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A computational study of the three-dimensional fluid–structure interaction of aortic valve

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ABSTRACT

In this paper, we describe a three-dimensional simulation of the fluid–structure interaction (FSI) of the aortic valve using a direct-forcing immersed-boundary method. The geometry of the valve is taken from a bioprosthetic valve, and the computational framework is based on a previous partitioned approach that is versatile for handling a range of biological FSI problems involving large deformations. When applying the approach in the heart valve simulation, we implemented an efficient parallel algorithm based on domain decomposition to handle the costly flow simulation. As compared with previous simulations of the aortic valve, our simulation was able to capture both realistic deformation of the leaflets and vortex structures in the flow, thus providing a balanced modeling approach for the flow and the valve. The results show that the pressure distribution on the leaflet surface is highly nonuniform and the jet flow contains a sequence of vortices during the opening process. After the valve is fully opened, both the three leaflets and the jet still experience significant oscillations. The drag resistance of the valve is also characterized, and it is found that the resistance is approximately equivalent to the inertial force of accelerating the fluid column of three diameter length. These details could be potentially used to characterize FSI of the aortic valve.

1. Introduction

Computational modeling of fluid–structure interaction (FSI) between the flexible leaflets of heart valves (e.g., the mitral valve and aortic valve) and blood flow has many potential applications in diagnosis, surgical repair and replacement, and design of prosthetic valves (Yoganathan et al., 2004; Sun et al., 2014). However, due to substantial simulation challenges involved in handling the large three-dimensional (3D) geometrical variations, topological change of the flow domain (i.e., on and off of the valves), and numerical instability of the FSI algorithm, to date few methods can solve this FSI problem to a satisfactory level. It remains an active area of research to develop computationally efficient and yet high-fidelity simulation tools to model the FSI of heart valves.

Given the complex 3D geometry of heart valves and large displacement of the valve leaflets in a cardiac cycle, it is extremely difficult to apply a conventional computational fluid dynamics (CFD) method that is based on boundary-conformal meshing of the fluid domain, as frequent mesh regeneration would be needed to avoid severe mesh distortion and deterioration. Therefore, the existing numerical methods for the heart valve FSI have mostly relied on non-boundary-conformal or immersed-boundary type of approaches to solve the fluid flow. In these methods, the mesh discretizing the flow domain is a typically fixed grid, either structured (e.g., Cartesian) or unstructured, and is independent from the Lagrangian mesh that discretizes and tracks the solid domain representing the elastic leaflets. As the Lagrangian mesh moves across the

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fixed grid, special treatment needs to be done in the flow solver to accounts for the presence of the immersed leaflets and the effects of their movement.

To do so, De Hart et al. (2003) introduced a Lagrange multiplier into the governing Navier–Stokes equation to replace the surface force at the boundary for simulation of the aortic valve. Even though the flow simulation was limited to a coarse resolution of less than 1000 finite elements, this work nevertheless represents one of the early studies of full 3D FSI models of the heart valves. Later Griffith et al. (2009) and Griffith (2012) advanced the aortic valve model by using a diffuse-interface immersed-boundary method that is based on the Cartesian grid for the flow. The use of a structured grid, and thus the efficient algorithms associated with the grid, allows the method to deploy much more mesh points to resolve the flow. In the diffuse-interface immersed-boundary method, the no-slip/no-penetration boundary conditions are not directly imposed when solving the Navier–Stokes equation; rather, in one computational step, the solid structure is first allowed to convect along with the fluid, and then the required stress from the fluid leading to the structural deformation is computed by solving the solid mechanics; this fluid force is then fed into the momentum equation of the fluid as a regularized volumetric force to advance the time step. Conceptually, this approach only changes the sequence of the solution process and would still provide the same solution once convergence is reached. However, such sequencing could be prone to spurious fluid forces and numerical instability, especially when the stiffness of the structure is high and the computed fluid stress is sensitive to the structural deformation. In those simulations, the valve leaflets have low bending stiffness and thus develop unnatural wrinkle deformations.

