Hemodynamics and Mechanical Behaviors of Aortic Heart Valves: A Numerical Evaluation

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Abstract — **This study investigates the hemodynamics of aortic heart valves under normal conditions as well as two severe diseases, which would be fundamentals for an assessment of mechanical behaviors of polyurethane (PU) prosthetic heart valves. Analysis results highlight that leaflet opening situation and valve geometry affect the shear stress distribution and vortex flow regime. The interactive impact between low and high wall shear stress on relation to heart valve diseases have been also demonstrated. The results show that low density PU material achieves good hydrodynamic function but produces high stress, while high density PU resists motion of leaflet but reduces the stress significantly. This study also proves that low Young's modulus PU leaflets achieve good hydrodynamic function while reducing the stress exerted upon the leaflets, and vice versa for high Young's modulus.**

Keywords - polyurethane prosthetic valve, aortic valve, hemodynamic, wall shear stress, Young's modulus.

I. INTRODUCTION

Nowadays, heart valve disease is one of the biggest health problems all over the world. Heart valve diseases occur in all four heart valves, but most frequently affect the aortic valve being responsible for high mortality rates. There are two main processes that can affect the aortic valve including aortic stenosis and aortic insufficiency. The most common treatment for a malfunctioning aortic valve due to stenosis or insufficiency is replacement of the valve, using mechanical valves, or biological prosthetic valves.

The viscous drag (shear stress) provided by flowing blood exerts a potent atheroprotective effect. Atherosclerotic lesions preferentially develop in areas with turbulent flow, whereas regions with uniform laminar flow are protected. The endothelial lining is the primary sensor of wall shear stresses, and functions as a transducer of these biomechanical stimuli into biological responses within the vessel wall. The shear stress is highly dependent upon the direction of blood issuing through the orifice of human aortic valve and the location of blood jet impingement. Meanwhile, the orifice shape of aortic valve as well as the location of blood jet impingement could be deformed under influence of valvular diseases. It is a demand to study effects of heart valve disease on blood shear stress.

Many works tried to combine all strong points of mechanical and biological prosthetic valve into single one. Although a prosthetic valve which is ideally suitable for all clinical situations is unavailable now, polyurethane valves have been considered as an optimal solution to solve this problem. Polyurethane valves have leaflet shape mimics that of natural valve, good biocompatibility but made of polymeric material whose flexibility, and withstand many cycles of stress and deformation capability. The properties of polyurethanes are also varied by altering the composition or proportion of either the hard or soft segment constituents.

Current study aims to investigate the hemodynamic characteristics of blood flow and heart valve leaflets under normal conditions as well as two severe diseases, which would be considered as fundamental for an assessment of mechanical behaviors of polyurethane (PU) prosthetic heart valve. The author utilizes a three dimensional Computational Fluid Dynamics (CFD) model to understand the complex flow pattern through a normal and diseased human aortic valve, concentrating on the analysis of the wall shear stress over the leaflets and endothelial lining, and specify the crucial requirement of prosthetic aortic heart valve materials in order to find out an appropriate polyurethane for prosthetic aortic valves.

II. METHODOLOGY

A. Geometry development

1) Normal aortic valve

The leaflet geometry design in this study is essentially developed by using quasi-steady method which combined works of Swanson [1] and Hsu [2].

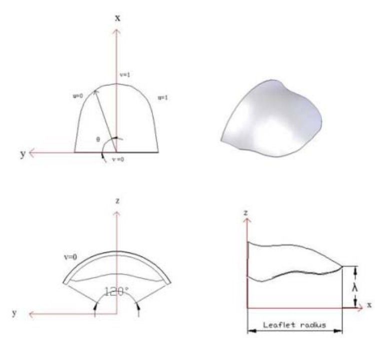

Figure 1. Coordinate system - valve leaflet.

2) Prolapsed valve

Valve prolapsed is characterized by the displacement of an abnormally thickened valve leaflet. Within limitation of current study, we assumed that one or two leaflets of aortic valve have problem and moves with different velocity to the others.

