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Characteristics of Hemodynamics in a Bileaflet Mechanical Heart Valve using an Implicit FSI Method

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***Abstract*—**Human heart valves diseased by congenital heart defects, rheumatic fever, bacterial infection, cancer may cause stenosis or insufficiency in the valves. Treatment may be with medication but often involves valve repair or replacement (insertion of an artificial heart valve). Bileaflet mechanical heart valves (BMHVs) are widely implanted to replace the diseased heart valves, but still suffer from complications such as hemolysis, platelet activation, tissue overgrowth and device failure. These complications are closely related to both flow characteristics through the valves and leaflet dynamics. In this study, the physiological flow interacting with the moving leaflets in a bileaflet mechanical heart valve (BMHV) is simulated with a strongly coupled implicit fluid-structure interaction (FSI) method which is newly organized based on the Arbitrary-Lagrangian-Eulerian (ALE) approach and the dynamic mesh method (remeshing) of FLUENT. The simulated results are in good agreement with previous experimental studies. This study shows the applicability of the present FSI model to the complicated physics interacting between fluid flow and moving boundary.

***Keywords*—**Bileaflet Mechanical Heart Valve, Fluid-Structure Interaction.

I. INTRODUCTION

THE heart has four valves which play a role to maintain the unidirectional flow of blood by opening and closing depending on the difference in pressure on each side. The heart valves could be diseased by congenital heart defects, rheumatic fever, bacterial infection, cancer and others. The diseased native heart valve may cause stenosis or insufficiency in the valve. The diseased valves should be replaced with a prosthetic valve in order for the patient to live a normal life.

Bileaflet MHVs (Mechanical Heart Valves) are the most commonly implanted prosthetic heart valves because they are permanently durable and reliable and have a hydrodynamic performance of low pressure drop. But, hemolysis caused by high velocity jets through the gap between the leaflets, the water hammer effects with the opening and shutting of leaflets and noise, etc have been a problem to be solved.

In vivo and *in vitro* experimental studies have yielded valuable information on the relationship between hemodynamic stresses and the problems associated with the implants. Recently, Computational Fluid Dynamics (CFD) has emerged as a promising tool, which, alongside experimentation, can yield insights of unprecedented detail into the hemodynamics of prosthetic heart valves. For CFD to realize its full potential, however, it must rely on numerical techniques that can handle the enormous geometrical complexities of prosthetic devices with spatial and temporal resolution sufficiently high to accurately capture all hemodynamically relevant scales of motion. Such algorithms do not exist today and their development should be a major research priority.

FSI approaches employed in the previous works can be broadly classified in moving-grid methods based on the Arbitrary-Lagrangian-Eulerian (ALE) approach [1] and fixed-grid methods such as the immersed boundary method [2]. A significant limitation of the ALE approach, however, stems from the fact that the mesh conforms to the moving boundary and as such it needs to be constantly displaced and deformed following the motion of the boundary. The mesh moving step could be quite challenging and expensive for complicated 3D problems. This situation is further exacerbated in problems involving large structural displacements for which frequent remeshing might be the only feasible approach for ensuring a well conditioned mesh at each time step of the simulation. Because of this inherent limitation, the ALE approach is only applicable to FSI problems involving relatively small structural displacements. In fixed grid approaches, on the other hand, the entire computational domain (including both the fluid and structure domains) is discretized with a single, fixed, non-boundary conforming grid system (most commonly a Cartesian mesh is used as the fixed background mesh). The effect of a moving immersed boundary is accounted for by adding forcing terms to the governing equations of fluid motion so that the presence of a no-slip boundary at the location of the interface can be felt by the surrounding flow. Because of the fixed grid arrangement, such methods are inherently applicable to FSI problems involving arbitrarily large structural displacement.

The aim of this study is to develop a Fluid-Structure Interaction (FSI) model using the moving-grid method with remeshing techniques, based on the Arbitrary Lagrangian Eulerian (ALE) approach for the physical interactions of MHVs, to investigate flows interacting with moving leaflets and to obtain guidance for further studies.

