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ON MODELING MULTI-LAYERED SOFT COLLAGENOUS TISSUES

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ABSTRACT

Several connective tissues consist of multiple layers and are composed primarily of collagen, a protein which gives the tissue strength which varies with, for example, age and pathology. Cerebral aneurysmal walls, for example, are multi-layered structures with mean fiber alignments distinguishing one layer from another [1]. Walls of conducting arteries such as the aorta, subclavian, carotid, and iliac (in health and disease) are other examples of soft collagenous tissues. They consist of multiple (up to 70) fenestrated elastic laminae separating the organized media into well-defined concentrically fiber-reinforced layers. The number of elastic laminae decreases toward the periphery, as the size of the vessels decreases. The fenestrated laminae allow substances to diffuse and nourish cells in the vessel wall. Quadricep tendons are also composed of multiple layers of collagenous tissue, which are separated to form two strands. Airway walls are multi-layered structures as well.

A new versatile constitutive model for the mechanical response of multiple layers in collagenous tissues is presented [4]. The model has five parameters (E_1, E_2, c, a, β), assumes isochoric deformation and is based on the constitutive theory of finite elasticity. It considers a set of layers with a mean collagen fiber alignment distinguishing one layer from another. The collagen fibers are embedded in a non-collagenous matrix material assumed to behave isotropically. The constitutive response of the collagen fabric is transversely isotropic for each layer and is based on an exponential function of the form proposed in [2]. The isochoric strain-energy function $\overline{\Psi}$ is

$$\overline{\Psi} = \frac{c}{2}\left(\bar{I}_1 - 3\right) + \sum_{i=1}^{n} \frac{k_i}{2a}\left\{\exp\left[a(\bar{I}_{4\,i} - 1)^2\right] - 1\right\}, \qquad \bar{I}_{4\,i} = \overline{\mathbf{C}} : \mathbf{A}(\phi_i, \beta) > 1,$$

where the index i denotes layer-specific entities, and n is the number of tissue layers. The first term in this equation relates to the non-collagenous matrix material (classical neo-Hookean model), while the second term is related mainly to the collagen fibers. The angle of the fibers in layer i is denoted by ϕ_i, and β describes the rotation of the principal directions of the collagen fabric with respect to a (local) reference system. The quantity $\mathbf{A}$ is a structural tensor, defined as $\mathbf{A} = \mathbf{M} \otimes \mathbf{M}$, with components

$[\mathbf{M}] = [\cos(\phi_i+\beta)\ \ \sin(\phi_i+\beta)\ \ 0]^{\mathrm{T}}$, $\overline{\mathbf{C}}$ is the modified right Cauchy-Green tensor, k_i are non-negative layer-specific material parameters, and the exponent $a > 0$ describes the degree of nonlinearity that the collagen fibers exhibit (assumed to be layer-independent). Finally, we introduce the two stiffness measures $E_1 = \sum_{i=1}^{n} k_i \cos^4\phi_i$, $E_2 = \sum_{i=1}^{n} k_i \sin^4\phi_i$ of the collagen fabric in the in-plane principal directions, which provide a relationship with k_i. It will be shown how this constitutive model can be used to numerically simulate, for example, extension-inflation tests of human adventitial tissues [5].

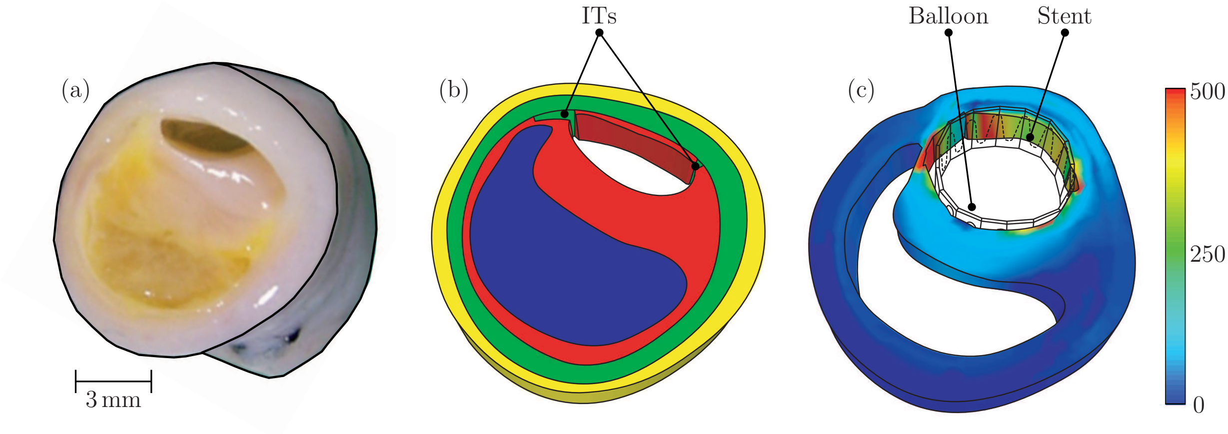

Figure 1: (a) Macroscopic view of an atherosclerotic-prone human iliac artery; (b) 3D geometric model from an extracted arterial slice obtained from MRI and reconstructed by means of NURBS. Four arterial tissues are modeled: adventitia (A), media (M), intima (I), lipid pool (I-lp). The introduced initial tears (ITs) are also indicated; (c) Deformed configuration and distribution of the maximum principal Cauchy stresses (in kPa) at a balloon and stent inflation of approximately 8 bar (the lipid pool is not shown).

Figure 1(a) shows an atherosclerotic-prone human iliac artery which consists of several layers of soft collagenous tissues, while Fig. 1(b) shows the related 3D geometry of an extracted arterial slice reconstructed by non-uniform rational B-splines (NURBS), which serves as a model for the finite element analysis. The talk concludes with a numerical analysis of the 3D contact between a balloon-stent and a stenotic lesion including two initial tears [3]. Figure 1(c) shows the deformed shape and the resulting stress field at full inflation. The balloon is inflated until its outer diameter reaches 4.5 mm, which corresponds to an approximate internal pressure of 8 bar. A stress value of approximately 500 kPa is obtained at both tips at full balloon inflation, which is the highest stress in the tissue. Human atherosclerotic plaques of iliac arteries rupture at a stress level similar to that computed here. It is interesting to observe that in the diseased part of the intima (I) a stress-shield in form of an arc is present.

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