3) Bicuspid valve

The bicuspid aortic valve model was referred to Pettersson model [3]. The valve was presented with one normal-looking cusp and one larger cusp, appearing to be the result of fusion of two cusps normally encountered in the tricuspid form of aortic valve (conjoint cusp).

B. Numerical model

1) Physical geometry

The physical dimensions of ventricle, sinus and aorta are utilized in this study as shown in Fig. 2. The left ventricular entrance radius is 29.50mm and the ventricular tract radius is 13.28mm. The largest radius of sinus is 16.76mm. Aortic exit radius is 14.85mm.

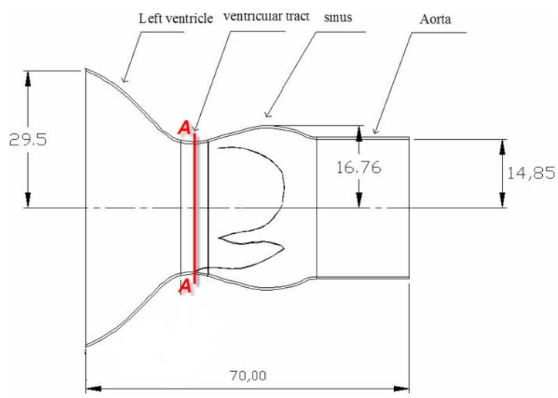

Figure 2. The Physical Geometry Model

2) Governing Equations

To simulate the hemodynamic characteristics of the blood flow, ANSYS CFX10.0 software with the finite volume method of analyzing the three-dimensional Reynolds-averaged Navier-Stokes equations is adopted. Blood is treated as Newtonian fluid with a viscosity of 3.9cP. The governing equations used for present viscous flow-field are Reynolds-averaged equations of Cartesian rectangular coordinate. The tensor forms of the Reynolds-averaged continuity and momentum are as follows:

The continuity equation

$$\frac{\partial}{\partial x_j}\left(\rho u_j\right)=0 \tag{1}$$

The momentum equation

$$\frac{\partial}{\partial x_j}\left(\rho u_j u_i\right)=-\frac{\partial p}{\partial x_i}+\frac{\partial}{\partial x_j}\left(\mu_{eff}\left(\frac{\partial u_i}{\partial x_j}\right)\right) \tag{2}$$

Where

u_i is velocity component in the x_i direction,

p is static pressure,

ρ is fluid density,

μ_{eff} is viscosity ($\mu_{eff} = \mu + \mu_t$; μ is the laminar viscosity; μ_t is the turbulent viscosity).

The *k-ε* turbulence model along with wall function is employed to solve important parameters including Reynolds stress, turbulent energy flux, and fluctuating viscous work. In accordance with the *k-ε* model of Launder and Spalding [4], turbulent viscosity μ_t is represented as follows:

$$\mu_t = \frac{c_\mu \rho k^2}{\varepsilon} \tag{3}$$

Where k is the turbulent kinetic energy, ε is the dissipation rate of turbulent kinetic energy, and can be individually solved from the equations of turbulent kinetic energy and dissipation rate of turbulent kinetic energy, respectively.

3) Boundary conditions

We assume uniform velocity distribution at inlet plane. Only axial velocity is considered and its magnitude is determined to accord with the physiological flow-rate. The physiological state of heart rate 72 beats per minute and stroke volume 70 ml is simulated. The peak systole volume flow rate is 380 ml per second. The inflow rate and the aortic pressure for the three cases demonstrated in the following sections are shown in Table 1. At the outlet plane, physiological pressure at the moment is specified. No slip and no-flux boundary conditions are imposed on all solid surfaces.

