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FSI Simulation of Intra-ventricular Flow in Patient-specific Ventricular Model with both Mitral and Aortic Valves*

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Abstract—**Investigating the intra-ventricular flow is the most important to understand the left ventricular function. In this study, we proposed a fluid-structure interaction (FSI) approach to simulate the blood flow in patient-specific model by combining both mitral and aortic valves. To accommodate the large mesh deformation, moving arbitrary Lagrangian-Eulerian (ALE) meshes were used for moving ventricular wall and rotating leaflets of valves. The left ventricular wall was predescribed according to the points acquired from magnetic resonance image (MRI). Mitral and aortic valves were integrated into the model by assuming each leaflet as a rigid body. Fluid-structure interaction (FSI) approach was adopted to capture the rapid motion of leaflets. The simulation results were qualitatively similar to the measurements reported in literatures. To the best of our knowledge, this is the first to simulate the patient-specific ventricular flow with the presence of both mitral and aortic valves.**

I. INTRODUCTION

Left ventricle is an important chamber of heart, which pumps blood through aortic valve. Intra-ventricular flow has been extensively investigated, as it is critical for understanding the left ventricular (LV) function. Mitral and aortic valves play an import role in keeping the unidirectional flow. It has been shown that the pressure gradient determines the opening and closing of valves [1]. The large diastolic vortex within the left ventricle has been reported to preserve the momentum of flow, i.e. minimize the energy consuming deceleration and reacceleration of blood, and redirects the flow direction towards the aortic valve [2].

To obtain the time-dependent intra-ventricular flow patterns, various techniques have been developed. Besides the success of the non-invasive blood flow imaging techniques, eg. phase-contrast magnetic resonance imaging (MRI), echo particle image velocimetry (E-PIV), computational fluid dynamics (CFD) simulation has played an important role in studying the complex ventricular hemodynamics [3]. There are two types of CFD methods, namely geometry-prescribed and fluid-structure interaction (FSI). The former one utilizes either patient-specific data acquired from clinical imaging or predefined equation to control the boundary motion [4]. The latter method couples the interaction between blood and tissue [5-7]. The reported numerical investigations on the intra-ventricular flow are mostly focused on the early diastolic phase, during which the critical feature of intra-ventricular hemodynamics (i.e. the mitral vortex) forms and develops [6]. Due to its complexity, most of the numerical studies do not include mitral valve. Although mitral valve is included in the numerical study of [7], the aortic valve is excluded in their model.

In this study, we simulated the blood flow in a patient-specific ventricle with both mitral and aortic valves. The ventricular wall was predescribed according to the data of MRI. The motions of both mitral and aortic valves were simulated using FSI method based on Arbitrary Lagrangian-Eulerian (ALE) approach. As the two dimensional simulations are capable of capturing the main flow characteristics [6], the patient-specific left ventricular and both mitral and aortic valves were modeled in 2D. To the best of our knowledge, this is the first to simulate the patient-specific ventricular flow with the presence of both mitral and aortic valves.

II. METHODS

A. Geometrical Model

The anatomical geometry of the left ventricle of a healthy adult was examined by a 1.5 T MR scanner (Avanto, Simens Medical Solution, Erlangen). The black dots in Fig. 1 show original points along the ventricle wall acquired from MRI. More points were added through cubic-spline interpolation to enhance the spatial resolution and were further used for the grid generation. As this study was focused on the intra-ventricular flow, the atrium was simplified as a straight tube. An ideal aortic sinus [8] was applied as shown in Fig. 1. In order to minimize the influence of uniform velocity boundary conditions on the intra-ventricular flow, the inlet was located 30 mm away from mitral valves. As the leaflets of valves were too thin to be clearly extracted from MRI, the leaflets were represented as rigid plates with round tip. The thickness of a mitral valve was assumed to be 1 mm, and the ratio of posterior to anterior leaflet was 0.613 [6]. For the aortic valve, two leaflets with equal length were modeled with the thickness of 0.5 mm. To keep the flow domain continuous,

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a 0.5 mm gap between a pair of leaflets was preserved at the fully closed position. The leaflets are connected to left ventricle through annulus, to enlarge maximum opening angle of leaflets. Both posterior and anterior annuluses were 2 mm in length so that each leaflet can open up to 120°.

