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PHENOMENOLOGICAL BASED CONSTITUTIVE MODELING OF JUGULAR VENOUS TISSUE

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INTRODUCTION

Venous valve tissues, while used in vein reconstruction surgeries and bio prosthetic valves with moderate success, does not have extensive studies on their structure and modelling. Their inherent anisotropic, non-linear behavior combined with severe diseases inflicting the veins like chronic venous insufficiency warrants understanding the structure and material behavior of venous valve tissues. Hence, before any bio-prosthetic grafts can be used in place of tissues, it is of utmost importance to know about the mechanical and structural properties of the tissue which can lead to an increase in success rates of valve replacement surgeries. The longevity of the bio-prosthetics would also increase if the manufactured grafts behave exactly similar to the native valves.

With the scant information on uniaxial and biaxial mechanical properties of jugular venous valve and wall tissue from any previous studies, the current focus of our study was to understand the material behavior by determining an established phenomenological strain energy based constitutive relation for the tissues. We used bovine veins to study the behavior of valve leaflet tissue and adjoining wall tissue (from proximal and distal end of the vein) under different biaxial testing protocols. We looked at the behavior of numerical partial derivatives of strain energy to select a suitable functional form of strain energy for wall and valve tissue from the literature. Using this strain energy descriptor, we determined Cauchy stress and compared it with experimental results under displacement controlled biaxial testing protocols to find material specific model parameters by Powell's method algorithm. Results indicated that whereas wall tissue strain energy can be explained using a polynomial nonlinear function, the valve tissue, due to higher nonlinearities, requires an exponential function. This study can prove helpful in primary stages of bio-prosthetic designs; replacement surgeries; can be of support to any future studies investigating structural models and to study valvular diseases by giving a way to understand material properties, behavior and to form a continuum model when required for numerical analyses and computational simulations.

METHODS

Structural exploration studies on jugular venous tissues have shown that tissue has incompressible homogeneous structure. Valve tissues are known to have bundles of aligned collagen fibers. The belly region of the valve has more homogeneous structure compared to the commissures or attachment area. Few studies done towards the mechanical behavior aspect of the Jugular venous tissue indicates non-linear and anisotropic behavior for the valve and wall tissue, valve having more nonlinearities. This information enables us to safely assume that the tissue under study, valve and wall alike, are pseudo-elastic, incompressible, homogeneous and locally transversely isotropic with respect to fiber axis.

Behavior of such hyper-elastic soft tissues can be explained using strain energy expression - 'W' which is a function of two invariants of the deformation [1] i.e. I_1 and I_4 (= α^2), along with the constitutive equation given as follows, where all symbols have their standard meanings.

$$\boldsymbol{t} = -p\boldsymbol{I} + 2W_1\boldsymbol{B}(\frac{W_\alpha}{\alpha})\boldsymbol{F}\ \boldsymbol{N}\otimes\boldsymbol{N}\ \boldsymbol{F}^T \quad \textbf{(1)}$$

The strain energy descriptor was decided on the basis of experimentation which involved biaxial testing where alternatively each invariant was held at a constant value, called "constant invariant testing." The plots of $W_1 = \dfrac{\partial W}{\partial I}$ and $W_\alpha = \dfrac{\partial W}{\partial \alpha}$ with respect to the invariants, termed as "response curves," provides an empirical basis to decide a suitable function for the strain energy descriptor.

Tissue Procurement & Preparation: Due to issues associated with procurement and testing human tissues and the fact that in practice, bovine tissues are used on a large scale as a suitable substitute from medical point of view in bio prosthetic grafts; in the current study, we used tissue samples from a mature bovine of Holstein breed, 10+ years old with an approx. weight of 1250 lbs. Valve and wall tissue samples were extracted from jugular veins of the animal. Vein samples were shipped overnight on ice, ~24 hours post-slaughter. Test specimens were cut from the belly region of the valve and avalvular region of the wall in square section of side 7-10 mm and stored in Hanks balanced salt solution (HBSS).

Testing Protocols: Discussed in detail in a previous study [2], we mount the tissue test specimens on the biaxial testing apparatus using rakes in a bath of HBSS pre-heated to 37 degrees Celsius to imitate the physiological conditions. The effective central region subject to stretching is 4.5 mm x 4.5 mm. The test rig has two load cells (10N±0.02N) mounted perpendicular to each other and two actuators opposite to load cells to simulate the loading or stretching. The displacement of the actuators is recorded along with the load cells readings. Test Specimen is preconditioned with a preload of 10mN for 8 loading-unloading cycle up to 30% strain at a strain rate of 1% per sec followed by a rest period of 300 secs. Thickness of the tissue samples is recorded before testing 4-5 times by using a dial gauge (±0.01 mm; The L.S. Starrett Co., Athol, MA) and then an average value was used for stress calculations. Based on an approximate 60% true strain breaking strength, we tested a total of 16 wall samples along with 8 samples of valve tissue for $3.2 < I_1 < 4.4$ and $1.1 < \alpha < 1.7$.

