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# BILEAFLET PROSTHETIC HEART VALVE DISEASE: NUMERICAL APPROACH USING 3-D FLUID-STRUCTURE INTERACTION MODEL WITH REALISTIC AORTIC ROOT

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## ABSTRACT

Surgical replacement, in the incidence of severely diseased heart valve, is vital in order to restore the normal heart function. Every year around 280,000 valve replacements occur around the world, half of them are bileaflet mechanical heart valves (BMHVs). Despite the remarkable improvement in valve design resulting in minimizing prosthetic valve complications (thromboembolic events or pannus formation), these complications are still possible with BMHV Implantation. As a consequence, an obstruction in one or both MHV leaflets could happen and threaten the patient life. In the present study, an obstructed bileaflet MHV with different percentages of malfunction was simulated assuming 3-D fluid structure interaction (FSI) adapting k-w turbulence as a robust model for the transitional flow using 2.5 million elements and creating a realistic aortic root model with three sinuses.

Velocity contours for different percentages of malfunction were compared mainly at B-datum plane and the perpendicular plane to the B-Datum. Also, the development of coherent structures was investigated. Clinically, the maximum pressure gradients were estimated by mimicking the Echo Doppler assumptions (using the simplified Bernoulli equation).

## INTRODUCTION

Valve stenosis or incompetence at severe levels reduce the performance of the heart and place additional stress and strain upon it. In many cases, surgical replacement of the diseased valve with a bioprosthetic or a mechanical heart valve is necessary to restore normal heart function. Due to a longer lifespan, around 50% of valve replacements worldwide are mechanical heart valves. Usually, a patient with a Bileaflet Mechanical Heart Valve (BMHV) must take anticoagulant medication lifelong because of risks of thromboembolic complications which, in turn, could restrict the movement of the leaflets. Another potential complication associated with mechanical valves is the pannus formation (Montorsi et al. 2003). Non-invasive diagnosis and evaluation of the severity of BMHV dysfunction using Doppler echocardiography and magnetic resonance imaging is not straight forward, usually due to theoretical, technical or accessibility limitations. It seems important, therefore, to investigate the flow downstream of a dysfunctional BMHV and to investigate the limitations of current diagnosis techniques.

Most numerical and experimental studies on BMHVs have focused on normal functioning valves with a high emphasis on the velocity field, transvalvular pressure drop and blood components damage. Numerically, different approaches were considered. For the geometry, 2-D and 3-D were conducted with simple or realistic aortic root. Also, a simple aortic arch (straight tube) and realistic curved one were adapted. In addition, both, steady and pulsatile flow, were simulated. Generally, steady flow was used to study the flow at the peak of systolic phase. While the pulsatile flow was concentrating more on the whole cardiac cycle with more than one of the three phases (acceleration, peak and deceleration phases) (Yoganathan et al. 2004, Bluestein et al. 2010).

Recently, Direct Numerical Simulation (DNS) with fully FSI was performed by Dasi et al. (2007), Nobili et al. (2008) and Tullio et al. (2009). However, the high computational cost required for DNS limits its applicability to practical clinical problems.

Finally, few studies investigated the blood flow through an obstructed BMHV. Baumgartner et al. (1993) showed, in vitro, that a defective BMHV (carbomedics valve with one leaflet blocked) leads to an increase in the energy loss through the valve resulting in a significant discrepancy between catheter and Doppler echocardiographic transvalvular pressure gradients. This reduction actually results from a less significant pressure recovery downstream of the defective BMHV. This was confirmed numerically, in a recent study performed by Smadi et al. (2009). In this study, the authors also suggested new diagnostic parameters to investigate non-invasively the severity of BMHV malfunction using Doppler echocardiography and magnetic resonance imaging.

## NUMERICAL METHOD

A 3-D model for 25 mm St. Jude Medical Hemodynamic Plus valve was created for the purpose of this study. The restriction of the leaflet motion was applied only on one of the two leaflets (Smadi et al. 2010). The position of the leaflet was varied from fully opened position (normal function) to fully closed position (100% dysfunction) with one intermediate position (50% dysfunction) (Fig. 1).

