A FLUID-STRUCTURE INTERACTION MODEL OF THE AORTIC VALVE INCORPORATING CORONARY FLOW

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INTRODUCTION

The aortic valve has received much attention in the biomedical research community, and for good reason. As a non-linear anisotropic material, the cardiac tissue is an ideal place to showcase purely structural simulations. Additionally, it has been shown that the opening dynamics of the passive aortic valve are driven exclusively by the blood flow – making this region an especially good candidate for fluid-structure interaction (FSI) models.

Regardless of the approach, the aim of all these studies is the same: to understand the intricacies of the healthy aortic valve and, perhaps, to mimic it in the realm of prosthetics. The structure is subjected to loading some 2.2 x 10^9 times during the average lifespan and the natural valve's ability to withstand this loading is far superior to either mechanical or biological surgical reconstructions. Mechanical solutions offer good long-term results, but sacrifice natural physiological dynamics and require anticoagulation medication. Biological valves improve upon mechanical valves in terms of reproducing the natural dynamics, but suffer in the area of durability

Previous studies on this topic have varied vastly, both in terms of their approach and the presentation of their results. Some authors have modeled only the leaflets, ignoring the possible effects of the sinus geometry.[1] Other studies have demonstrated a more robust geometric model, incorporating the sinuses and various material properties.[2,3,4,5] Some common metrics have been the leaflet stress pattern, the valve orifice area versus time, and the opening pattern. The goal of this particular study was to expand on these methods. A model was created and tested that included the leaflets, sinuses, and coronary arteries. In addition to examining the opening dynamics, we were now free to examine how coronary flow may be affected by various parameters.

While stress patterns, orifice areas, and closing dynamics are all crucially relevant parameters in prosthetic design, it is our belief that the addition of coronary flow will vastly improve current FSI models. With this model it may be possible to pinpoint the cause of ischemia in operated valves that otherwise appear to be functioning normally.

METHODS

An idealized, parametric, geometric model of the aortic valve was created in Ansys and dimensions for these parameters were taken as the average of various reported values. The model is shown in figure 1, in an exploded view along the axial direction. Three components comprise the solid domain: the aortic root, the leaflets, and the sinuses. Additionally, the sinus component includes a portion of the ascending aorta and the two coronary ostia. The fluid domain consists of three components: an inlet, outlet, and central region.

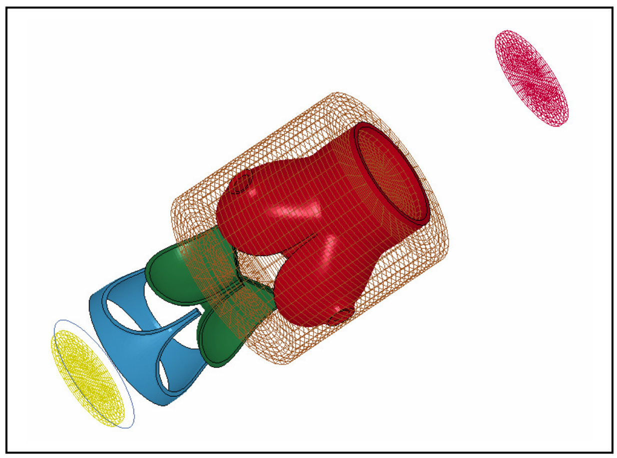

Figure 1: An exploded view of the aortic valve model. The wire-framed objects are components in the fluid domain. The other parts comprise the solid domain.

The model was meshed in Ansys, discretizing the solid components into 10,892 shell elements. The fluid domain consisted of 24,500 solid elements. An isotropic linear elastic material property was used for the shell elements. While other, more complex, properties have been used with success in the past this model was selected for the sake of computation time. Considering this limiting assumption, most of the results will be drawn from the fluid domain, as values for stress in the shell elements may be quite far from their actual physiological values.

A time dependent, spatially uniform, pressure condition was applied to the fluid domain to impel the valve dynamics. The magnitude of this curve was calculated as the pressure in the left ventricle less the aortic pressure over the cardiac cycle. This is in line with other studies.[6] Furthermore, a time-dependent pressure condition was applied to the coronary ostia. The magnitude of the boundary condition was taken from the literature.[7]

The model was constrained in all 6 DOF at the level of the ascending aorta and the coronary outlets. Constraints at the inlet were limited to the axial direction, allowing expansion of the aortic root.

RESULTS

Numerical simulations, particularly those of the FSI variety, produce a daunting amount of raw data. Information on just about any conceivable facet of the object is available and, for this reason, it is of extreme importance to examine the results in a methodical and logical fashion. In the following section two types of results will be shown. First, the closing morphologies and bulk flow properties will be presented and compared to known physiological values. This will, hopefully, inspire some confidence in the simulation for the second set of data: values of the flow in the coronary arteries. Previous simulations have ignored this feature. It is our hope that these results will demonstrate the feasibility and importance of their inclusion.

