NUMERICAL SIMULATION OF BILEAFLET HEART VALVE PROSTHESES IN AORTIC POSITION

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ABSTRACT

Modeling and simulation of heart valves had been a demanding biomechanical problem due to the variation in anatomy of heart valves, 3D anisotropic material behavior and the pulsatile physiological behavior. The application of Fluid Structure Interaction (FSI) models had become a key tool in optimizing the current existing designs of mechanical heart valves which eventually offers constructive and valuable information to the cardiologists. As such, a FSI numerical simulation was carried out to analyze the dynamic behaviors of the leaflets during its opening and closing stage with the presence of pulsatile aortic velocity blood flow and the local phenomena experienced by the blood flow throughout the simulation. Coupling was performed to reach this objective using a commercial computational fluid dynamics package in ANSYS. The prosthetic mitral valve was modeled together with the inlet and outlet of the aortic tract to simulate its actual position in the heart. The pressure and velocity distribution in 1 cardiac cycle were obtained at specified time to differentiate their distributions during systole, peak systole, diastole and peak diastole region. The results obtained were compared with the existing results from related journals. Future work on performing simulation that are closer to the aortic root geometry, and parameters that need to be taken into consideration in future research had been discussed as well to discover the flaws present in the current design and the steps that can be taken to improve them.

ABSTRAK

Pemodelan dan simulasi injap jantung telah menjadi masalah mendesak biomekanikal kerana perubahan anatomi injap jantung, kelakuan bahan tak isotrop 3D dan tingkah laku fisiologi yg berdebar-debar. Aplikasi model Interaksi antara jasad dengan bendalir (FSI) telah menjadi alat yang penting dalam mengoptimumkan reka bentuk yang sedia ada semasa injap jantung mekanikal yang akhirnya menawarkan maklumat yang konstruktif dan bernilai kepada pakar kardiologi. Oleh itu, simulasi berangka FSI telah dijalankan untuk menganalisis tingkah laku dinamik daripada risalah semasa majlis perasmian dan peringkat tutup dengan kehadiran halaju aliran darah aorta yg berdebar-debar dan fenomena tempatan yang dialami oleh aliran darah di seluruh simulasi. Gandingan dilakukan untuk mencapai objektif ini menggunakan pakej komersial dinamik bendalir pengiraan dalam ANSYS. Injap mitral prostetik telah dimodelkan bersama-sama dengan bahagian masuk dan keluar saluran aorta untuk simulasi kedudukan sebenar di tengah-tengah. Taburan tekanan dan halaju dalam 1 kitaran jantung diperolehi pada masa yang ditetapkan untuk membezakan pengagihan mereka semasa rantau diastole Sistole, Sistole puncak, diastole dan puncak. Keputusan yang diperolehi dibandingkan dengan keputusan yang sedia ada dari jurnal yang berkaitan. Kerja masa depan kepada perlaksanaan simulasi yang lebih dekat kepada geometri akar aortic, dan parameter yang perlu diambil kira dalam penyelidikan masa depan telah dibincangkan serta untuk mencari kecacatan yang hadir dalam reka bentuk semasa dan langkah-langkah yang boleh diambil untuk memperbaiki mereka.

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Title of Project Paper/ Research Report/ Dissertation / Thesis ("this work"):
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To my revered father Radhakrisnnan Jaggaiah and to my beloved mother Kohilammal Raman who has made endless sacrifice throughout their life. Thank you very much. Not a day goes by without remembering their love and sacrifices.