Model and Parameters: Based on the behavior of the response curves (linear for wall tissue and exponential for the valve tissue), we selected suitable descriptors from the literature: 5-parameter polynomial type strain energy descriptor [1, 3] for the wall tissue and a 3-parameter exponential type function [4] for the valve tissue as shown.

$$W^{wall} = c_1(\alpha-1)^2 + c_2(\alpha-1)^3 + c_3(\mathrm{I}_1-3) + c_4(\mathrm{I}_1-3)(\alpha-1) + c_5(\mathrm{I}_1-3)^2$$

$$W^{valve} = c_0(\exp(c_1(\mathrm{I}_1-3)^2 + c_2(\alpha-1)^4)-1)$$

Note that c's in the above two strain energy functions are different parameters. After selecting the descriptor function, the next step of the process required us to find material parameter values (5 for W^{wall} and 3 for W^{valve}). The expressions for model predicted stress for the wall and valve tissues were found by substituting W_I and W_α in the constitutive relation in **Equation 1**, and as below:

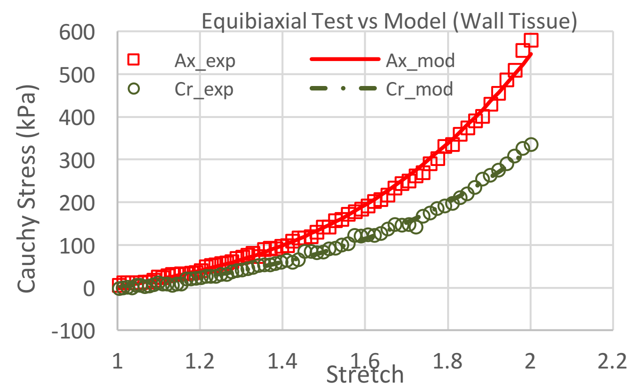
$$\boldsymbol{t}_{11}^{wall} = 2(\lambda_1^2-\lambda_3^2)[c_3 + c_4(\alpha-1) + 2c_5(\mathrm{I}_1-3)] + \lambda_1[2c_1(\alpha-1) + 3c_2(\alpha-1)^2 + c_4(\mathrm{I}_1-3)] \quad \textbf{(2)}$$
$$\boldsymbol{t}_{22}^{wall} = 2(\lambda_2^2-\lambda_3^2)[c_3 + c_4(\alpha-1) + 2c_5(\mathrm{I}_1-3)]$$

$$\boldsymbol{t}_{11}^{valve} = (4c_0c_1(\lambda_1^2-\lambda_3^2)(\mathrm{I}_1-3) + 4c_0c_2\lambda_1(\alpha-1)^3)\exp(c_1(\mathrm{I}_1-3)^2 + c_2(\alpha-1)^4), \quad \textbf{(3)}$$
$$\boldsymbol{t}_{22}^{valve} = 4c_0c_1(\lambda_2^2-\lambda_3^2)(\mathrm{I}_1-3)\exp(c_1(\mathrm{I}_1-3)^2 + c_2(\alpha-1)^4)$$

where λ's are stretches. For parameter estimation, a separate set of biaxial testing was conducted in which separate preconditioned tissues were subjected to 4 off-axial and 1 equibiaxial stretching protocol consecutively (in the order of A: C for wall tissue samples or C: R for valve tissue sample = 2:1, 1:2, 1.5:1, 1:1.5, 1:1) where the ratio indicates the ratio of strains in the orthogonal directions. Best fit values were found for each sample separately by minimizing the sum of the squares of the residuals using the Powell's method algorithm. The residual was defined as the error between experimental measured and theoretically predicted Cauchy stress. A maximum of 70% true strain was selected for this phase of biaxial testing.

RESULTS AND DISCUSSION

Parameter fitting was done for three testing ratios (i.e., 1:1, 1.5:1, 1:1.5) for every wall and valve samples. The material parameters found were used to predict the experimental data for the two remaining test ratios to verify the usefulness of the model. Stress vs. stretch data along with the model fitting for the wall and valve tissue for equi-biaxial test ratio (1:1) is shown in **Figure 1**. The correlation coefficients for all the wall tissues were found above 0.9; whereas for the valve tissue, they were as low as 0.7, indicating some prediction error. The response curves for the wall matched very well to the ones in the myocardium study [1, 3], but the response curves for the valve, while exponential in nature, had some notable differences from the ones presented in mitral valve study [4], and hence the prediction error was expected. It was also observed that the distal wall tissue samples have more strain energy density per unit volume indicating higher forces at distal end compared to proximal end at any instant; which could possibly be attributed to the higher hydrostatic pressure at the distal end of the vein.



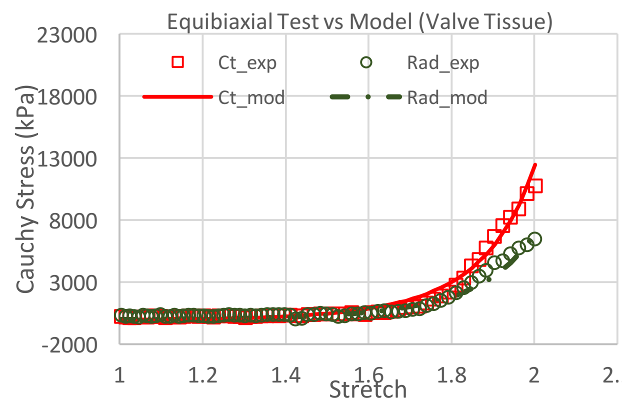


Figure 1: Top: The model fits using the individual material parameters for the wall tissue (Equation 2, $c_1 = 29.7$, $c_2 = 33.7$, $c_3 = 12.3$, $c_4 = -9.90$, $c_5 = 3.82$). Bottom: The model fits using the individual material parameters for the valve tissue (Equation 3, $c_0 = 472.7$, $c_1 = 0.043$, $c_2 = 0.34$).

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