II. METHODS

An accurate solution of the Navier Stokes equations for deforming meshes is provided by the use of the ALE

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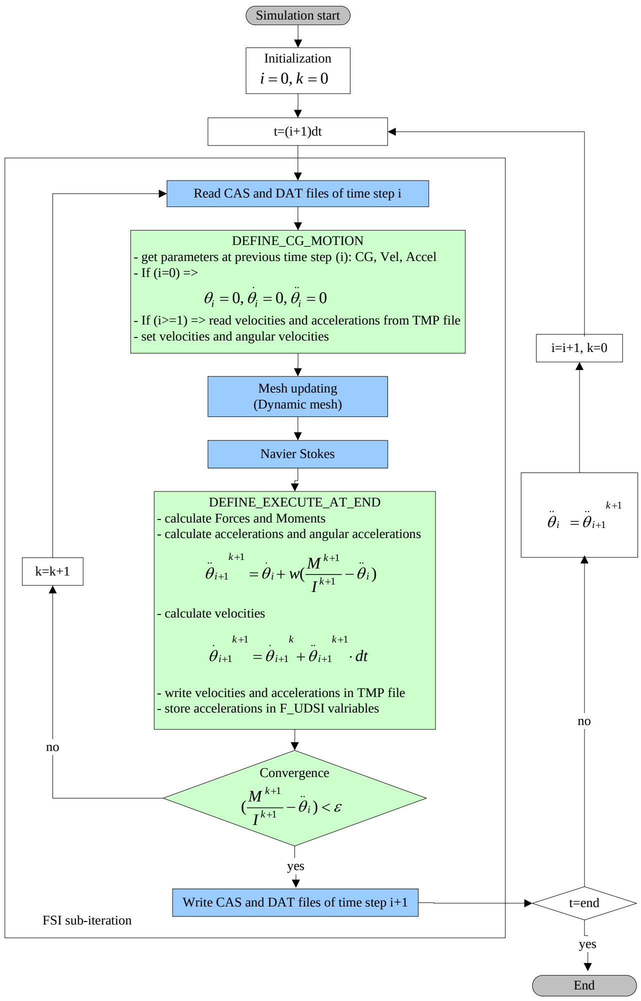


Fig. 1 Implicit coupling procedure scheme of the fluid-struc- ture interaction

formulation, which makes it possible to include grid velocities in the momentum and continuity equation of the fluid domain.

The ALE description conjugates Lagrangian and Eulerian features. The computational grid is neither moved with the boundary (Lagrangian) nor held fixed (Eulerian). Rather, it is moved in some arbitrarily specified way to give a continuous reconfiguration capability. Because of this freedom in moving the computational mesh offered by the ALE description, greater distortions of the continuum can be handled better than would be allowed by a purely Lagrangian method, with more resolution than that afforded by a purely Eulerian approach. The partitioned approach is used to simulate the interplay between leaflets and blood. This strategy preserves the fluid and the structural solvers as separate. Both parts are alternately integrated in time and the interaction is taken into account by the boundary conditions of both the solvers. As a direct consequence there exists an intrinsic time lag between the integration of the fluid and the structure, which can be avoided by repeating the interaction until both the solution consistently produce the same result.

The general scheme of the coupling procedure is shown in Fig. 1. As mentioned above, the fluid domain is solved using the finite volume method computational code Fluent (Ansys Inc., USA), which provides a number of features well suited to handle the specific problem of rotating boundaries. We will use a spring-based moving, deforming mesh module, which allows a robust mesh deformation handling by assuming that the mesh element edges behave like an idealized network of interconnected springs. In order to maximize the influence of the boundary node displacements on the motion of the interior nodes, no damping was applied to the springs. To preserve the quality of the mesh during the valve motion, the maximum admissible skewness of the computational cells is set. The Fluent remeshing algorithm is adopted to properly treat degenerated cells, which agglomerates cells that violate the skewness criterion, and locally remeshes the agglomerated cells. If the new cells satisfy the skewness criterion, the mesh is locally updated with the new cells (with the solution interpolated from the old cells); otherwise, the new cells are discarded (FLUENT Users Manual, 2007).