TABLE I. ANALYSIS CONDITIONS

	Inflow Rate (ml/s)	Inflow Velocity (m/s)	Outlet Pressure (mmHg)
Case 1	50	0.0189	90
Case 2	175	0.0663	110
Case 3	290	0.1098	120

III. RESULTS & DISCUSSIONS

A. Flow field analysis

Flow field analyses of the aortic valve pointed out the corresponding velocity distribution trends and changes at different leaflet openings. Once the leaflets begin to open (case 1), shedding vortex appears obviously right behind the leaflets. These vortexes affect the factors on blood coagulation. When the leaflets completely open (case 3), the central flow field becomes smoother, while turbulence appears along the sinus wall. From the fluid mechanics viewpoint, differences in flow situation may lead to different problems with blood such as damage of red blood cells, damage and activation of platelets [5, 6]. Note that turbulence is the most important factor in the activation of platelets, and platelets are much easier damaged by the shear stress. In addition, platelet damage is not influenced by the maximum shear only, but also by the duration of the stress.

The flow velocity pattern varies according to the leaflet geometry of the valve. A strong turbulent flow might be related to the valve opening motion. The curved leaflet valve provides a wider opening area of the central orifice when the valves are fully opened. The increase in central orifice area is the direct reason for the decrease in velocity and smoother of fluid flow. The role of the leaflets' curvature can be positive or negative, in term of turbulence shear stress exerting on blood cells, depending on the considered position in the flow field. The formation of thrombi may be initiated by turbulent flow and other fluid dynamic factors such as increased shear stress. Furthermore, a gradient associated with turbulent flow plays the most important role in the induction of prothrombotic state.

Pressure near leaflets under surface is greater than that near the upper surface, and thereby leading flow to roll up and to form a trailing vortex. Blood flow velocity changes dramatically for the location of leaflet trailing edge and sinus wall where the jet impingement happens.

As observed from Fig. 3 and Fig. 4, bicuspid valves and prolapsed valves have severely impacted upon the flow field, velocity pattern, and magnitudes. These two heart valve diseases can cause an obstruction to flow (valve stenosis), a leakage backward (valve regurgitation), or both. As a result, these abnormalities can induce an extreme effect upon the volume of flow issuing the valve orifice and severe symptoms such as shortness of breath (dyspnea), chest pain (angina), and dizziness (near-syncope).

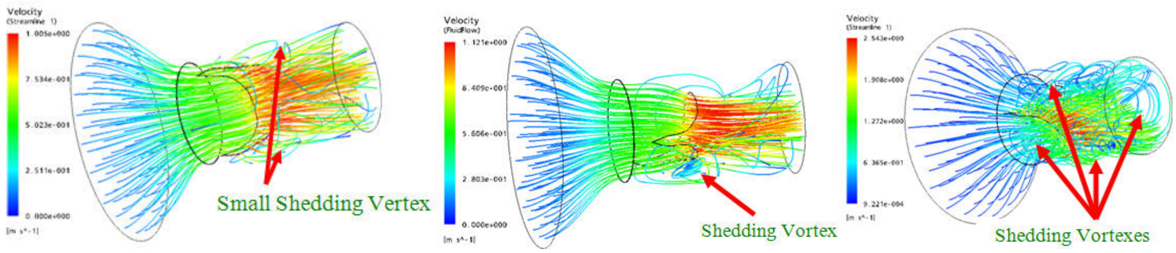

Figure 3. Flow field at case 3: (Left) Normal (Mid) Bicuspid (Right) Prolapsed

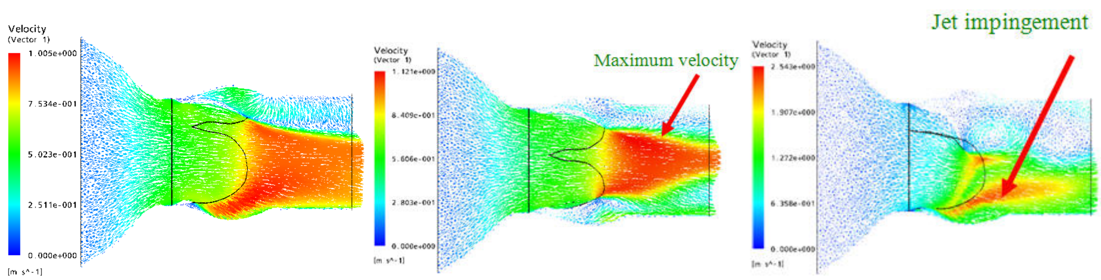

Figure 4. Velocity distribution at case 3: (Left) Normal (Mid) Bicuspid (Right) Prolapsed

