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FULLY-COUPLED FLUID-STRUCTURE INTERACTION SIMULATION OF ASCENDING THORACIC AORTA BASED ON THE "SMALL ON LARGE THEORY"

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Abstract. Numerical hemodynamics simulations remain a challenge in the world of clinical research. In this context, complex interactions require to be modeled to have reliable results. In particular, the interaction between the vascular mechanics and the fluid dynamic conditions of the vessels cannot be neglected. This aspect is fundamental especially for the aortic structure. Additionally, the vascular tissue require hyperelastic and anisotropic mechanical models. The presented work presents a new numerical approach for Fluid Structure Interaction (FSI) simulation of the aortic structure. The proposed method includes a local linearization of a fiber-based constitutive model, accounting for both hyperelasticity and anisotropy.

1 INTRODUCTION

Hemodynamics simulations are an important field of research in the context of computational fluid dynamics. Different computational techniques are available within the state of the art. Particular interest is given to large blood vessels, like the aorta. The Fluid-Structure Interaction (FSI) stands out as the most reliable approach to this kind of research issues. Nevertheless, limitations and space for improvements remain. The requirement for always more accurate patient specific conditions remains. Additionally, the computational load of FSI aorta simulations remains onerous for most cases. Non linearities and anisotropy cannot be neglected for aortic vessels. It is well established that vascular tissue is highly hyperelastic and anisotropic. These aspects contribute to the increment of the computational time and cost. Different research groups assume elastic material approximations for the sake of simplicity [1]. Nevertheless, this assumption remains a strong limitation and a source of inaccuracies. It would be suitable to provide an approach for introducing an acceptable level of approximation, without neglecting the hyperelastic/anisotropic features of the aortic tissue.

The current work presents a new computational approach for the implementation of the Small On Large (SOL) linearization technique [2] on a fully-coupled FSI environment. In particular, a patient specific case of an ascending aortic aneurysm (aTAA), both in terms of geometry and fluid dynamic conditions, is reported. The SOL approach allowed for the definition of linearized local material properties, pre-stress inclusion and anisotropic behavior. The results both in terms of fluid dynamic and mechanical features are presented.

2 MATERIALS AND METHODS

The entire FSI-SOL implementation was performed by integrating both in-vivo imaging and ex-vivo data. In-vivo 4D flow MRI and CT datasets were used to reconstruct the 3D geometries at diastolic phase and to extract the corresponding inlet flow condition at the aortic level, respectively [3]. The aTAA ex-vivo specimen was harvested after surgical replacement of the ascending aorta. A biaxial traction setup was adopted to obtain the complete mechanical characterization of the tissue and to fit a fiber-based hyperelastic and anisotropic constitutive model [4]. The computational framework for FSI-SOL was implemented in ANSYS. The SOL procedure was divided in three main steps: (i) definition of 0-pressure configuration, (ii) large deformation (LD) and (iii) small deformation (SD) simulation step.

0-pressure configuration - The 0-pressure state was obtained from the diastolic configuration by implementing a reverse displacement algorithm [5]. Firstly, the diastolic geometry is pressurized with a physiological load, then, the resulting displacement is reversed-weighted and re-applied to the starting geometry to obtain a candidate. The candidate is then pressurized again to obtain a score by imposing a comparison with the diastolic configuration. The process is repeated by modifying the reverse displacement weight until the score reaches a minimum.

LD simulation step - Since the deformation occurring from the unloaded to the diastolic configuration is assumed to be large, the linearization assumption cannot be made in this step. The 0-pressure configuration is loaded with 80 mmHg through a static mechanical simulation step. The material was modeled according to the fitted constitutive model from the experimental data.

SD simulation step - Since the deformation occurring from the diastolic to the systolic step is small, the linearization is allowed. At this step, the fully-coupled FSI-SOL simulation is setup: the linearization is imposed, according to the SOL theory, on the basis of the LD step results. In particular, the Cauchy stress is expressed to contain a pre-stress and a linear elastic component according to:

$$t_{ij} = t^0_{ij} + \mathbb{C}^0_{ijkl}\epsilon_{kl} \tag{1}$$

where t^0_{ij} is the pre-stress, ϵ_{kl} is the deformation and $\mathbb{C}^0_{ijkl}$ is the tangent stiffness tensor component deriving from the linearization of the LD step results. The linearized tangent stiffness is defined according to:

$$\mathbb{C}^0_{ijkl} = \delta_{ik}t^0_{ij} + t^0_{il}\delta_{jk} + 4\, F^0_{iI}F^0_{jJ}F^0_{kK}F^0_{lL}\frac{\delta W(\mathbf{C})}{\delta C_{IJ}\delta C_{KL}}\bigg|_{\mathbf{C}=\mathbf{C}^0} \tag{2}$$

where $\delta_{ab} = 1 \iff a = b$ (0 otherwise), F^0 and C^0 are the deformation gradient and the right Cauchy-Green deformation tensor from the LD step respectively. The W function is the strain energy density function defined according to the adopted fiber-based constitutive model. The tangential stiffness tensor $\mathbb{C}^0$ was modified at each mesh element to maintain the local tissue anisotropy. Concerning the fluid-dynamic part, the patient specific inlet flow condition was imposed. The outlets were managed according to three-element Windkessel models, imposed at the supra aortic vessels and descending aorta levels. The simulation results are presented in terms of linearized stiffness distribution both along circumferential ($\mathbb{C}^0_{\theta\theta\theta\theta}$) and longitudinal ($\mathbb{C}^0_{zzzz}$) direction. Additionally, the resulting stresses and the fluid dynamic features from the FSI-SOL are reported.