The moving deforming mesh module is used in conjunction with two user-defined subroutines, named DEFINE _EXECUTE_AT_END (moving body dynamics; MBD) and DEFINE_CG_ MOTION (center of gravity motion; CGM), respectively; at the beginning of each step the first one calculates and updates the kinematics of the leaflets on the basis of the moment applied to the leaflet, which is calculated by the second subroutine at the end of the time step, once the time step convergence has been achieved. An iterative call to the fluid solver is performed by an external subroutine in order to update the solution of the fluid dynamic field and achieve the convergence of the FSI cycle, until the difference between the external momentum divided by the inertia of the fluid (calculated by MBD) and the angular acceleration (imposed by CGM) is not below a threshold value. More in detail, since the valve leaflet is rigid body in rotation on a fixed pivot, the angular position is the only degree of freedom and leaflet dynamics can be calculated using:

$$I\ddot{\theta} = M_p + M_s \tag{1}$$

where I is the angular inertia, $\ddot{\theta}$ the angular acceleration of the leaflet, M_p the torque applied on the leaflet external surface by the pressure field, and M_s is the moment generated by shear stresses.

The acceleration value for the subsequent iteration within the generic time step i is updated through an under-relaxation scheme as

$$\ddot{\theta}_{i+1}^{k+1} = \dot{\theta}_i^{k} + w\left(\frac{M_p^{k+1} + M_s^{k+1}}{I^{k+1}} - \ddot{\theta}_i^{k}\right) \tag{2}$$

where k is the iteration index and w is the under-relaxation factor, which plays the role of damping changes in the acceleration produced during each iteration. Starting from the acceleration obtained in Eq. (2), the velocity and the displacement of the leaflet are calculated using the Newmark method.

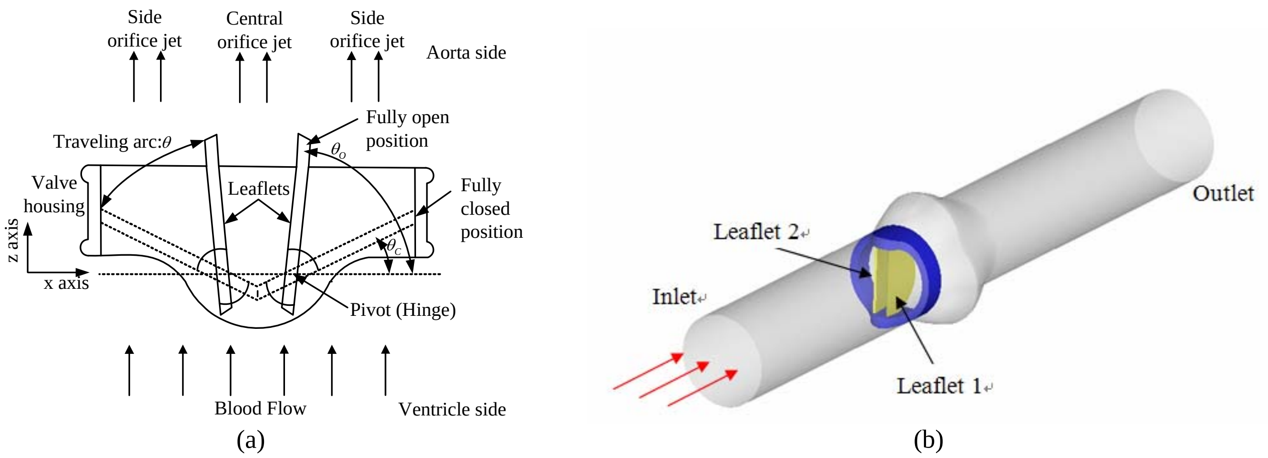


Fig. 2 3D model of the BMHV and aortic root with three sinuses: (a) kinematics of the BMHV (b) 3D model