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LIST OF ABBREVIATIONS

BMHV	- Bileaflet mechanical heart valve
DOF	- Degree of freedom
MHV	- Mechanical heart valve
FSI	- Fluid structure interface
ALE	- Arbitrary Lagrangian Eulerian

CHAPTER 1. INTRODUCTION

1.1 OVERVIEW

Human heart is a muscular organ, which pumps blood throughout the circulatory network of arteries and veins called the cardiovascular system. Cardiac cycle includes all the events related to the blood flow or pressure that happens throughout the heart during one complete heart beat. Blood pressure and flow velocity entering the arterial tree rise and fall during each cardiac cycle, while their magnitude is largely controlled by the autonomic nervous system (Rogers, 1999). There are four chambers in the heart; namely right atrium, right ventricle, left atrium and left ventricle. A contraction activity of the heart chamber is referred as systole while diastole refers to the relaxation of the heart chamber. The heart valve allows the blood to flow in only one direction throughout the heart (Klabunde, 2012). The heart valves open and close in response to differences in the blood pressure. There are 2 main types of valves, namely atrioventricular valves and semilunar valves to allow blood to flow unidirectional through the heart. Atrioventricular valves are present between the atria and ventricles; whereas the semilunar valves are present in the arteries which are leaving the heart.

Mitral valve, which is a bileaflet valve placed in between the left atrium and left ventricle, enables the blood to be pumped from the lungs into the heart through the help of pulmonary veins. During the relaxation phase of left ventricle, the mitral valve opens as blood gushes from the left atria into the left ventricle. The mitral valve closes to prevent blood from flowing into the left atrium when the left ventricle is experiencing contraction. During this period, blood is pumped from the left ventricle to the aorta through the aortic valve, and was then sent to the whole body (Raven et. al, 2010). The mitral valve consists of 2 leaflets; anterior leaflet which has a semicircular shape and posterior leaflet which has a quadrangular shape. It functions to regulate the blood flow from the left atrium to left ventricle while avoiding regurgitation of blood and prevent any reverse backflow (Horia Muresian, 2006). It has a typical surface area of 4-6 cm^2.

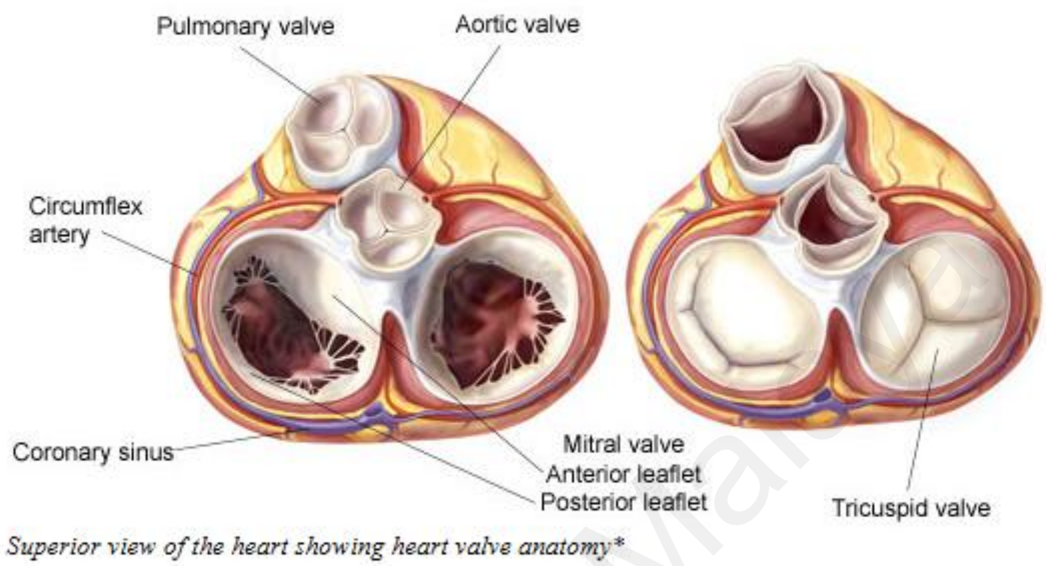


Figure 1.1 : Mitral valve location in the heart (Reproduced from (Carpentier, 2010))

A fibrous ring, known as mitral valve annulus surrounds the opening of the heart valve, in which the anterior leaflet covers a larger part; approximately two-thirds of this ring. It serves as an insertion site for the leaflet tissue (Ho, 2002). The tendon present on the mitral valve, namely chordae tendineae prevents the anterior and posterior leaflet by prolapsing into left atrium. This tendineae also prevents both valves from opening in the wrong direction (Carpentier A, 2010).