The malfunction was present, only, during the leaflet opening phase (stenosis) while the leaflet is functioning properly during the closure (no extra regurgitation) (Aoyagi et al. 2000). The simulations were performed under unsteady conditions with an experimental pulsatile flow as inlet condition (Fig. 1) and ambient pressure at the outlet. The mean cardiac output was 5 L/min and the heart rate was 70 bpm (systolic phase duration 0.3 s). Blood was simulated as a Newtonian fluid with a density of 1060 kg/m3 and a dynamic viscosity of 0.0035 Pa s. The assumption of a Newtonian fluid behavior is realistic for blood flow in large arteries as the aorta. The inlet conditions corresponded to a $Re_{max}$ = 8023; $Re_{average}$= 3820 and Womersley number = 16.2.

The current simulation was carried out using commercial software (FLUENT) and adapting Arbitrary Lagrangian Eulerian method (ALE) for re-meshing the fluid domain after the solid part (the leaflets) moved to a new position. Briefly, a spring-based method was used for treating the deformed mesh where the element edges behave like an idealized network of interconnected springs. The spring constant was adjusted to be zero (no damping was applied to the springs). Moreover, the maximum skewness of the mesh elements was set to be 0.7. Therefore, at each time step the skewness will be checked and the elements with the higher value than the threshold will be re-meshed (Nobli et al. 2008). Meshing the geometry was done by GAMBIT 2.4 (Fluent Inc.) and by using 2.5 million elements. This method was used and validated by Dumont et al. (2004), Dumont et al. (2007) and Nobili et al. (2008).

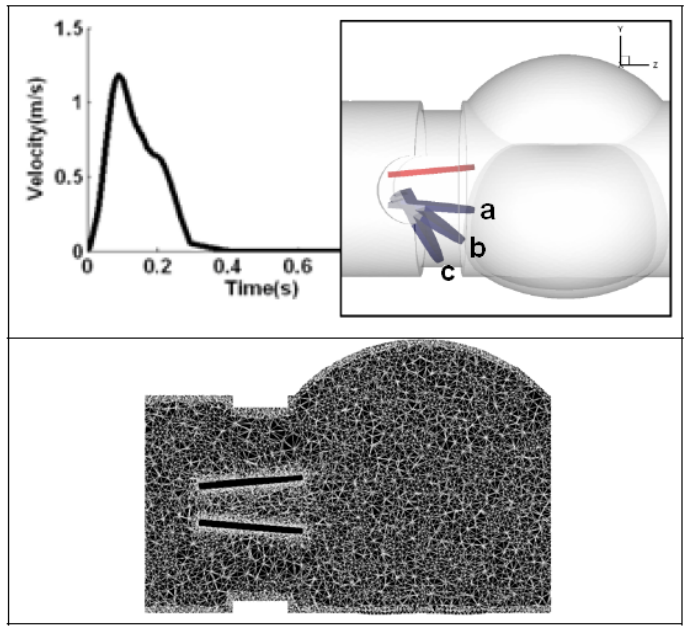


**Figure 1. The geometry and the mesh quality of 25 mm St. Jude HP BMHV with the instantaneous velocity as an inlet condition. Three different opening restrictions were applied on one leaflet: a) 0% obstruction (completely opened), b) 50% obstruction (partially opened) and c) 100% obstruction (completely closed).**

### Turbulent FSI Approach

While the *K-E* approach is appropriate for high-Re flows, it may have some limitations for some low-Re flows. As physiological blood flow in the low-Re range (Re<10,000), and the blood flow through a BMHV is characterized with laminar-transitional-turbulent behavior, the Wilcox's low-Reynolds model was found able to accurately predict its main flow characteristics (Wilcox, 1998, Bluestein et al., 2000).

In the present study, two-equation transitional $k-\omega$ model was used to capture the laminar-transitional-turbulent flow phenomenon. Time-averaging or Reynolds averaging has been used as a mean of analyzing turbulence by separating fluctuating properties with their time-mean values. Thus, the true velocity ($u_i$) is defined by: $u_i = \overline{u}_i + u_i'$, where the overbar refers to time-average and prime refers to fluctuation from this average. When this is substituted in the general

 