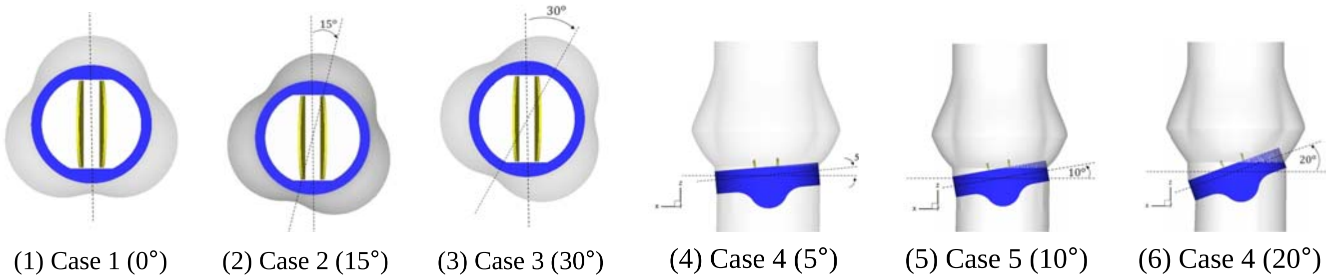


Fig. 3 Calculation cases

III. Analysis Model

The analysis model includes a BMHV and three sinuses. Ventricular and aortic vessels are assumed to be straight tracts. A three-dimensional model is made based on a 25 mm aortic BMHV of St. Jude Medical. The valve consists of one housing and two leaflets connected through pivots, and the leaflets can rotate between 25° and 85° (Fig. 2a). The external and internal diameters of the housing are 25 mm and 22.3 mm, respectively. The leaflets are 12.8 mm long and 0.65 mm thick. The hinge pivot is 2.25 mm apart from the central axis.

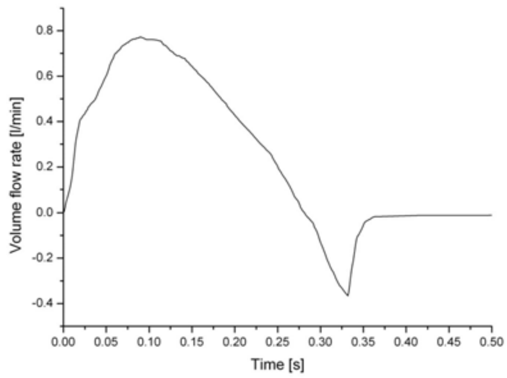


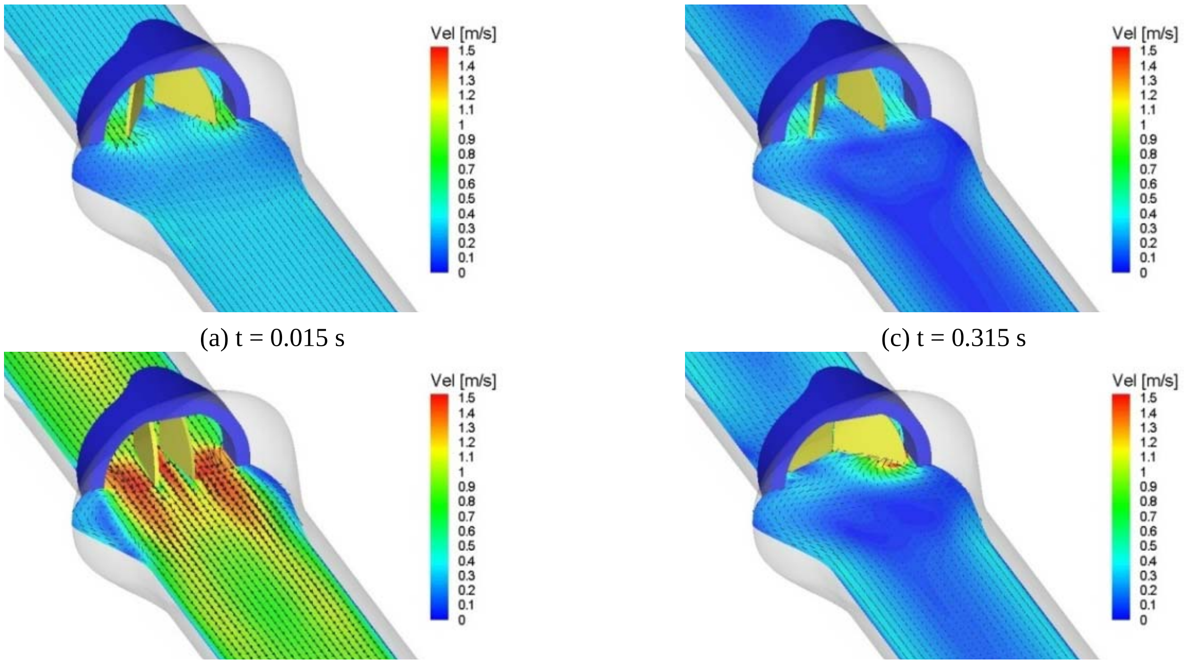
Fig. 4 Flow rate for inlet boundary condition