Borazjani (2013) used a sharp-interface immersed-boundary method to model the flow for both a mechanical and a bio-prosthetic aortic valve. The method in this work was also based on a fixed but curvilinear grid, and fine resolution was applied to capture the detailed vortex structures in the flow. However, the structural model in the work utilizes membrane finite elements, which do not include the bending stress in the structure; thus, the leaflets also develop unnatural wrinkly deformations. Later, this method was extended to include shell elements to handle bending of the leaflets (Gilmanov et al., 2015; Gilmanov and Sotiropoulos, 2016). The closing phase was not considered in those studies. Recently, Kamensky et al. (2015) and Hsu et al. (2015) used a so-called immersogeometric FSI method to model both the flow and structure. Several advancements have been made in their study. For example, the mesh for the flow is not boundary-fitted but is adaptive around the solid surface; both stretching (in-plane) and bending (out-of-plane) deformation were included; a contact model was introduced for leaflet collision; and wall compliance was also incorporated. One limitation is that their study has not yet examined the flow field in detail. More recently, Mao et al. (2016) combined a commercial finite-element package with a custom flow solver based on the smoothed particle hydrodynamics to simulate FSI of the aortic valve. With the help of the commercial package, their structural model is detailed and displays realistic pattern of deformation. However, their discussion on the flow field is also very limited.

To summarize this brief literature review, the FSI model of the heart valves has been advanced significantly in recent years, but there is still lack of modeling study that provides reasonable details for both the valve deformation and the flow pattern. To address this issue, we propose a 3D FSI simulation of the aortic valve to investigate both the leaflet deformation and vortex pattern in the flow. Studies like this are significant because it is understandable that for systems like the heart valves, the flow pattern is intricately associated with the structural deformation through the interaction, and understanding the relationship between the two parts may lead to useful tools to diagnose any abnormality of one part based on available information about the other.

In our study, we will use a Cartesian grid based sharp-interface immersed-boundary method combined with a finite-element method for the solid (Tian et al., 2014). This computational framework was previously developed to simulate 3D biological FSI problems with large deformations. It has the capability of handling both thin-walled structures and general 3D bodies, and it incorporates both geometric and material nonlinearities. In the present work, we will describe its application in the aortic valve FSI and will discuss the results for both the valve and the flow.

2. Model description and the numerical approach

2.1. Model setup

The three-dimensional computational model used in this paper is illustrated in Fig. 1. The aorta is simplified to a cylindrical tube of diameter $D = 2.1$ cm and length $L = 19$ cm. It has a three-lobed dilation to model the aortic sinuses. These dilation regions were believed to play a significant role in the dynamics of the valve (Bellhouse et al., 1968; Leyh et al., 1999; Salica et al., 2015). The dimensions of the aortic sinuses in the current model are based on the measurements of the human aortic root (Swanson and Clark, 1974; Reul et al., 1990). According to the geometric description in Reul et al. (1990), the cross-section of the aortic sinuses is built with an epitrochoid equation. The aortic valve consists of three flexible semi-lunar leaflets that can independently deform from one another. The overall geometry of the aortic sinuses and leaflets is axisymmetric about the x-axis. Despite of the complexity of anatomy of human aorta, simplified computational domains similar to ours are often used for the FSI study of native aortic valve and its prostheses (Griffith, 2012; Borazjani, 2013; Marom et al., 2013; Hsu et al., 2015; Kamensky et al., 2015; Mao et al., 2016). In our study, the geometry of the leaflets (Fig. 1(b)) was provided by Prof. Wei Sun's lab at Georgia Tech by courtesy and was extracted from a transcatheter aortic valve (TAV) model (Mao et al., 2016).

For spatial discretization, the aorta wall is divided into 20,735 triangular elements with refinement in the sinuses region, where the element size is about 0.3 mm. Each leaflet is 0.1 mm thick and consists of a total of 535 finite-element serendipity