3 RESULTS

The distribution maps of the linearized tangent stiffness and the peak stress at systole are reported in Figure 1 a-b and Figure 1 c-d, respectively.

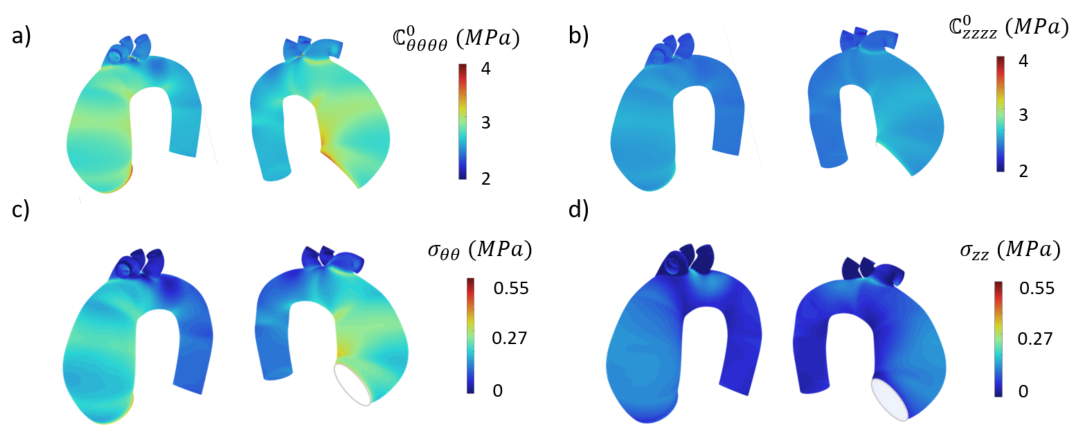

Figure 1: Linearized tangent stiffness distribution for the circumferential (a) and longitudinal (b) direction. Peak stress maps at systole for for the circumferential (c) and longitudinal (d) direction.

Figure 2 depicts the stress distribution range and the velocity streamlines resulting from the SD step. The simulation conditions were revealed to maintain the physiological pressure range.

4 DISCUSSION

The reported results revealed the effect of anisotropy. The resulting stiffness values exhibited a significant heterogeneity, with a stiffer behavior within the internal curvature of the aTAA and along the circumferential direction. An analogous trend was revealed from the stress maps. These features suggest the importance of anisotropy modeling and the effect of the geometry on the mechanical features.
The results reveal that, if a homogeneous hyperelastic material model is assumed, significant heterogeneities of tangent stiffness are still expected. The computational workflow

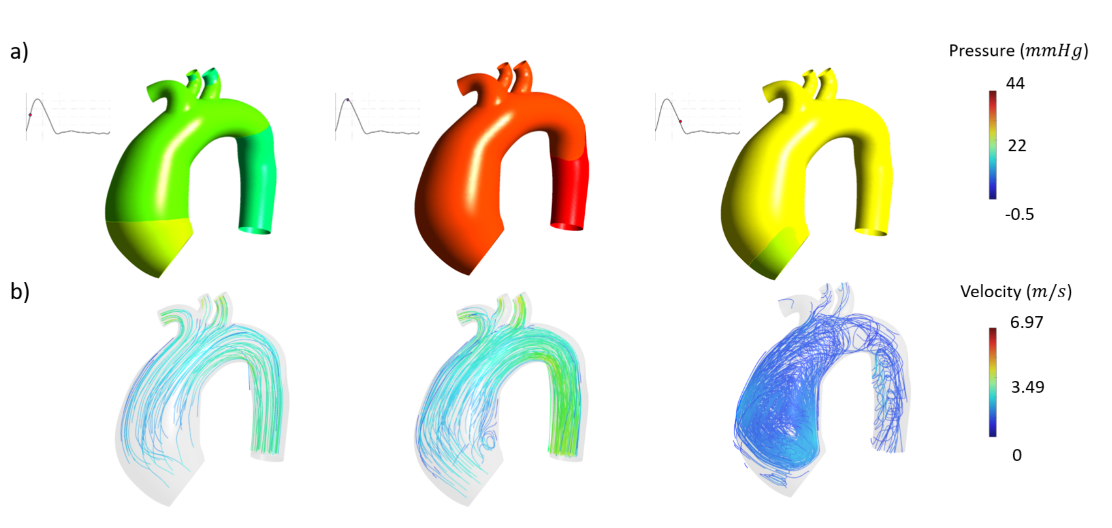

Figure 2: Pressure (a) and velocity (b) distributions at different cardiac phases from the fully-coupled FSI step.

for the SOL was demonstrated to be feasible with the FSI analysis of a patient-specific aTAA case.

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