The valve is made of pyrolytic carbon with a density of 2,116 kg/m^3.

Rotating and tilting of the bileaflet valves are considered. In the cases of rotating the valves, three rotation angles (0°, 15° and 30°) are considered (Fig. 3a). When the rotation angle is 0°, the axis of reflection symmetry of the aortic sinus is parallel to the rotation axis of the two leaflets (Fig. 3 a-1). In the cases of tilting, three tilting angles (5°, 10° and 20°) are considered (Fig. 3b).

Blood is assumed to be an incompressible Newtonian fluid with a density of 1,000 kg/m^3 and a dynamic viscosity of 3.5 cP. The computational domain shown in Fig. 2b is discretized with 200,000 tetrahedral cells in a curvilinear grid system. The volume flow rate measured in *vitro* study (3) is imposed as the inlet condition (the ventricular side) and the pressure in the outlet is set to 0 mmHg. Leaflet behavior is calculated in the present FSI algorithm. No-slip conditions are imposed for the walls. At the beginning of the calculation (t=0 s), the flow is assumed to be at rest and the valve is closed. The calculation is carried out over three cardiac cycles to confirm cyclic independence and this paper presents results from the third cycle.

The time increment Δt is very carefully controlled between 0.1 ms to 10 ms in association with the calculated open angle variance of the leaflet. The time increment Δt^{n+1} for the next step is decided based on the open angle variance ($\Delta\theta^n = \theta^n - \theta^{n-1}$) at the previous time step, which allows us to keep the variation of the open angle less than 2° between consecutive time steps.



(a) t = 0.015 s

(c) t = 0.315 s

(b) t = 0.090 s

(d) t = 0.319 s

Fig. 5 Transient velocity field

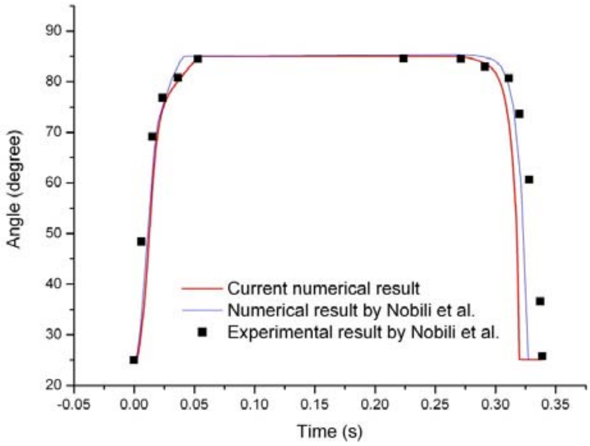


Fig. 6 Comparison of transient angle of leaflet

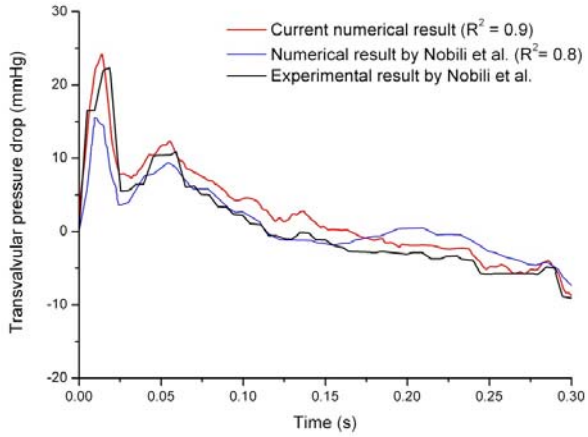


Fig. 7 Comparison of pressure drop

IV. Result

Validation of current numerical calculation

The data used for validation of the present FSI calculation were obtained from a experimental and numerical study on the dynamics of a bileaflet valve by Nobili et al. [4]. Fig. 5 shows the velocity fields of current numerical calculation.

