Constitutive Modeling and Fluid-Structure Interactions of 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.

Materials and Methods: First, we used bovine veins to experimentally study the behavior of valve leaflet tissue and adjoining wall tissue (from proximal and distal end of the vein) under different biaxial testing protocols. Second, 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 [1-2]. 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. The expressions for model predicted stress for the wall and valve tissues were shown below:

$$\boldsymbol{t}_{11}^{wall} = 2(\lambda_1^2 - \lambda_3^2)[\mathrm{c}_3 + \mathrm{c}_4(\alpha - 1) + 2\mathrm{c}_5(\mathrm{I}_1 - 3)] + \lambda_1[2\mathrm{c}_1(\alpha - 1) + 3c_2(\alpha - 1)^2 + \mathrm{c}_4(\mathrm{I}_1 - 3)]$$
$$\boldsymbol{t}_{22}^{wall} = 2(\lambda_2^2 - \lambda_3^2)[\mathrm{c}_3 + \mathrm{c}_4(\alpha - 1) + 2\mathrm{c}_5(\mathrm{I}_1 - 3)]$$

and

$$\boldsymbol{t}_{11}^{valve} = (4c_0c_1(\lambda_1^2 - \lambda_3^2)(\mathrm{I}_1 - 3) + 4\mathrm{c}_0\mathrm{c}_2\,\lambda_1(\alpha - 1)^3)\exp(\mathrm{c}_1(\mathrm{I}_1 - 3)^2 + \mathrm{c}_2(\alpha - 1)^4),$$
$$\boldsymbol{t}_{22}^{valve} = 4c_0c_1(\lambda_2^2 - \lambda_3^2)(\mathrm{I}_1 - 3)\exp(\mathrm{c}_1(\mathrm{I}_1 - 3)^2 + \mathrm{c}_2(\alpha - 1)^4)$$

where λ's are stretches. We then implemented our selected constitutive models along with material specific model parameters using user defined material subroutines (UMAT) in a commercial FEM package ABAQUS with fluid-solid interactions (FSI) functionality. We observe the blood flow behavior differences around a bileaflet versus a trileaflet valve during opening/closing in the vein. Isolated and coupled effects from hemodynamic, hydrostatic, and hydrodynamic variations are studied based on the FSI simulations.

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, 2], but the response curves for the valve, while exponential in nature, had some notable differences from the ones presented in mitral valve study [3], 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.

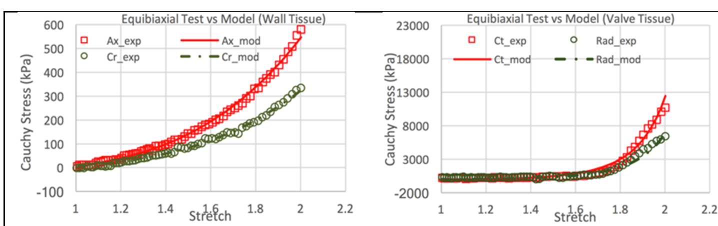

Figure 1: Left: The model fits using the individual material parameters for the wall tissue ($c_1 = 29.7$, $c_2 = 33.7$, $c_3 = 12.3$, $c_4 = -9.90$, $c_5 = 3.82$). Right: The model fits using the individual material parameters for the valve tissue ($c_0 = 472.7$, $c_1 = 0.043$, $c_2 = 0.34$).

Conclusions: 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, and to form a continuum model required for numerical analyses and simulations.

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References:

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