During each heartbeat, the bileaflet valve functions ideally by allowing the blood to flow through the atrium chamber to the ventricle chamber due to the increased pressure exerted by left atrium. It is estimated that about 70 – 80% of blood travels during this early filling phase of the left ventricle (Kevin Lau, 2011). The leaflets also functions in forming a rigid and closed seal in preventing the backward flow of blood into atrium. Mitral valve sustains a transvalvular pressure load of 16kPa or 120mmHg during systole when the valve is being closed (Ho, 2002). The resistance to fluid flow is low during diastole when the valve is being opened which allows ventricular filling.

Aortic valve is a tricuspid valve that is present in between left ventricle and aorta to transfer the blood to the whole body. Its three flaps functions as a one-way gate in which it only allows the oxygenated blood to flow from left ventricle to the aorta. The aortic valve opens when the pressure increases in the left ventricle during ventricular systole. This will cause the blood to exit the left ventricle into the aorta. Likewise, when the pressure in the left ventricle drops rapidly, the aortic valve closes drastically due to a decrease in the ventricular pressure.

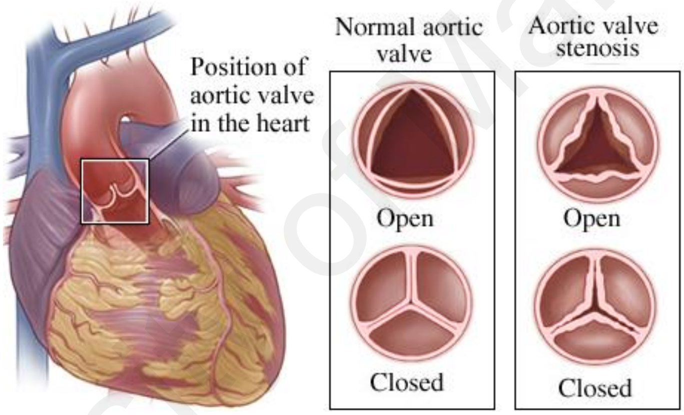


Figure 1.2 : The position of the aortic heart valve in the heart and the comparison between a normal and stenotic heart valve. Reproduced from (Healthwise Incorporated, 2009)

There are two main conditions that can lead to the failure of both aortic and mitral valve; namely valvular stenosis and valvular regurgitation (insufficiency). 95% of the heart valve replacement occurs to both mitral and aortic valves (Nebras Hussein Ghaeb, 2009). Aortic valve stenosis occurs due to the narrowing of aortic heart valve. This causes the heart valve to be inhibited from being fully opened in which this causes the obstruction of blood flow from the heart to the aorta and to the rest of body (Mayo Clinic, 2011). This obstruction causes the need for the heart to pump blood even harder in which this can eventually lead to fatigue, dizziness and weakening of the heart muscles. Mitral stenosis is characterized by the obstruction to left ventricular inflow at the level of mitral valve due to structural abnormality of the mitral valve apparatus (A.D.A.M. Medical Encyclopedia, 2010). The most common cause of mitral stenosis in adults is rheumatic fever and arthrosclerosis while congenital birth defects for children. This scenario causes atrium chamber muscles to swell as a result of build up pressure which causes blood to flow back into the lungs and eventually causes breathing difficulties.

Incomplete closure, known as incompetence or regurgitation, is the most common aortic and mitral valve disease. It happen either due to the disease of mitral and aortic root or pathology of the leaflets themselves (Novaro, 2011). Blood regurgitates into left ventricle from aorta for aortic regurgitation while blood leaks back into the pulmonary vein from left ventricle in mitral regurgitation (WebMD Medical Reference, 2010). Main causes of regurgitation include predominantly degenerative (67.8%), endocarditis (14.9%), rheumatic (12.4%), and ischemic (4.9%) (Maisano et.al, 1998). If regurgitation is left untreated, it increases the risk of left ventricular dysfunction, atrial fibrillation, pulmonary hypertension, and eventually heart failure (Verma and Mesana, 2009).