Fig. 6 and Fig. 7 show the comparison between the numerical and experimental results for transient angle of a leaflet and transvalvular pressure drop, respectively. The pressure was evaluated at a distance of 2.5 diameters upstream and downstream of the valve site. Even though there is slightly difference in the closing phase of the leaflet, the time duration of the opening phase of the leaflet is predicted very accurately in the present result. It can be seen clearly that the present numerical result for the angle of the leaflet is good agreement with the experimental data qualitatively. Also, as shown in Fig. 18, the pressure drop of the present result is good agreement with the experimental data qualitatively.

Obviously, it can be revealed from the comparison between the numerical and experimental results that the current numerical method is applicable for modeling the blood flow passing through the bileaflet valves and leaflets' motions.

Effect of the rotation of the valve

Transient variations of the opening angle of the left and right leaflets with different rotation angles are obtained and plotted in Fig. 8. It is observed that, in the case 1, the left and right leaflets behave symmetrically in conjunction with physically symmetric boundary conditions. In the cases of rotation, In the cases with rotation, the transient angles of leaflet 1 and leaflet 2 are analogous to each other in the opening phase but they are slightly different each other in the closing phase.

The averaged pressure drop is generally increased with the rotation angle (Fig. 9) and the averaged pressure drop in case 3

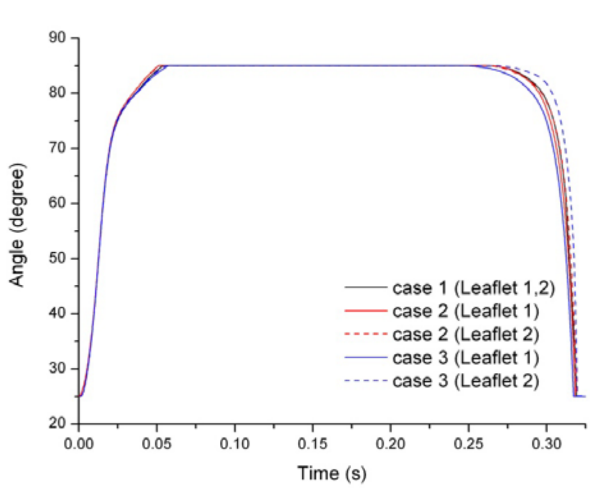


Fig. 8 Transient angle of leaflets in cases of rotation

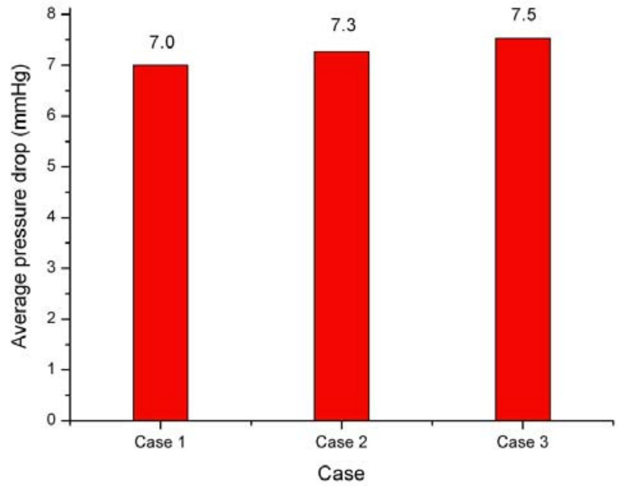
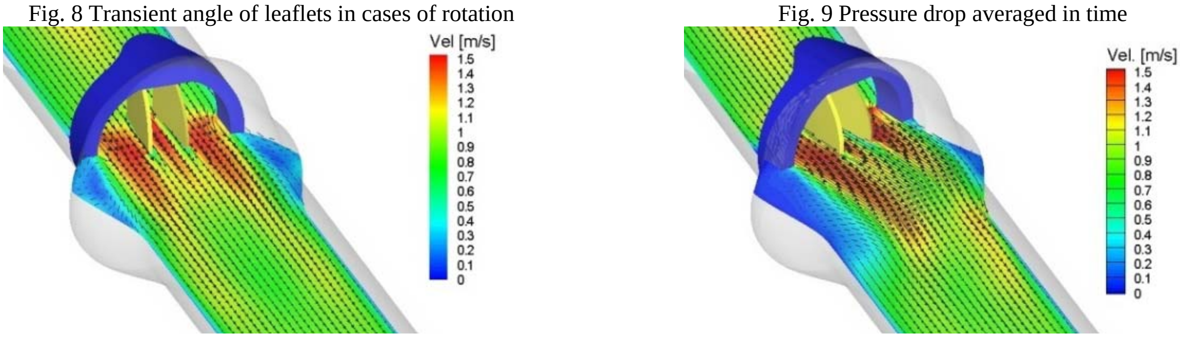