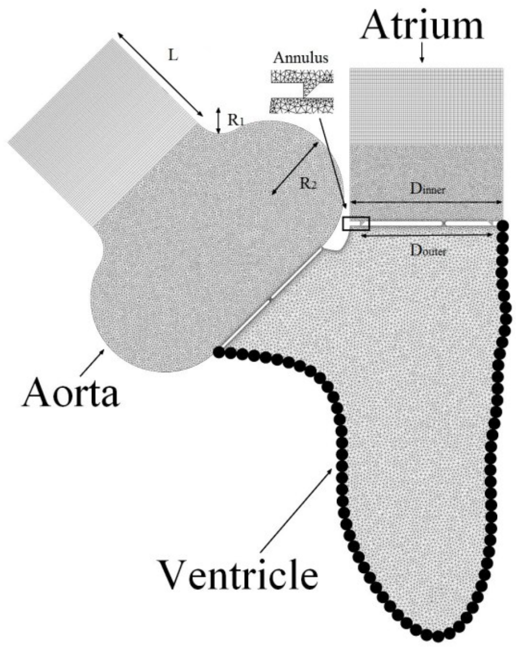

Figure 1. Ventricular contours acquired from MRI and initial grid generation. The outer and inner diameters of the mitral annulus are 26 and 30mm, respectively. The diameter of aortic valve is 30 mm. $L = R_2 = D_A = 30\ mm$ and $R_1 = 3/8 D_A = 11.25\ mm$

B. Numerical Method

In this study, the intra-ventricular flow was simulated by a commercial CFD solver, ANSYS™ FLUENT™ (Version 14.0). It utilizes the finite volume method to solve the continuity and Navier-Stokes equations as shown in (1) and (2):

$$\frac{\partial}{\partial t}\int_V \rho dV + \int_S \rho\left(\vec{v} - \vec{v}_b\right)\cdot\vec{n} dS = 0 \quad (1)$$

$$\frac{\partial}{\partial t}\int_V \rho\vec{v} dV + \int_S \left(\rho\vec{v}\left(\vec{v} - \vec{v}_b\right) + pI - \vec{\tau}\right)\cdot\vec{n} dS = 0 \quad (2)$$

where ρ is the fluid density; $\vec{v}$ is the velocity vector of fluid; $\vec{v}_b$ is the velocity vector of the moving boundary; $\vec{n}$ is the outwardly directed vector normal to dS; S is the boundary of the control volume (V); p is the pressure; I is the unit tensor; and $\vec{\tau}$ is the viscous stress tensor. In this study, the blood was assumed to be incompressible and Newtonian, with constant density of 1,050 kg/m^3 and dynamic viscosity, μ, of 3.5 *cP*.

To take the movement of ventricular wall into considerations, the ventricular wall was pre-described according to the data acquired from MRI measurement. Cubic-spline interpolation was used to enhance the temporal resolution from the original 25 frames to 10,000 frames per cardiac cycle and the spatial resolution of the ventricular wall.

Each leaflet was assumed as a rigid body in rotation around the annuluses, and the motion of the leaflet was solved by the backward Euler first-order implicit scheme implemented in the dynamic mesh model of FLUENT™. To ensure the momentum equilibrium, the partitioned approach coupled the interaction by updating both flow and structure boundary conditions iteratively till the difference of motions predicted by FSI cycle and flow solver was below the threshold. Aitken relaxation method [9] was implemented in this study to update the optimal relaxation factor of FSI sub-iteration in each time step due to its overall simplicity and computational efficiency [10].

In this study, only the systolic and diastolic phases of cardiac cycle were modeled. Velocity waveform was applied to the inlet and the outlet pressure was assumed to be constant. The velocity waveform was derived from the volume variation of the ventricle as shown in Fig. 2(b), and the value was set to zero immediately after the mitral valve was fully closed. No slip boundary condition was applied to the ventricle wall and leaflets. The motion of numerical grids was controlled by user defined function (UDF) at the beginning of each time step. Both spring-based smoothing and re-meshing methods were adopted to handle the large mesh deformation during the simulation. After trial and error, the time step size was set as 0.0002T*. Here T* represents the time for one cardiac cycle. A total of 4.8×10^4 cells were found to be sufficient to provide grid-independent results.