Global valve dynamics

The leaflet morphologies during the opening and closing phases, as shown in figure 2, compare favorably with previous research. Similar time-lapse images presented by other authors have demonstrated the same triangular orifice area and leaflet billowing.[2,3,5,8] Additionally, the timing of the opening and closing phases are within the acceptable limits.[4] Speaking in strictly qualitative terms, the leaflet displacement results are encouraging.

For a more quantitative analysis of the global functionality of the model, velocity history was sampled at the center node at the level of the commissures. A peak velocity of 144 cm/s occurred at 0.085 s. This is in agreement with previous studies, which have reported peak velocities of 1.35 ± .35 m/s.[9] The fluid also exhibits a rapid acceleration followed by a deceleration slightly smaller in magnitude – this is also known to be the case physiologically. The time history of this velocity is shown in figure 3.

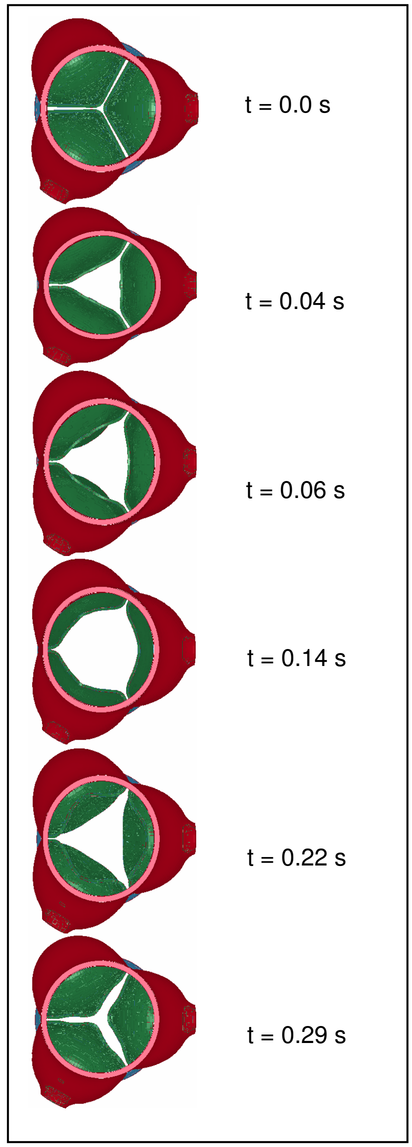

Figure 2: The opening and closing patterns of the leaflets, as seen from the aorta, during certain moments of the cardiac cycle.

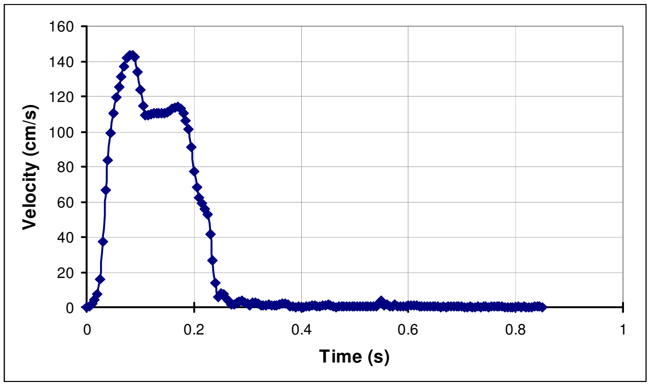

Figure 3: Velocity values, sampled over the cardiac cycle, at the central nodes at the level of the commisures.

As a final measure of the global valve dynamics, the maximum orifice area was calculated and compared to known values. Figure 4 shows the morphology of the valve at the time of maximum orifice area, as well as the transverse plane used to calculate this value. In agreement with previous studies, the valve opened to 69% of the cross section of the valve at the same level.[4]

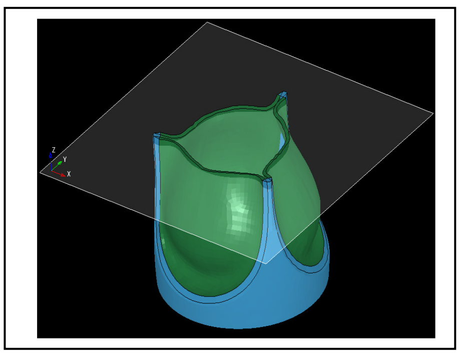

Figure 4: The displacements of the leaflets at the point of maximum orifice area and the transverse plane used to determine this area.