There are two types of heart valves. They are distinguished as mechanical and biological heart valve. Biological heart valves are made either from bovine pericardial tissues or porcine tissues which are treated with gluteraldehyde for preserving and antigenic proteins removal purposes whereas mechanical heart valves are made from synthetic biomaterials such as metals, polymers and ceramics (Nebras and Al-Timemy, 2009). Mechanical heart valves lasts for almost 20 years on average while biological heart valves need to be changed every 5 years on an average basis. The downside of mechanical heart valves is that the patient needs to consume anticoagulation medication such as warfarin, Coumadin and panwarfarin lifetime to avoid blood from clotting on the valve which can cause the valve to malfunction (Benjamin, 2003).

The first heart valve replacement was invented by Harken in aortic position and Starr in mitral position in 1960 which were made of titanium and they lasted for 20 to 25 years. There are three main types and designs invented for mechanical heart valve; namely caged ball valve, single leaflet (tilting disc) and the bileaflet valve. The bileaflet valve was first invented and introduced in 1978, with the improved hemodynamic and structural ability; namely the St Jude Medical (SJM) bileaflet mechanical valve (Pierrakos, 2002). SJM bileaflet valves have set a gold standard with the valve housing made from graphite with pyrolytic carbon while the leaflets impregnated with tungsten to allow valve visualization. Leaflets act as an obstruction to the blood flow through the heart valve and high velocity blood jets through the gap between the leaflets when the valve is fully opened. Also, it cases elevated shear stresses which may cause red blood cell to be damaged or reduced platelet activation. Arterial walls and the heart muscle are compliant to each other and play an important role in the process of opening and closing of heart valves (Amanifard, 2011). The study of the fluid mechanics of prosthetic heart valves became an important research field in bioengineering since their introduction back in 1960 (DeWall, Qasim and Carr, 2000).

1.2 RESEARCH OBJECTIVES

There are a few crucial objectives set in this research. The first objectives set for this study is to model and develop a 3-dimensional fluid dynamics model of the bileaflet prosthetic heart valve in the aortic position. Besides, the flow (e.g. vortices) and transvalvular characteristics (e.g. pressure gradients and effective geometric orifice areas) across the valve are to be investigated as well once the modeling is done. The velocity distribution for the blood flow using the mitral heart valve will be explored and studied as well on inclined angles from 10° which is near opening position to 85° which is near to the bileaflet closing position. The flow and simulation results will be studied and examined through 3 main approaches; which are the current theoretical method, computer simulation through existing computational fluid dynamics simulation software and real experimental measurement data which are obtained from patients who had undergone the mitral heart valve replacement. Besides, the deformation of leaflets will be predicted as well in one complete cardiac cycle. The key parameters that will be taken into account is the hydrodynamic of the mitral heart valve such as effective orifice area (EOA), energy lost with forward flow, pressure drop of forward blood flow, and the volume of regurgitated blood experienced when the valve is fully closed in a complete flow cycle.

1.3 REPORT OUTLINE

This thesis is organized as follows. Chapter one is an introductory chapter, which begins with a brief introduction on history of artificial heart valves, and the anatomy of the heart valves by itself. Also, the failures of heart valve had been explained in detail in which this leads to the objectives of this research followed by the significance of the study in this report.

Chapter two reviews the relevant literature, providing the ground work for the methodology of this work to study the efforts and previous methods and trials that had been worked out by researchers to design the bileaflet heart valve in numerical simulation manner.

Chapter three, “Methodology”, presents the details on the design of the mechanical heart valve, the techniques that had been carried out using ANSYS to obtain the results as expected in the objectives.

Chapter four, “Results”, presents the results that were obtained from ANSYS software while the “Discussion” in Chapter 5 completes the section by reasoning out the results that had been acquired in this research.