III. Results

A. Opening and Closure of Valves in a Cardiac Cycle

Fig. 2(a) shows the opening angles of both mitral and aortic valves from onset of ventricular dilation to the end of contraction. The opening angles of posterior and anterior leaflets of mitral valve vary in a cardiac cycle and have the similar trend. The difference in magnitude is mainly due to the asymmetric leaflets of the mitral valve. During the early diastole, the mitral valve opens rapidly and it partially closed in the mid-diastole. Before the full closure, the mitral valve reopens when the atrial contracts. Due to the inertia of leaflet, the mitral valve is only partially closed at the end of ventricular diastole ('g'), i.e. maximum ventricle volume shown in Fig. 2(c). It can be seen that the mitral valve fully closed at the time of 'i', while the incoming flow into ventricular has zero velocity at the time of 'g'. During the time instants from 'g' to 'i', blood flows retrogradely into the atrium, which leads to the decrease of left ventricular volume in Fig. 2(c), called the closing volume. The simulation result shows that the closing time is 0.09 T* and the closing volume is 5.4% of the stroke volume. In the systole, the aortic valve opens rapidly without partial closure. Although the leaflets of the aortic valve are assumed to be symmetric in this study, the opening and closing behaviours of these two leaflets are different. This is due to the non-uniform intra-ventricular flow. Similar to the mitral valve, the aortic valve is not fully closed at the end of ventricular contraction. As this study is mainly focused on the flow pattern from the onset of ventricular dilation to the end of ventricular contraction, the aortic regurgitation is not included.

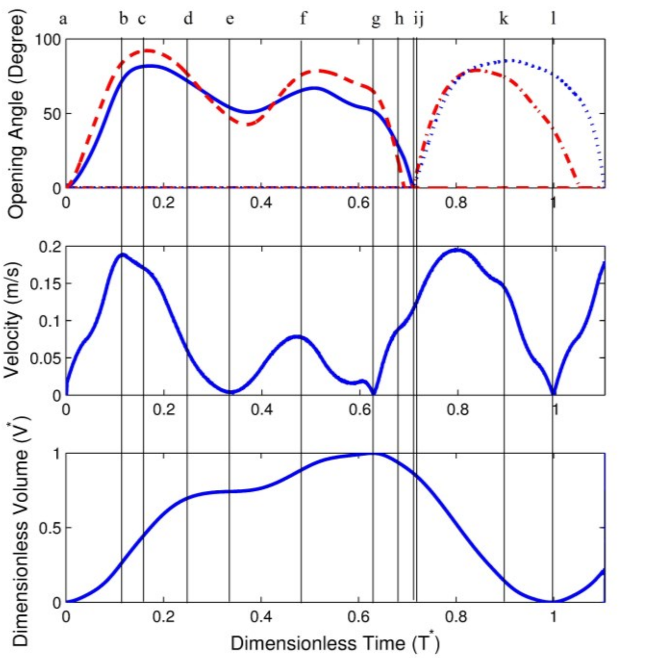

Figure 2. (a) Opening angles of mitral and aortic valves; (b) Inlet velocity; and (c) Dimensionless volume. Posterior mitral leaflet: - - -; Anterior mitral leaflet: ——; Left aortic leaflet: ··········; Right aortic leaflet: - · - (Note: the letters of 'a' to 'c' represents the time instant in the upper panel of Fig. 2(a).)

B. Patterns of Intra-ventricular Flow

Fig. 3 shows the streamlines of flow in the left ventricles from the onset of ventricular dilation to the end of ventricular contraction, with superimposed contours of vortices magnitude.

At the onset of ventricular dilation, the pressure gradient from atrium to apex leads to the opening of mitral valve as demonstrated in Fig. 3(a). With the advancement of diastolic phase, the leaflets of mitral valve gradually rotate towards the ventricle. The boundary layer near the leaflets of mitral valves develops at the peak of early filling phase at 0.12T* (Fig. 3(b)), which initiates vortex formation. With the further decrease of left ventricular pressure, the flow decelerates. At 0.16T* (Fig. 3(c)), vortex is found near the tip of both anterior and posterior leaflets of the mitral valve due to boundary layer separation. These vortices further develop in mid-diastole phase due to the decrease of left ventricular pressure, which leads the leaflet rotating inwards as shown in Fig. 3(d) and 3(e). The intra-ventricular flow is dominated by these vortices, which play an important role in preserving the kinetic energy of the blood and redirecting the flow towards the aortic valve.

Due to the increase of the atrial pressure at the phase of atrial contraction, the transmitral flow re-accelerates and leads to the reopening of mitral valve as shown in Fig. 3(f). During the second deceleration, the left ventricular pressure increases and results in inward motion of the leaflets of mitral valve. At the end of ventricular dilation, the ventricular volume is at its maximum and mitral valve is partially closed as shown in Fig. 3(g).

Regurgitation occurs from the onset of ventricular contraction to the complete closure of mitral valve, as shown in Fig. 3(h). When the mitral valve is fully closed (Fig. 3(i)), the vortices decay but do not vanish completely.

In the systolic phase, the left ventricle contracts and has high pressure, which leads to the rotation of the aortic leaflets outwards. The intra-ventricular flow is characterized by flow acceleration from ventricular wall to aortic valve. At 0.72T* (Fig. 3(j)), the vortices diminish. These vortices are flushed out of the ventricle at 0.90T* as shown in Fig. 3(k). Near the end of ventricular contraction, the decrease of the left ventricular pressure leads to the formation of another vortex inside the ventricular as seen in Fig. 3(l).

