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# Image-based immersed boundary model of the aortic root

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ABSTRACT

Each year, approximately 300,000 heart valve repair or replacement procedures are performed worldwide, including approximately 70,000 aortic valve replacement surgeries in the United States alone. Computational platforms for simulating cardiovascular devices such as prosthetic heart valves promise to improve device design and assist in treatment planning, including patient-specific device selection. This paper describes progress in constructing anatomically and physiologically realistic immersed boundary (IB) models of the dynamics of the aortic root and ascending aorta. This work builds on earlier IB models of fluid–structure interaction (FSI) in the aortic root, which previously achieved realistic hemodynamics over multiple cardiac cycles, but which also were limited to simplified aortic geometries and idealized descriptions of the biomechanics of the aortic valve cusps. By contrast, the model described herein uses an anatomical geometry reconstructed from patient-specific computed tomography angiography (CTA) data, and employs a description of the elasticity of the aortic valve leaflets based on a fiber-reinforced constitutive model fit to experimental tensile test data. The resulting model generates physiological pressures in both systole and diastole, and yields realistic cardiac output and stroke volume at physiological Reynolds numbers. Contact between the valve leaflets during diastole is handled automatically by the IB method, yielding a fully competent valve model that supports a physiological diastolic pressure load without regurgitation. Numerical tests show that the model is able to resolve the leaflet biomechanics in diastole and early systole at practical grid spacings. The model is also used to examine differences in the mechanics and fluid dynamics yielded by fresh valve leaflets and glutaraldehyde-fixed leaflets similar to those used in bioprosthetic heart valves. Although there are large differences in the leaflet deformations during diastole, the differences in the open configurations of the valve models are relatively small, and nearly identical hemodynamics are obtained in all cases considered.



## 1. Introduction

Worldwide, 300,000 heart valve repair or replacement procedures are performed each year [1–3], and this rate is projected to increase to 850,000/year by 2050 [2]. Treatment for severe stenosis of the aortic heart valve is generally to replace the native valve with either a mechanical or a bioprosthetic valve [4], and in the United States alone, approximately 70,000 aortic valve replacements are performed every year [5].

Surgical valve replacement via open-heart surgery has been performed since 1960, but over the past decade, transcatheter aortic valve replacement (TAVR) has emerged as a less invasive alternative to conventional valve replacement surgery [6,7]. In TAVR, a stent-mounted bioprosthetic heart valve is percutaneously implanted via a catheter within the diseased valve. First-in-man TAVR implantation was performed in 2002 [8]. In 2011, the Edwards *SAPIEN* valve became the first TAVR device to be approved by the U.S. Food and Drug Administration, and approval of a second TAVR device, the Medtronic *CoreValve*, followed in 2014. TAVR is now approved in the U.S. for use in both high- and intermediate-risk

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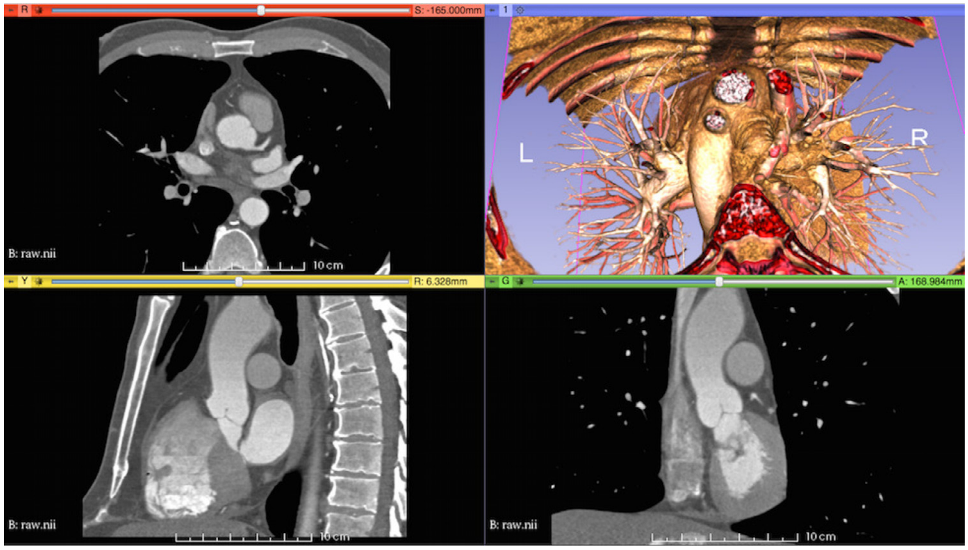


**Fig. 1.** The computed tomography (CT) image data used in this study.

patients, and is poised to become an increasingly common alternative to conventional surgical valve replacement.

