is the number of experimental stress values. Experimental results of minimum, mean and maximum stress–strain curve are also illustrated in Fig. 3. The probability density function (PDF) of the stress values calculated using Eqs. (1) and (2) at three different strain values are plotted in Fig. 3 which shows the normal distribution of stress values. The experimental data is tested using Shapiro–Wilk test [39]. The average probabilities of normality in circumferential and longitudinal directions are 0.7185 and 0.7237, which validate the assumption of normal distribution.