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Assessment of calcified aortic valve leaflet deformations and blood flow dynamics using fluid-structure interaction modeling

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ABSTRACT

Aortic valve diseases are among the most common cardiovascular defects. Since a non-functioning valve results in disturbed blood flow conditions, the diagnosis of such defects is based on identification of stenosis via echocardiography. Calculation of disease parameters such as valve orifice area or transvalvular pressure gradient using echocardiography is associated with substantial errors. Computational fluid dynamics (CFD) modeling has emerged as an alternative approach for accurate assessment of aortic valve hemodynamics. Fluid-structure interaction (FSI) modeling is adapted in these models to account for counter-interacting forces of flowing blood and deforming leaflets for most accurate results. However, implementation of this approach is difficult using custom built codes and algorithms. In this paper, we present an FSI modeling methodology for aortic valve hemodynamics using a commercial modeling software, ANSYS. We simulated the problem using fluid flow solver FLUENT and structural solver MECHANICAL APDL under ANSYS and coupled the solutions using System Coupling Module to enable FSI. This approach minimized adaptation problems that would raise if separate solvers were used. As an example case, we investigated influence of leaflet calcification on hemodynamic stresses and flow patterns. Model geometries were generated using b-mode echocardiography images of an aortic valve. A Doppler velocity measurement was used as velocity inlet boundary condition in the models. Simulation results were validated by comparing leaflet movements in the simulations with b-mode echo recordings. Wall shear stress levels, pressure levels and flow patterns agree well with previous studies demonstrating the accuracy of our results. Our modeling methodology can be easily adopted by researchers that are familiar with ANSYS and other similar CFD software to investigate similar biomedical problems.

1. Introduction

Aortic valve separates the left ventricle from the aorta. It consists of three half-moon-shaped pocket-like flaps of leaflets housed within three sinuses [1]. A healthy aortic valve fully opens at ventricular systole and fully closes at ventricular diastole ensuring unidirectional flow with minimal regurgitation. The valve functions according to transvalvular pressure difference. Valve leaflets are exposed to complex hemodynamic forces during cardiac cycle: while the front ventricularis surfaces are exposed to unidirectional jet flows at peak systole, back fibrosa surfaces are exposed to circulatory flows while the valve is closing [2–4]. Valve calcification and bicuspid valve are the most common types of aortic valve defects. Incidence rate of congenital bicuspid valve is 0.5–2% [5]. Calcification affects mainly elderly with incidence rate of 2–7% in the population above 65 years of age [6].

Aortic valve defects prevent efficient opening of leaflets and result in aortic stenosis (AS), which is the formation of a high velocity jet at the valve orifice at peak systole. This disturbed flow condition also results in an increase in transvalvular pressure gradient (TPG) which is associated with heart attack risk [7]. Determination of jet velocity and TPG is essential for diagnosis of the condition. Doppler echocardiography is the most commonly used technique for that purpose. Doppler can measure the maximum blood velocity at the valve orifice, but not the axial velocity profile of the jet. The assumption of constant velocity along jet orifice will lead to some potential errors in further calculations for TPG [8]. Effective Orifice Area (EOA) is another parameter used by physicians for the clinical assessment of AS severity. EOA is the minimal cross sectional area of the aortic flow jet. EOA can be calculated from

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continuity equation by relating Doppler measured velocities at aortic valve inlet and at jet orifice [9]. Doppler velocities can be used to calculate TPG using simplified Bernoulli equation, where viscous terms are neglected. Due to simplifying assumptions and inaccuracy of the measurements, calculating EOA and TPG from Doppler velocities were shown to be associated with significant errors [10].

Accurate diagnosis of severity of aortic valve disease is crucial for therapy planning. However, as explained above, current approaches are associated with significant errors in hemodynamics analysis. Computational fluid dynamics (CFD) modeling has emerged as an alternative useful approach for elucidating complex cardiovascular flows where clinical measurement schemes would provide only limited information [11]. Patient-specific aortic valve CFD models are generated using medical images from patients [12]. These models enable a detailed disturbed flow analysis and precise determination of defect severity. Disturbed hemodynamics can trigger mechano-biological mechanisms (i.e. up/down regulation of gene expressions etc.) leading to further complications [13]. Therefore, qualitative and quantitative hemodynamics analysis for defected aortic valves will be important in biological investigation of the disease as well.

There are numerous previous numerical studies on aortic valve hemodynamics [14–16]. The initial studies focused on the investigation of blood flow and determination of hemodynamic force levels on the leaflets at a specific time point in the cardiac cycle (in most cases ventricular systole) using static geometries. In these studies, only the structural domain (aortic root and leaflets) was modeled and fluid flow behavior through the valve was excluded. Grande et al. generated 3D finite element models from MRI images using ANSYS and defined physiological pressure on the leaflets as boundary condition [17]. The study showed that, patient specific asymmetries are important in stress distribution on the aortic root and the leaflets. With the current advancements in simulation techniques and computational speeds, dynamic leaflet movements could be modeled in more recent studies. Weinberg et al. developed 3D aortic valve models using LS-Dyna and defined transient dynamic hydrostatic pressure on the structural zones as the boundary condition in the models [18]. Influence of valve calcification on leaflet movement behavior was investigated in the study. In a more recent investigation on calcification, patient specific 3D models were generated from MRI images of aortic valve patients. Using ABAQUS software, transient pressure boundary condition was applied on the leaflets [19]. The valve orifice area decreased when calcification level increased in the study.