TAVR device selection, including device sizing, can be challenging. For instance, leakage flows between the stented valve and the aortic root, which are referred to as paravalvular leaks, are clearly linked to increased long-term mortality [9–11], and interactions between the device and calcification lesions within the aortic root can determine the degree of residual paravalvular leak following TAVR [10]. Improved methods for device selection could also reduce the occurrence of severe complications such as heart block [12,13]. Computer simulation promises to facilitate device selection and treatment planning for patients who require these devices.

Models also can provide insight into physiological mechanisms, ultimately facilitating improved device design, and can accelerate the testing of implantable medical devices such as prosthetic heart valves. It is known, for instance, that many of the difficulties of prosthetic heart valves are caused by the fluid dynamics of the replacement valve [1,2]. Computational models of cardiac dynamics can predict the flow patterns of native valves as well as mechanical and tissue valve prostheses. Simulations also can predict the kinematics and loads experienced by the valve leaflets, which could lead to novel designs that improve device durability. Indeed, the major limitation of bioprosthetic heart valves is their limited durable lifetime, which is typically 10–15 years.

Fluid–structure interaction (FSI) models can predict the dynamics of the aortic valve leaflets [14–21], and recent work provides more complete descriptions of the dynamic FSI-mediated coupling between the valve leaflets and the sinuses [22–24]. For instance, to account for patient-specific geometry, a finite element-based fluid–structure interaction model of the aortic root was constructed using magnetic resonance imaging that accounts for geometric variability among a population of 10 patients [25]. In this work, an emphasis is placed on correctly modeling the nonlinearities associated with the mechanical response of the valve leaflets. In another study, data from transesophageal echocardiography were used to construct aorta models, integrating tissue properties derived from the patient's age [26]. The model is constructed by determining anatomical landmarks for leaflet and vessel response are further tailored to the patient through the use of age dependent material properties. The efficacy of the model is assessed by comparison with patient imaging data. Recently, a study using an arbitrary Lagrangian–Eulerian (ALE) approach compared the performance of native valves as well as stentless and stented bioprosthetic valves in realistic patient anatomies [24].

This paper describes ongoing work to develop a simulation framework based on the immersed boundary (IB) method [27] to model the dynamics of the aortic root, with the goals of facilitating prosthetic valve design and personalized approaches to treatment planning. We previously developed three-dimensional FSI models of the aortic root using several different versions of the IB method, including a cell-centered IB method based on an efficient approximate projection method [28,29], a staggered-grid IB method with improved volume conservation [30,31], and an IB method with support for finite element elasticity models [32,33]. These initial models used realistic driving and loading conditions, including realistic diastolic pressure loads on the closed valve, and produced physiological stroke volume and cardiac output at realistic Reynolds numbers. However, all of these studies employed a highly stylized aortic geometry derived from patient image data [34] that did not account for the asymmetry of the aortic root or the curvature of the ascending aorta. In addition, these earlier studies all used simplified descriptions of the mechanics of the aortic valve cusps based on idealized fiber-based models that were discretized using systems of springs and beams.

This paper extends these earlier IB models of aortic valve dynamics [29,30,32] toward clinical utility by incorporating a realistic, three-dimensional anatomical model of the aortic root and ascending aorta. Within this image-based anatomical geometry, we construct a rule-based model of the fiber structure of the aortic valve leaflets using an approach based on Poisson interpolation that is similar to methods used previously to generate models of the fiber architecture of the heart [35–37]. The biomechanics of the valve leaflets is described by a fiber-reinforced hyperelastic constitutive model for the aortic valve leaflets [38] with constitutive parameters fit to experimental tensile test data from fresh or glutaraldehyde-fixed porcine aortic valve leaflets [39,40]. Three-dimensional FSI simulations are performed by an IB method that supports finite-strain continuum mechanics structural models discretized via a nodal displacement-based finite element scheme [33]. These methods are well suited for complex anatomical geometries, experimentally based constitutive models, and large-scale simulation. The model is driven using clinical hemodynamic data [41], and afterload is modeled by a three-element Windkessel model fit to the same data [42]. The FSI model is demonstrated