Navier-Stokes equations, a new term will be introduced, i.e., the Reynolds stresses ($-\rho\overline{u_i'u_j'}$). To close the governing equations with the new extra variables, two-equation transitional $k-\omega$ model was used through which these Reynolds stresses are approximated using the Boussinesq relation for incompressible flow (Wilcox, 1998)

$$-\rho\overline{u_i'u_j'} = \mu_t\left(\frac{\partial u_i}{\partial x_j}+\frac{\partial u_j}{\partial x_i}\right)-\frac{2}{3}\left(\rho k+\mu_t\frac{\partial u_i}{\partial x_i}\right)\delta_{ij} \quad (1)$$

Where, $u_i$ is the average velocity in $i$ direction, $\mu_t$ is the turbulent eddy viscosity and $k$ is the turbulent kinetic energy. A second-order upwind scheme was selected to be the discretization scheme for the convection terms of all governing equations. For all transient calculations, a second-order temporal discretization scheme was used. The mass-momentum equations were solved using the PISO solver and all results were converged to residuals of < $10^{-5}$, unsteady simulation in general required 5-10 iterations per time step. Moreover, additional care was taken close to the wall and leaflet surfaces to maintain y+ << 1 (y+ < 0.4). The time step was set to be 0.05 ms and two cycles were simulated before extracting the results.

The motion for the leaflets is assigned and controlled by an external subroutine based on experimental data. Therefore, the full nonlinear fluid structure interaction concept is not considered. Moreover, as the scope of the current study is to evaluate the influence of the presence of a dysfunction on the blood flow downstream of the MHV and to compare the healthy and the dysfunctional cases with each other, the angular velocity for the leaflets was kept the same for the all cases.

## RESULTS AND DISCUSSION

### Velocity Contours

Figure 2 shows axial velocity contours at three different percentages of dysfunction at the peak of systolic phase. In healthy model (0% obstruction), the influence of the orientation of the valve leaflets on the flow field was noticed by creating two wakes and three jets (one central and two laterals). Also, the circulation within the sinus area was found and represented by negative value of the axial velocity.

The strength of rotation within the sinus area was proportional to the percentage of malfunction. The negative axial velocity magnitude reached the maximum with 100% malfunction case ($V_s$= -1.5 m/s). Moreover, more circulation zones appeared with the defective cases

Introducing 50% malfunction to one leaflet alerted significantly the flow distribution. Although the three jets still exist, the velocity profile distribution is different from the healthy model. The central and upper lateral jets get close from each other by shifting the central jet peak velocity toward the upper lateral side, while the lower lateral jet goes more toward the lower wall. For 100% malfunction, the flow behavior is close from 50% malfunction case except having only two dominant central and upper lateral jets instead of having three jets. It is clear with the dysfunction that the central common flow is no more dominant and the lateral orifices gain more importance as the majority of the flow passes through one of them.

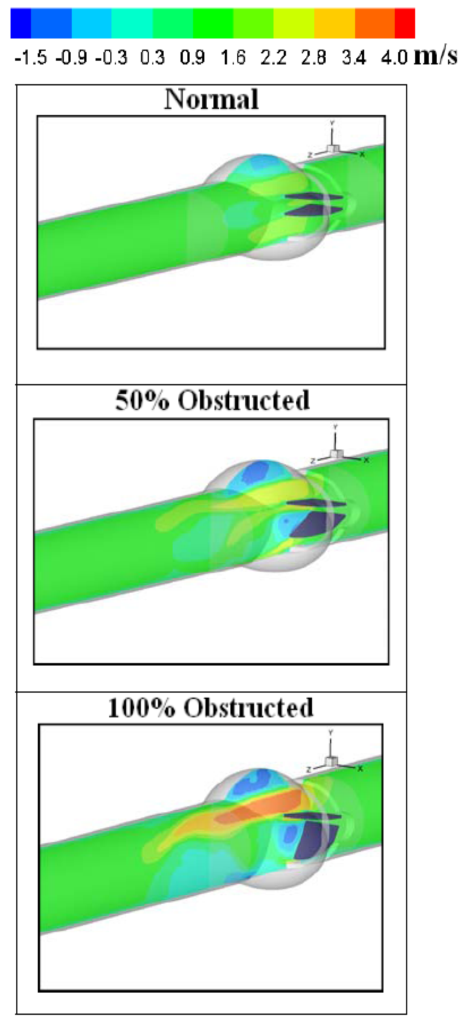


**Figure 2. Contours of velocity magnitude through the bileaflet valve for different percentages of malfunctions at the peak of the systolic phase.**

The maximal velocity increased dramatically (from ~ 2.3 m/s for the healthy case to ~4 m/s for 100% obstruction), which in turn will increase the Doppler peak pressure gradient (from 21 mmHg to 64 mmHg). However, this dramatic pressure gradient is not enough to confirm the presence of valve obstruction as there are other factors that could lead to the same result (i.e high flowrate, left ventricle out flow obstruction and prosthesis patient mismatch).

