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THE ROLE OF MATERIAL ANISOTROPY IN ABDOMINAL AORTIC ANEURYSM WALL MECHANICS

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ABSTRACT

The prevalence of AAA is growing along with population age and according to different studies AAA rupture is the 13th most common cause of death in the U.S., causing an estimated 15,000 deaths per year. In biomechanical terms, AAA rupture is a phenomenon that occurs when the developing mechanical stresses within the aneurysm inner wall, as a result of the exerted intraluminal pressure, exceed the failure strength of the aortic tissue. To obtain a reliable estimation of wall stress, it is necessary to perform an accurate three-dimensional reconstruction of the AAA geometry and model an appropriate constitutive law for the aneurysmal tissue material characterization. In this regard, a recent study on the biaxial mechanical behavior of human AAA tissue specimens [1] demonstrates that aneurysmal arterial tissue behaves mechanically anisotropic. The objectives of the present work are to determine the effect of material anisotropy of the aneurysmal abdominal aorta on wall stress distribution and to establish a comparison of wall mechanics between ruptured and unruptured aneurysms.

We consider the aneurysm wall as a hyperelastic material and postulate the existence of a strain-energy function (SEF) W, from which the stress-strain behavior of the material can be derived [2]. In this investigation, isotropic and anisotropic constitutive models have been assumed for the aneurysmal wall. For the case of isotropy the material response of the aneurysm is characterized by the following SEF [3]: $W(\mathbf{C}) = U(J) + c_{10}(\bar{I}_1 - 3) + c_{20}(\bar{I}_1 - 3)^2$, where c_{10} and c_{20} are constants with dimension of stress. For the case of anisotropy, the aneurysmal tissue has been modeled as a hyperelastic material reinforced with two families of fibers aligned along two directions arranged in a double-helix pattern. Therefore, the SEF is expressed as $\overline{W} = \overline{W}(\overline{\mathbf{C}}, \mathbf{n}_0, \mathbf{m}_0)$, with $\mathbf{n}_0$ and $\mathbf{m}_0$ denoting the directions of collagen fibers within the tissue. In addition, $\overline{W}$ is additively decomposed into an isotropic contribution, corresponding to the matrix material, and an anisotropic contribution related to the (two families of) collagen fibers [4], i.e. $W(\mathbf{C}) = U(J) + c_{10}(\bar{I}_1 - 3) + \frac{k_1}{2k_2}\left[e^{k_2(\bar{I}_4 - 1)^2} - 1.0\right] + \frac{k_3}{2k_4}\left[e^{k_4(\bar{I}_6 - 1)^2} - 1.0\right]$, where c_{10} is a stress-like material parameter and $k_1 \ldots k_4$ are material parameters corresponding to the fibers. These parameters were obtained by means of a nonlinear regression analysis of the membrane solution to the experimental data reported by Vandegueest et al. [1]. Since the presence of intraluminal thrombus redistributes the stress at the inner wall, in this study the ILT has been included and modeled as an isotropic hyperelastic material with the SEF [5] $W(\mathbf{C}) = U(J) + c_{01}(\bar{I}_2 - 3) + c_{02}(\bar{I}_2 - 3)^2$, where c_{01} and c_{02} are material constants with units of stress. The three dimensional material constitutive models

have been implemented within the multi-purpose finite element program ABAQUS by means of user material subroutines (UMAT). The orientation of collagen fibers has been included in the model by defining a tangent at each integration point. The vectors are oriented according to the angle between the fiber reinforcement and the circumferential direction of the wall. CT scans were obtained retrospectively from 6 AAA subjects (3 ruptured and 3 unruptured aneurysms) treated at Allegheny General Hospital in Pittsburgh, Pennsylvania. The corresponding DICOM images were imported into an in-house Matlab based image segmentation code (VESSEG v.1.0.1, Carnegie Mellon University), for the lumen, inner wall, and outer wall segmentations. Three-node shell elements (quadratic, with 5 integration points through the thickness) were used for meshing the arterial wall, while ten-node tetrahedral elements (quadratic, with 4 integration points) were used to mesh the thrombus. The number of shell elements for the AAA wall ranges from 60000 to 165000 while the number of tetrahedral elements for the ILT ranges from 23000 to 100000, dependent on the size and shape of each patient-specific geometry. Quasi-static analyses were performed at a peak systolic arterial pressure load of 120 mmHg applied on the intraluminal surface of the AAA. To estimate peak wall stress, the maximum principal stress was evaluated at the third integration point of the wall elements. Figure 1 illustrates the wall stress distribution for two representative unruptured AAA models, U1 and U3, along the anterior wall surface. While the actual distribution of the stress is similar for both material models, incorporating collagen fiber directions and anisotropy in the characterization of the AAA wall material results in a significantly higher peak wall stress. The mean peak wall stresses (in kPa) are 307.8 ± 49.7 and 702.6 ± 166.2, for the isotropic and anisotropic material models, respectively, of the three unruptured AAA geometries analyzed in the present work. A comparison between ruptured and unruptured wall mechanics on the basis of the anisotropic material model yields mean peak wall stresses (in kPa) of 977.9 ± 179.6 and 702.6 ± 166.2, for the ruptured and unruptured geometries, respectively. The anisotropic characteristics of the AAA wall, previously verified by means of biaxial tensile testing of AAA tissue specimens, also play a role in the distribution of wall stress yielding higher peak wall stresses that likely influence the assessment of rupture potential in *individual* AAAs.

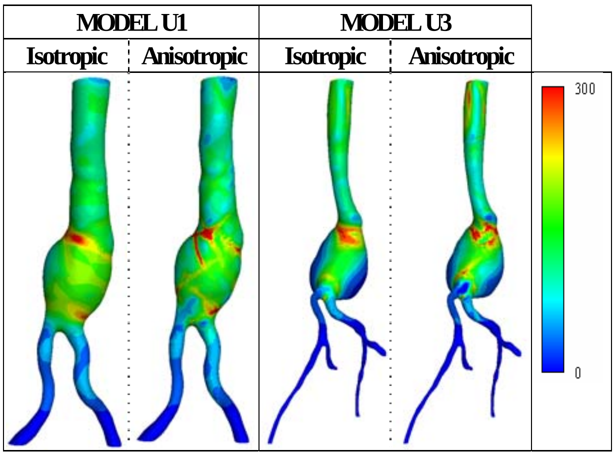

Figure 1. Comparison of isotropic and anisotropic wall stress distributions (in kPa) for models U1 and U3.

REFERENCES

[1] J.P. Vande Geest, et al., “The effects of aneurysm on the biaxial mechanical behavior of human abdominal aorta,” *J Biomech*, Vol. **39**, pp. 1324-1334, (2006).

[2] G.A. Holzapfel, *Nonlinear solid mechanics. a continuum approach for engineering*, John Wiley & Sons, Chichester, 2000.

[3] M.L. Raghavan and D.A. Vorp, “Toward a biomechanical tool to evaluate rupture potential of abdominal aortic aneurysm: identification of a finite strain constitutive model and evaluation of its applicability,” *J Biomech*, Vol. **33**, pp. 475-482, (2000).

[4] G.A. Holzapfel, et al., “A new constitutive framework for arterial wall mechanics and a comparative study of material models,” *J Elast*, Vol. **61**, pp. 1-48, (2000).

[5] E.S. Di Martino and D.A. Vorp, “Effect of variation in intraluminal thrombus constitutive properties on abdominal aortic aneurysm wall stress,” *Ann Biomed Eng,* Vol. **31**, pp. 804–809, (2003).