These studies contributed substantially to our understanding of the aortic valve behavior during disease. However, such structural models cannot be used to investigate disturbed hemodynamics, which was shown to be important in the progression of aortic valve disease. Aortic valve behavior is a complex dynamic event since it involves both fluid and structure movements. To model this complex dynamic problem, fluid-structure interaction (FSI) approach should be adapted. In this approach, valve leaflets are modeled as deformable structural domain and blood flow is modeled as fluid domain. Two domains are coupled and mathematical solutions of the fields are determined simultaneously. This approach enables researchers to investigate the dynamics of leaflet movements and blood flow throughout the cardiac cycle in a reliable manner.

In a pioneering FSI study by De Hart et al., aortic valve hemodynamics was modeled in 2D and associated fluid and structure equations were solved simultaneously using custom algorithms developed by the authors [20]. In another study, ADINA commercial software was utilized for generating 2D FSI models to investigate hemodynamic forces on bicuspid and tricuspid aortic valves [21]. Here, transient pressures on the leaflets were used as boundary condition and the solution was converged when the solution approximated physiological values. In more recent studies, aortic valve hemodynamics was simulated in 3D. Halevi et al. investigated the influence of calcification on valve hemodynamics using 3D FSI models [22]. Here, structural model was generated using ABAQUS and flow field was solved using Flow Vision. Fedele et al. developed a patient-specific FSI modeling approach based on moving resistive immersed implicit surfaces [14]. Here, a mathematical model was generated to represent a fluid in a general domain with an immersed structural domain into the fluid domain. FSI approach was adapted to evaluate the performances of aortic valve replacements as well. Yun et al. investigated the blood damage through bi-leaflet mechanical aortic valves [23] and Vahidkhah et al. studied blood stasis on prosthetic aortic valves using such 3D FSI models [24]. These studies were very important FSI modeling examples of the aortic valve. Advanced mathematical modeling approaches adapted in most of these studies make them difficult to be adopted by other researchers.

In this study, we aimed to simulate aortic valve hemodynamics in a practical and reliable way, that would make the models easily adoptable by others. For this purpose, we generated 2D FSI aortic valve models under a single modeling platform, ANSYS. Fluid and structure fields were defined on patient specific models that were prepared in ANSYS Workbench. ANSYS FLUENT and ANSYS MECHANICAL APDL were used to solve fluid and structure fields respectively. Two fields were coupled using System Coupling FSI Module in ANSYS Workbench. Since all the solvers and modules work under one platform, adaptation problems that might raise due to utilization of separate software packages and modules were minimized. The influence of leaflet calcification on flow and leaflet movement dynamics was investigated using this approach. The modeling technique that we developed in this study will be readily adoptable to researchers using ANSYS for heart valve modeling.

2. Methods

In this study, we simulated aortic valve hemodynamics using ANSYS package program. For this purpose, we generated a 2D model geometry based on aortic valve images from a patient. Surfaces and regions in the geometry were defined appropriately to allow FSI. Simultaneous solutions were obtained for fluid and structure zones using separate solvers under ANSYS WORKBENCH. Below, we summarize our solution steps in detail:

2.1. Echocardiography imaging

Aortic valve from a healthy individual at 27 years old was imaged via echocardiography using 7 and 3 MHz scanners (GE Vivid 7 Ultrasound Machine). Dimensions of the valve (i.e. inlet, orifice, sinus and root diameters) were measured from short axis b-mode images (Fig. 1A). Time dependent velocity profile at the aortic valve inlet was measured via Pulsed Doppler Mode (Fig. 1B, left). These velocity measurements were used as transient velocity boundary condition for the simulation (Fig. 1B, right).

2.2. Numerical model

The geometry of the model was developed in ANSYS Workbench based on b-mode measurement data (Fig. 1C). The numerical model is divided to two separate fields: The flow field and the Structure Field.

The flow field consists of blood inlet surface, blood outlet surface and the blood flow region. Inlet surface of the geometry (left surface in Fig. 1C) was defined as velocity inlet boundary condition. The velocity profile for aortic valve inlet was measured using Doppler technique and applied as the transient velocity boundary condition on this surface. Flow at the inlet was assumed plug flow [25]. At a specific time-point, inlet velocity is the spatially averaged velocity of the blood flow entering to the aortic valve from left ventricle. This average value equals to the maximum axial velocity measured by Doppler at valve inlet (plug flow assumption). The transient velocity profile was defined as an UDF file to be interpreted by FLUENT solver. This file includes the transient velocity data for inlet surface for each time step. The outlet surface (right surface in Fig. 1C) was defined as pressure outlet and the outlet pressure was set