Figure 3 shows the maximum transvalvular pressure gradient (TPGmax) for different percentages of malfunctions using echo Doppler velocity measurements from Smadi et al. (2010) and numerical simulation results. The TPGs were determined using the simplified Bernoulli equation ($TPG=4V^2$).The results are in good agreement with the 2-D model used by Smadi et al. (2010). Therefore, the 2-D model is valid when the scope of the study is focusing on pressure or velocity magnitude.

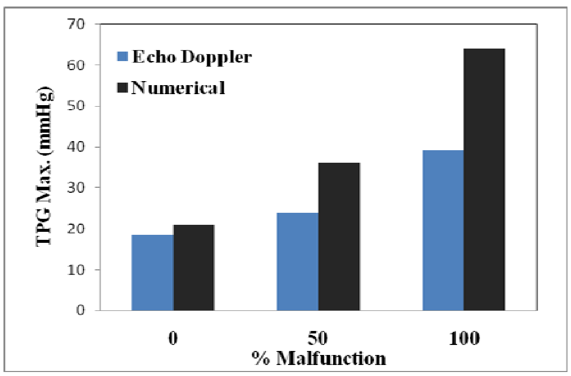


**Figure 3. Comparisons between numerical and Echo Doppler maximum Transvalvular Pressure Gradient (TPGMax).**

### The Coherent Structure

Figure 4 shows the isosurface of the vortical structure, considering Q-criterion, downstream of the valve for three different malfunctions at the peak of the systolic phase. The value of the isosurface was selected to have a high magnitude (100,000 $S^{-1}$) for all cases.

At the peak of systolic phase, and for the healthy case, three major vortex rings were found within the three sinuses which is consistent with the findings in the literature. However, introducing the malfunction significantly disturbed the flow downstream of the valve and the coherent structures were dominant through the whole aortic root. In case of 100% dysfunction, the vortex structures traveled for largest distance downstream of the valve comparing to the other two cases. In the mean time, in the case of 50% obstruction, the vortex structures covered more area within the sinuses comparing to the other cases. Generally, in healthy BMHVs, the maximum viscous and shear stresses is found at the peak instant and downstream of the trailing edges of the two leaflets where the vortex shedding (von Kármán vortex street) occurs (Ge et al. 2007). On the other hand, in the dysfunctional BMHVs, the vortical flow is covering larger area. As a consequence, blood elements remain for longer period in a region of elevated shear stress. Therefore, the platelet activation and/or red blood cells damage can occur. As the coherent structure in the partially defective valve was more dominant comparing to the completely defective one, the partially defective bileaflet MHV could be developed to a severe case (100% obstruction) with relatively high rate and within a short period. Therefore, detecting the malfunction at early stages is crucial and any delay in diagnosis could have life threatening consequences.

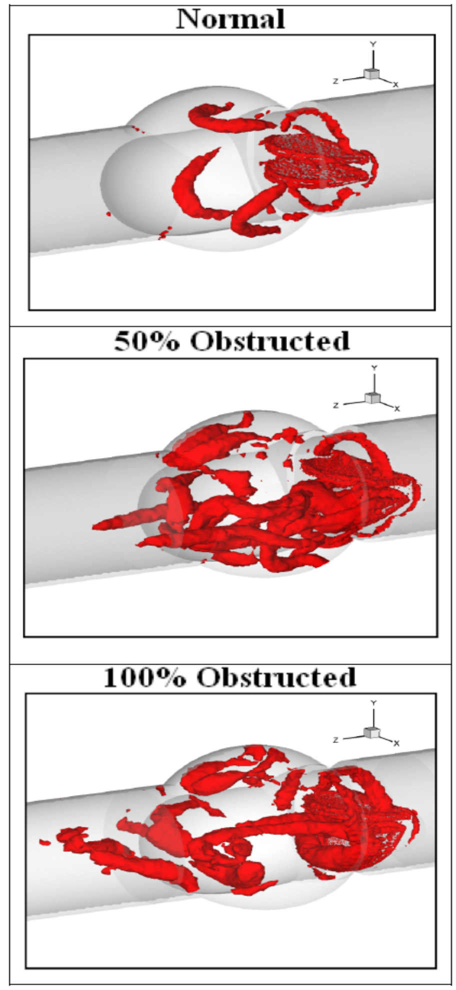


**Figure 4. 3-D coherent structure based on Q-criterion at the peak instant of the systolic phase for different malfunctions.**