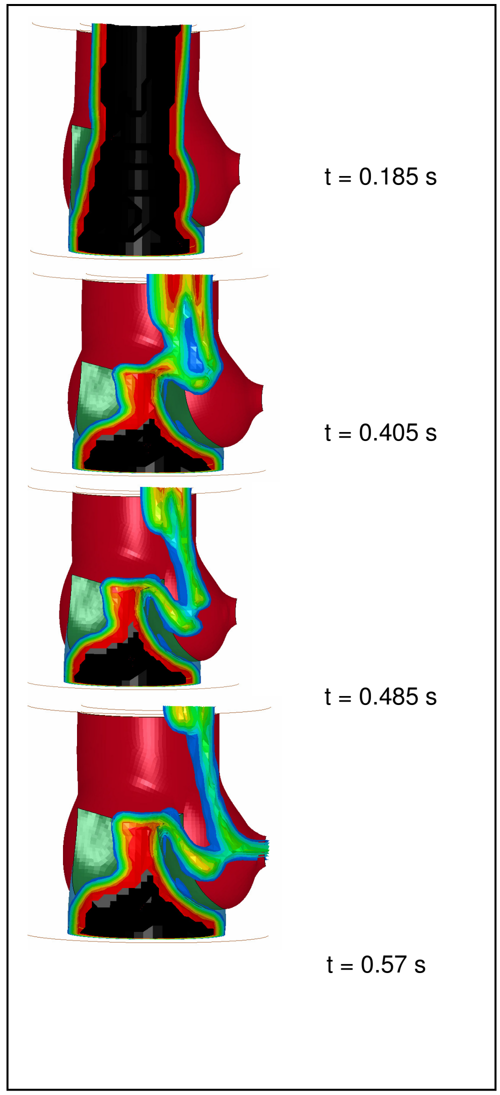

Figure 5: A plot of the volume fraction in LS-Prepost demonstrates the nature of the fluid dynamics in the sinus over the cardiac cycle.

Coronary flow data

In figure 5 the flow pattern in a sinus is depicted at four instances. All four images are shown from the same view – the model sliced in half along the axial direction. In the first image, at 1.85 seconds into the cardiac cycle, the valve is fully open the fluid is flowing entirely into the ascending aorta. The second image demonstrates the backflow induced by the phase of deceleration. The low momentum fluid near the wall is affected more greatly and the blood begins to tend into the sinus. The third image shows the familiar formation of a vortex in the sinus.[10] Finally, the blood moves from the sinus out through the coronary artery.

With a global idea of the flow patterns driving the coronaries, it would now be useful to examine the structure in a more quantitative mode. Figure 6 shows the peak velocity of the blood through each of the coronary arteries. These are very close to the known physiological values for coronary flow with an overall coronary flow of 200 ml/min, or, approximately 4% of the cardiac output. Additionally, the flow exhibited an impressively smooth parabolic shape.

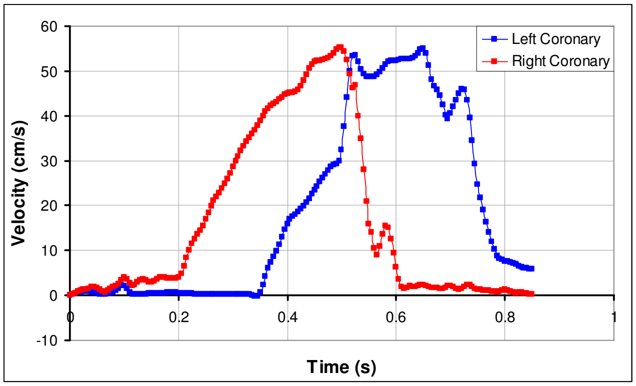

Figure 6: The velocities in the coronary arteries over the cardiac cycle.

CONCLUSION AND LIMITATIONS

With this model of an aortic valve in healthy working order, the next logical step would be to explore pathologies and surgical procedures. The variations on the coronary flow to these alterations will provide a new metric by which to measure surgical options.

While this model is in very good agreement with previous research, and while the coronary flows are a good match with known values, there are still a number of serious limitations to our approach. First, the material properties of the solid components are linear elastic and isotropic. Future work will incorporate more complex material models to determine if, or to what effect, this change may have on the coronary flow model. Secondly, for the sake of computational time, the model has been mass scaled by a factor of ten. This method is not terribly uncommon, but, once again, it is, strictly speaking, anatomically incorrect. Finally, it is important to stress the idea that this idealized realization of the aortic valve can only serve as a broad tool to aid our understanding of the mechanics of the object. Such a general geometry, as it stands, has very little use to any patient specific case.

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