A Method for Incorporating Residual Stresses into Patient-Specific Finite Element Simulations of Arteries with an Example on AAAs

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ABSTRACT

Through progress in medical imaging, image analysis and finite element (FE) meshing tools it is now possible to extract patient-specific geometries from medical images of abdominal aortic aneurysms (AAAs), and thus to study clinically-relevant problems via FE simulations. Such simulations allow additional insight into human physiology in both healthy and diseased states. Medical imaging is most often performed in vivo, and hence the reconstructed model geometry in the problem of interest will represent the in vivo state, e.g., the AAA at physiological blood pressure. However, classical continuum mechanics and FE methods assume that constitutive models and the corresponding simulations begin from an unloaded, stress-free reference condition.

Two problems exist when applying such classical approaches to patient-specific simulations of arteries: (i) the in vivo determined 'initial' geometry is not an unloaded reference configuration, and (ii) the unloaded tissue itself is residually-stressed [1, 2]. The first problem is overcome by computational methods of prestressing the FE model so that, e.g., the initial (image derived) geometry is in equilibrium with the known loads, cf. Gee et al. [3]. The second problem, that of in vivo residual stresses in patient-specific simulations, has still not been satisfactorily addressed in the biomechanics literature.

We propose a pragmatic approach to incorporate experimentally-determined residual stresses (stretches) into FE simulations of arterial tissues, including geometrical models built from patient-specific image data. We calculate a residual deformation gradient for each element, which reflects the kinematic transformation between a (nearly) stress-free reference configuration and the actual (possibly patient-specific) geometry. We demonstrate the accuracy of the method by recreating analytically-determined 3D residual stress distributions from Holzapfel and Ogden [4] using detailed experimental data published on residual deformations from Holzapfel et al. [5], cf. Fig. 1. Finally, we exercise the method on a complete patient-specific FE simulation of a human AAA using realistic anisotropic constitutive models (Gasser et al. [6] for the aortic tissue and Holzapfel et al. [7] for the intraluminal thrombus (ILT)), and material and structural parameters. Furthermore we compare simulated stress distributions with and without the inclusion of residual stresses (stretches).

Including residual stresses in patient-specific FE simulations of arterial tissues is very important as both the global (organ-level) deformation and the internal stress distribution are significantly impacted. Residual stresses in the aortic wall may also influence the stress distribution in the ILT.
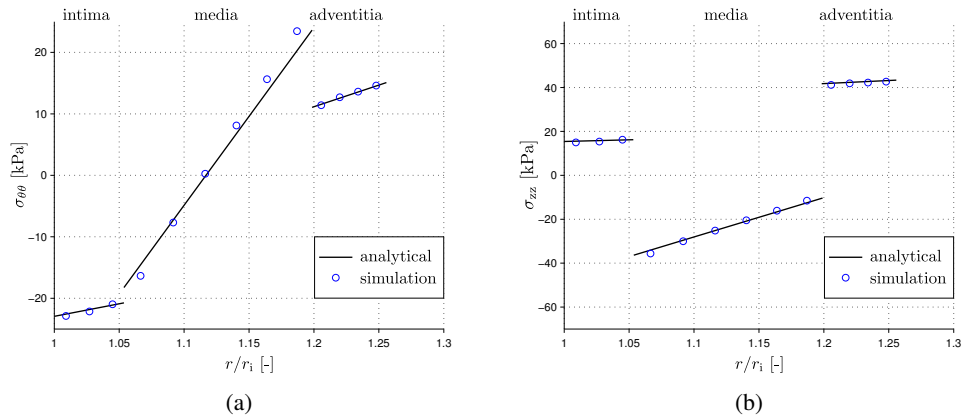
Figure 1: Stress vs. normalized radius for 3-layer arterial tissue in the unloaded, residually-stressed state, comparison of analytical result [5] and simulation: (a) circumferential $\sigma_{\theta\theta}$, (b) axial σ_{zz}.

Our method is applicable to any residually-stressed soft biological tissue where experimental data on the residual deformation are available. Such methods, coupled with appropriate experimental data, aim at increasing the accuracy of classical FE analyses for patient-specific studies in computational biomechanics and may lead to increased clinical application of simulation tools.

References

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