B. *Wall shear stress analysis*

The maximum wall shear stress of the aortic root distributes somewhere near the jet impingement. The flow field analysis and shear stress analysis show that the maximum wall shear stress of the leaflets distributes among the leaflets trailing edges, near the centre orifice and on the endothelial lining area where the jet impingements take place. At very beginning of the systolic phase (case 1), the maximum shear stress is around 10 N/m2 and locates somewhere near endothelial lining area of jet impingements. The location of the maximum shear stress area shifts to the center of the leaflet trailing edge as leaflets further opening (case 2).

The simulation results of natural heart valve indicate that the calculated maximum wall shear stress is far from the damage limit of endothelial lining. However, this low shear stress combined with shedding vortexes appears between tip of valve leaflet and sinus could induce sclerosis in valve leaflet, thus lead to exceptional displacement of leaflet namely prolapsed disease.

In cases of bicuspid valve, the wall shear stress were observed dramatically increase (nearly 68% in case 1, 45.9% in case 2 and 30% in case 3). The location of the maximum shear stress area concentrates on the central of leaflets trailing edges and on the aortic root where two jet impingements occur.

Meanwhile, for the aortic prolapse valve, the maximum wall shear stress is extremely high in comparison with normal case; maximum value is 35.8 N/m^2. In view of the shear stress value, the possibility of endothelial lining damage which occurs at shear stress around 40 N/m^2, therefore, these shear stress high values will severely harmful effect on endothelial lining. The simulation results have illustrated the complex flow of diseased heart valve with unpredictable phenomenon.

Note that hemodynamic shear stress plays a significant role in the function and morphology of endothelial cells, and shear stress acts as a stimulus for endothelial cell migration and inhibits the adhesion of lymphocytes to endothelial cells, and thus affecting cell attachment and vascular modeling. Damage of endothelial lining by high shear stress could lead to sclerosis in valve leaflet, and results in abnormally thickened leaflet.

The shear stress has also a strong influence on damage on red cells and platelets. The enhancement of red cell damage occurs at high shear stress has been studied by several investigators [5, 6]. During high shear stress, platelet deposition and thrombosis to be a major problem. High shear stress is also associated with higher convective mass transport, stimulating platelet deposition [7].

In contrast, low wall shear stress is also considered as a main factor causes sclerosis, low wall shear stress theory. The low wall shear stresses impair mass transport between blood and the vessel/valve leaflet wall, potentially impacting on not only the uptake of nutrients and oxygen by the vessel/valve leaflet wall from the blood, but also the release of waste products and carbon dioxide into the blood from the wall. This latter effect would help explain how material components typically found in atherosclerotic lesions and plaques may build up in the vessel/valve leaflet wall with slow or stagnant flow.

The analysis results of flow field and shear stress indicate that high turbulence which concentrates near the leaflet trailing edges is the cause of generating high shear stress areas. Strong flows through the central orifice as well as the corresponding high shear stress could be observed. When the valve opens, high speed flow is generated and can cause blood cell trauma. Recently, several works have shown that the increased shear stress generated by the accelerated blood flow is also associated with the prothrombotic state. Flow field analysis have conducted, particularly in relation to turbulent stress levels, has become essential in attempts to understand the causes of platelets activation, thrombus formation, and other clinical problems.