### Velocity Magnitude (two vs. three velocity components)

Figure 5 shows the discrepancy between using the three velocity components and using only the two in-plane velocity components to calculate the velocity magnitude. In the current study, only, B-datum plane is presented where the maximum discrepancy was found. In normally functioning BMHV, a minor discrepancy was found in the sinus area, which is in a good agreement with Kaminsky et al. 2007 findings. However, the discrepancy was more significant when the malfunction is present and it was proportional to the percentage of the malfunction. In the healthy case, the disagreement between the two calculated velocities was mainly in the sinus area. In the mean time, in the case of obstructed bileaflet MHV, the discrepancies covered larger areas (the core area and the area downstream of the sinuses).

It is important to mention that 2-D Particle Image Velocimetry (PIV) systems used frequently to explore the flow dynamics downstream of normal mechanical heart valves (Burker 1997, Bladucci et al. 2004, Chi-Pei et al. 2009).

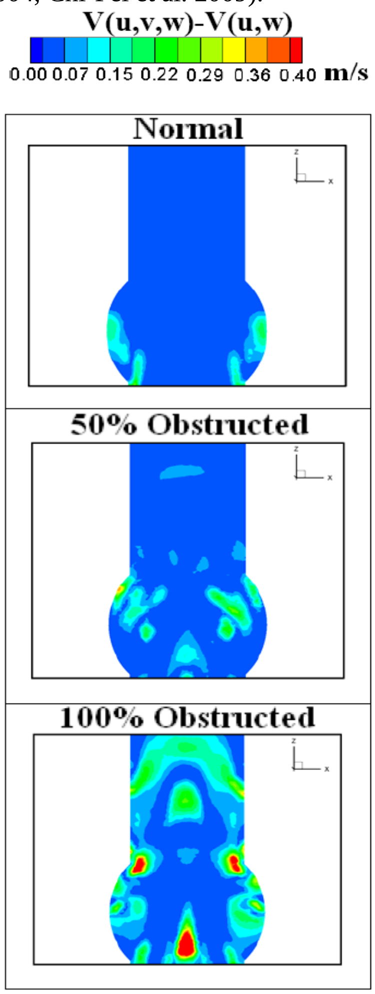


**Figure 5. The differences between the two-component and three-component velocity magnitude (V(u,v,w) and V(u,w)).**

However, the contribution of the out of plane velocity component to the total velocity magnitude, under healthy conditions, was studied by Kaminsky et al. 2007. There was a slight difference between the velocity calculated by three components comparing to the one calculated by only two components, especially in the sinus area. On the other hand, as shown in fig. 4.4, introducing a dysfunction to the leaflet/s disturbs the flow downstream of the BMHV and increases the importance of out of plane velocity components and as a consequence alerts the importance of using a stereoscopic PIV system instead of using 2-D PIV system.

## Conclusion

In conclusion, this study showed that the flow nature is strongly three dimensional and time dependent especially with the existence of the dysfunction. Therefore, with the presence of valve obstruction, the pulsatile 3-D simulation should be adapted when the evolution of the vortical structure downstream of the bileaflet MHV is the objective of the study. Also, there was a strong presence for the vertical structures downstream of the defective BMV. Hence, the residential time for the blood elements in a relatively high shear stress region could, significantly, increase the likelihood of platelets activation and/or blood hemolysis.

Finally, for capturing the velocity field downstream of the defective BMHVs, especially in the sinus of valsalva region, stereoscopic PIV measurements are more accurate compared to 2-D PIV ones.