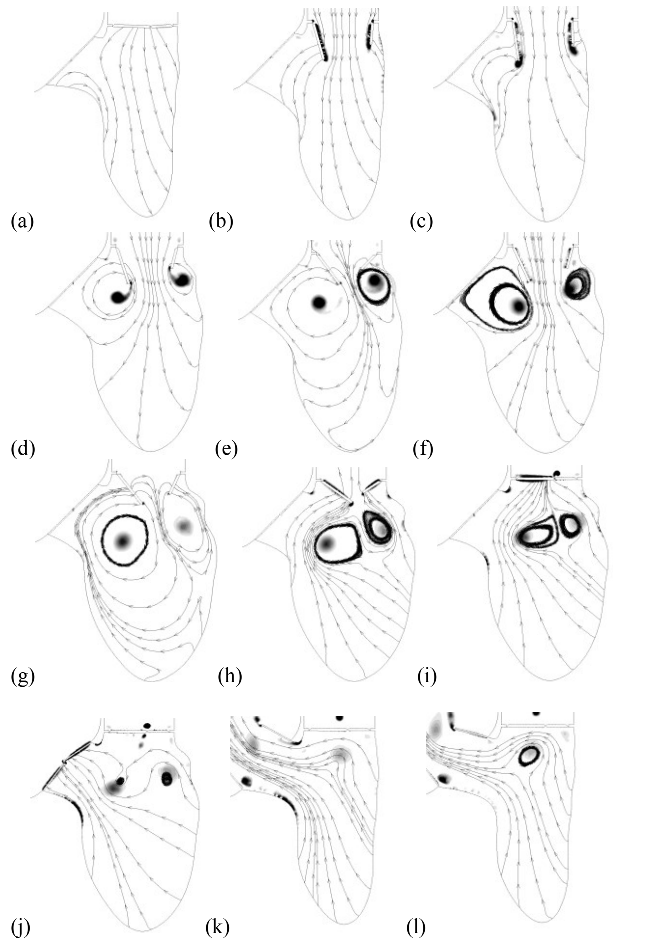

Figure 3. Development of intra-ventricular flow pattern at various time instants (a) 0.01T*, (b) 0.12T*, (c) 0.16T*, (d) 0.25T*, (e) 0.34T*, (f) 0.48T*, (g) 0.63T*, (h) 0.67T*, (i) 0.71T*, (j) 0.72T*, (k) 0.90T*, (l) 1.0T* (Note: these time instants are labelled in the upper panel of Fig 2(a))

IV. CONCLUSION

In this study, we have modeled the left ventricle with both mitral and aortic valves integrated in 2D. The geometry of left ventricle wall was pre-described according to the data acquired from patient-specific MRI. Cubic-spline interpolation approximation was adopted to generate the intermediate frames of the left ventricular wall and keep both velocity and acceleration of each boundary node continuous. FSI approach was applied to the leaflets of valves to simulate the interaction between the blood and heart valves, which play an important role in the intra-ventricular flow.

Generally, the numerical predicted flow patterns are similar to the measurements reported in literatures [11]. Only more persistent and sustained vortices are reported in [6]. In the study of [12], a large vortex was found near the root of aortic valve at the first peak of diastolic phase, with a 2D model without mitral valve. The formation of this vortex is due to the large cross-sectional area expansion in the streamwise direction and the intra-ventricular flow pattern varies considerably among normal subjects. However the vortex forms near the leaflet tip in the model with leaflets, and it develops mainly in the wake of leaflet. Therefore, the mitral valve is essential to predict the intra-ventricular flow.

The relatively large closing volume predicted in this numerical study should be related to the rigid leaflet, which functions like a mechanical valve. It is worth noting that the closing volume of mechanical valve was reported to be 3 times larger than that of the tissue valve [13]. An attempt was made to minimize the closing volume of the mitral valve by halving the moment of inertia of leaflet, but no significant difference was found. This is consistent with the study of [14].

The current 2D simulation is limited and leads to several complications. At first, the velocity at the inlet was derived from the motion of 2D ventricular wall. Therefore the velocity magnitude is smaller than that extracted from 3D ventricle. Secondly, it cannot predict the secondary flow induced by the 3D motion of ventricular wall. Thirdly, only longitudinal motion of mitral annulus was considered.

Despite the limitations of our model, the numerical study successfully demonstrates the vortex formation and development in the left ventricle. Therefore, the 2D approach is able to qualitatively predict the intra-ventricular flow, and it is beneficial to develop 3D numerical studies in the future.

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