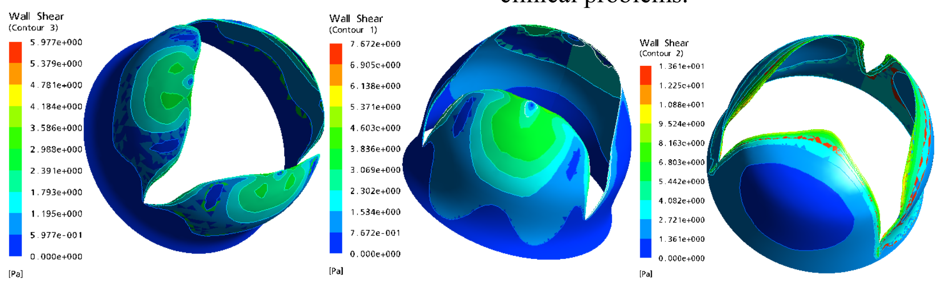

Figure 5. Shear stress distribution at case 3: (Left) Normal (Mid) Bicuspid (Right) Prolapsed

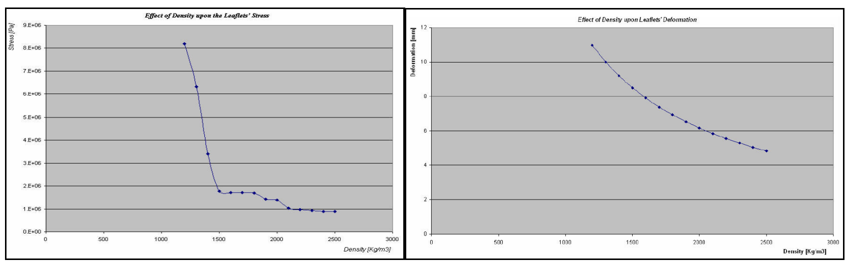

Figure 6. (Left) Density vs. Stress (Right) Density vs. deformation

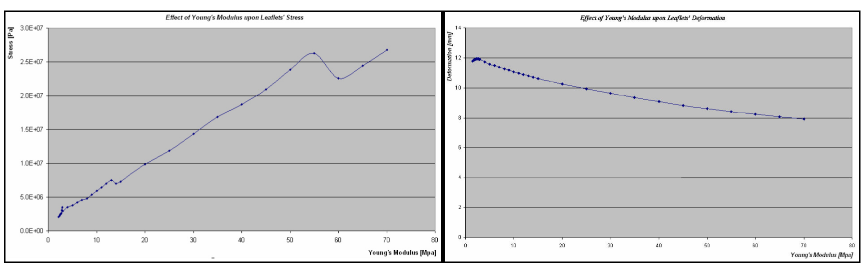

Figure 7. (Left) Young's Modulus vs. Stress (Right) Young's Modulus vs. deformation

C. *Leaflet deformation analysis*

Mechanical valves require the recipient to maintain long term systemic anti-coagulation with associated risks of bleeding episodes or thromboembolic complications, while bioprosthetic valves have lifetimes which are limited due to pannus overgrowth, calcification and primary tissue. The development of prosthetic heart valves making use of synthetic polymers has been the subject of much research effort over many years. Polyurethanes, in particular, have been applied to this problem in view of their relatively low thrombogenicity and good blood compatibility characteristics.

To determine an appropriate polyurethane material for prosthetic heart valve, relying on the flow field analysis, the authors conduct the deformation analysis which intended to examine the hydrodynamic behavior of polyurethane heart valves with a range of Young's Modulus and density. The calculated pressure from flow field analysis has been used as loading exerting on leaflet surfaces in the deformation analysis. The effects of density and Young's Modulus upon the leaflets Von Mises stress, pressure, and deformation have been shown in Fig. 6 and Fig.7.

Polyurethanes illustrate considerable versatility with aspect to their properties since their primary structure may be readily modified. The properties of polyurethanes may be varied by altering the composition or proportion of either the hard or soft segment constituents. The results indicate that very low density materials achieve good hydrodynamic function but also receive high pressure and stress, and hence may be resulting in poor durability. Meanwhile the higher densities reduce considerably the deformation of the leaflets, but the stress was affected significantly.

On the other hand, the results demonstrate that low Young's modulus assist the valve leaflets in improving hydrodynamic function, i.e. increasing the opening level, and reducing the pressure and stress exerted upon the leaflet surfaces. In contrast, the higher modulus results in a stiffer material, and hence decreases the aperture of the valve leaflets.