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## REFERENCES

Aoyagi, S., Nishimi, E., Kawano, M., Tayama, H., Fukunaga, S., Hayashida, N., Akashi, H., 2000, “Obstruction of St. Jude Medical Valves in the Aortic Position: Significance of a Combination of Cineradiography and Echocardiography,” *American Society for Thoracic Surgery Vol. 10 (4)*, pp. 339–344.

Baumgartner, H., Schima, H., and Kuhn, P., 1993, “Effect of Prosthetic Valve Malfunction on the Doppler-Catheter Gradient Relation for Bileaflet Aortic Valve Prostheses”, *Circulation, vol. 87*, pp. 1524-4539.

Bluestein, D., Chandran, K. B., Manning, K. B., 2010, “Towards Non-Thrombogenic Performance of Blood Recirculating Devices,” *Annals of biomedical engineering, Vol. 38(3)*, pp. 1236–1256.

Dasi, L.P., Ge, L., Simon, H.A., Sotiropoulos, F., Yoganathan, A.P., 2007, “Vorticity dynamics of a bileaflet mechanical heart valve in an axisymmetric aorta, *Physics of Fluids Vol.19 (6)*, pp. 067105-17.

De Tullio, M.D., Cristallo, A., Balaras, Verzicco, E., 2009, “Direct Numerical Simulation of the Pulsatile Flow Through an Aortic Bileaflet Mechanical Heart Valve,” *J. Fluid Mech. Vol. 622*, pp. 259–290.

Dumont, K., Astijen, J.M.,Van De Vosse, F.N., and Verdonck, P.R., 2004, “Validation of a Fluid-Structure Interaction Model of a Heart Valve Using the Dynamic Mesh Method in Fluent”, *Computer Methods in Biomech. And Biomed. Eng., vol. 7(3)*, pp. 139-146.

Dumont, K., Vierendeels, J., Kaminsky, R., “Van Nooten, G., Verdonck, P., Bluestein, D., 2007, “Comparison of the Hemodynamic and Thrombogenic Performance of Two Bileaflet Mechanical Heart Valves Using a CFD/FSI Model,” *J Biomech Eng. Vol. 129(4)*, pp. 558-65.

Ge, L., Dasi, L., Sotiropoulos, F., Yoganathan, A., 2008, “Characterization of Hemodynamic Forces Induced by Mechanical Heart Valves: Reynolds vs. Viscous Stresses,” *Annals of biomedical engineering, Vol. 36(2)*, pp. 276–297.

Kaminsky, R., KALLWEIT, S., WEBER, H.J., Claessens, T., Jozwik, K., Verdonck, P., 2007, “Flow Visualization through Two Types of Aortic Prosthetic Heart Valves Using Stereoscopic High-Speed Particle Image Velocimetry,” *Artificial Organs Vol. 31(12)*, pp. 869-879.

Montorsi, P., Cavretto, D., Alimento, M., Muratori, M., and Pepi, M., 2003, “Prosthetic Mitral Valve Thrombosis: Can Fluoroscopy Predict the Efficacy of Thrombolytic Treatment?”, *Circulation, vol. 108*, pp.79-84.

Nobili, M., Morbiducci, U., Ponzini, R., Del Gaudio C., Balducci A., Grigioni, M., Montevecchi , F.M., Redaelli, A., 2008, “Numerical Simulation of the Dynamics of a Bileaflet Prosthetic Heart Valve Using a Fluid-Structure Interaction Approach, *Journal of Biomechanics Vol. 41 (11)* , pp. 2539–2550.

Smadi, O., Fenech, M., Hassan, I., Kadem, L., 2009, “Flow Through a Defective Mechanical Heart Valve: A Steady Flow Analysis, *Medical Engineering & Physics Vol. 31 (3)* , pp. 295–305.

Smadi, O., Hassan, I., Pibarot, P., Kadem, L., 2010, “Numerical and Experimental Investigations of Pulsatile Blood Flow Pattern Through a Dysfunctional Mechanical Heart Valve,” *Journal of Biomechanics* , In Press, Corrected Proof.

Wilcox, D.C., 1998, “*Turbulence Modeling for CFD*,” 2nd Edition, DCW Industries, La Canada, California.

Yoganathan, A. P., He Z., and Jones, S.C., 2004, “Fluid Mechanics of Heart Valves*”, Annu. Rev. Biomed. Eng., vol. 6*, pp.331–62.

 