Fig. 9 Pressure drop averaged in time



(a) case 1

(b) case 6

Fig. 10 Comparison of velocity fileds ofuntilted case and tilted case at peak systole

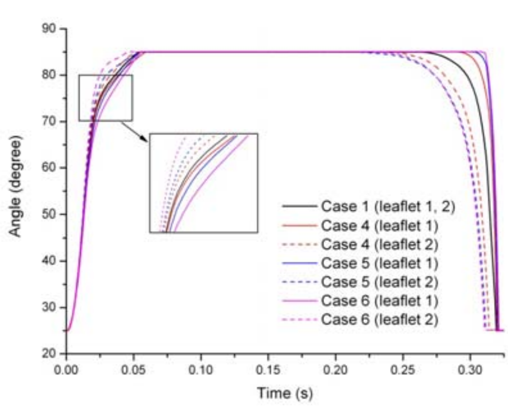


Fig. 11 Trangient angle in cases of tilt

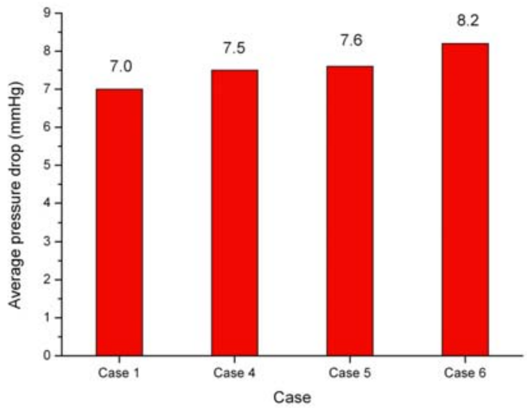


Fig. 12 Pressure drop in cases of tilt

is 7.5 mmHg which value is increased about 7 % in compared to the averaged pressure drop in case 1. Higher pressure drop means that the heart have to work harder.

Effect of the tilt of the valve

The flow pattern in case of tilt is obviously different from the flow pattern in untilted case (case 1) (Fig. 10). Transient variations of the opening angle of the left and right leaflets with different rotation angles are obtained and plotted in Fig. 11. In tilted cases, the one leaflet moves faster than another leaflet in the opening phase as well as in the closing phase. When the bileaflet valve is tilted, a blocking effect of the one leaflet on the blood flow is greater than that of another leaflet since the bileaflet valve is tilted to the one side.

The averaged pressure drop is generally increased with the angle of tilt (Fig. 12) and the averaged pressure drop in case 7 is 8.2 mmHg which value is significantly increased about 17 % in compared to the averaged pressure drop in case 1.

V. Conclusion

A numerical simulation of three-dimensional, physiological flow interacting with the moving leaflets of the BMHV has successfully carried out. The predicted results have been in good agreement with available results in the literature. The applicability of the present FSI model in the complicated physics of the BMHV has been validated. The results are important for identifying approaches to further improve BMHV design and performance, and to provide guidelines for further studies.

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