As the main outcome of the analysis, the material density for leaflets valve in a range of 1250 - 1500 kg/m3, and the Young's Modulus values of 1.5 MPa - 10 MPa have been selected to conduct further study. The obtained simulation results are really promising when compared with some in vitro experiment results [8] (Fig. 8).

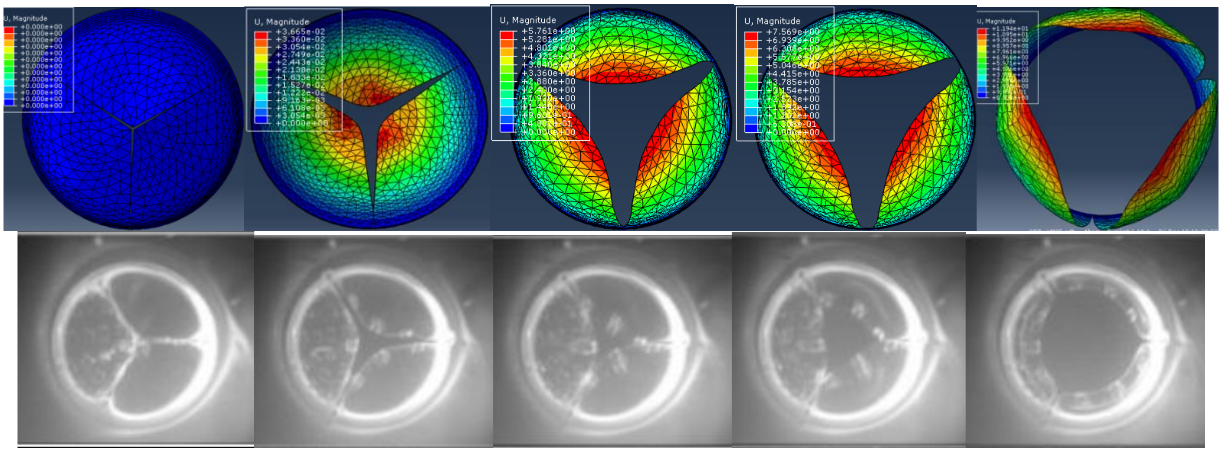

Figure 8. Heart valve leaflet deformation:(Up) Numerical (Down) Experimental [8]

IV. CONCLUSIONS

- The flow velocity pattern varies according to the leaflets geometry. A strong turbulent flow might be related to the valve opening motion. The high turbulent shear stresses occur at location of high velocity gradients and at locations immediately distal to the valve leaflets. The bicuspid valves and prolapsed valves have severely impacted upon the flow field velocity pattern, and magnitudes; they also cause an obstruction to flow. As the results, these abnormalities can induce an extreme effect upon the volume of flow issuing the valve orifice and serve symptoms such as dyspnea, angina, and dizziness.

- In cases of bicuspid and prolapsed valves, the wall shear stress increase dramatically. The location of the maximum shear stress areas concentrate on the central of leaflets trailing edges and on the aortic root where jet impingements occur. These high shear stress values have severely harmful effect and may induce damage endothelial lining which occur at shear stress around 40 N/m2.

- The interactive relation between low and high wall shear stress has been demonstrated in this study. Low wall shear stress acting on valve leaflet could lead to a valvular disease somehow. This disease in contrast induces higher shear stress, and thus, makes the disease become worse.

- The density and Young's modulus of polyurethane have a significantly effect on the hydrodynamic properties of valve leaflets. The low density PU material achieves good hydrodynamic function but also induce high stress which resulting in poor durability while high density PU resist motion of leaflets but reduces the stress significantly. In addition, it has been proved that low Young's modulus PU leaflets achieve good hydrodynamic function whereas reducing the stress exerted upon the leaflets, and vice versa for the high values. The study suggests that the density in a range 125 to 1500 kg/m3, and Young's modulus about 1.5 MPa to 10 MPa are the most appropriate option for polyurethane prosthetic valves.

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