Chapter six, “Conclusion and Recommendation”, concludes by summarizing what the research work in this thesis has achieved. It gives a precise account of the extent to which the objectives have been accomplished, and elaborates on further ideas that are suggested to be carried out as an extension of this work.

CHAPTER 2. LITERATURE REVIEW

2.1 INTRODUCTION

Computational Fluid Dynamics (CFD) is emerging as one of the most promising instrument in which it enables one to have insights of hemodynamic of prosthetic MHV. Numerical simulation and analysis had become a significant technique used for biomechanics research as the handling data algorithms become more refined and provides influential computational resources. Moreover, numerical simulation offers an alternative way by allowing various realistic geometrics modeling, detailed information for optimization of valve design besides a major attainment in solving the dynamics of the intended design with high degree of resolution in both space and time (Sebastiaan et.al, 2010).

2.2 BACKGROUND

Physical modeling can be a very accurate and effective method in testing for research purposes; however, this experimental simulation consumes much longer time and is a very expensive method due to numerous tests that are conducted to get the results during the research and development phase. Besides, numerical simulation itself can provide the exact useful, constructive responses (Dasi, 2009), deeper insight on the blood flow and shear stress regions; and particularly efficient for sensitivity studies (Verdonck, 2002) as the experimental ones to surgeons on deciding on the best implantation strategy with a much lesser costs to the manufacturers (Annerel et.al, 2011). Interaction of blood-leaflet interplay through the application of Fluid Structure Interaction (FSI) models moves into the direction of providing more in-depth insight into the MHVs hemodynamics, including the valve opening and closure assessment. Prosthetic valves have a complex geometry and the leaflets shape and motion progress dynamically in response to the instantaneous flow settings. Also, the pulsatile nature of blood flow together with the resulting FSI causes an inclination to a very complex, borderline turbulent flow and highly unsteady which are being differentiated according to their 3D separation, reversal in flow, shedding and vortex formation (Yoganathan, 2005). Non-Newtonian effects are essential when the flow within the valve components is being analyzed, such as at the hinge pivot region of the heart valve (Sotiropoulos, 2004). Blood vessels exhibit non-linear properties when distended in order that the elastic stiffness increases with the degree of loading. It is considered to be a challenge for CFD to investigate on the mechanics of BMHV due to the large rotations that are encompassed by the leaflets together with the fully coupled FSI features which are needed to represent the complete physical phenomenon of BMHV in an aortic position (Palmieri, 2007), the complex aortic and ventricular anatomies and the complexity in geometry of a prosthetic BMHV which is joined with the dynamic nonlinear nature between pulsatile blood flow (Yoganathan, 2005) . BHMV had been widely used and still remains as the most widely implanted valve design for over two decades. BMHV are well known to have high durability and superior hemodynamic performance which is inclusive of higher effective orifice area (EOA) which generates smaller drop observed in transvalvular pressure, pattern of central flow and low pressure drop (Sotiropoulos, 2004). The better hemodynamic property is due to the fact that their hinged semi-lunatic leaflets can open almost parallel (85°) with the mainstream flow which can minimize the blood flow resistance (Qi, 2003). However, BHMV implants carry some significant risks that are unavoidable despite the continuous design improvements and widespread clinical use. Hemolysis which occurs at the gap between the leaflets due to high velocity jets, thromboembolic events and water hammer effects during the opening and closing of leaflets are major concerns that need to be solved in CFD (Tae-Hyub, 2009).

The mechanical properties of mitral valve tissues are assumed to be non-linear and elastic (Emiliano et. al, 2010). The mechanical response was described by means of proper strain energy potentials in which leaflets anisotropic response were accounted for by means of a Fung-like strain energy potential (Prot, 2009). Analyzing the dynamics of blood flow is crucial in computational fluid dynamics to enable researchers to investigate the complex flow patterns within the valve and flow downstream of heart valves (Kim, 2001). The five main performance characteristics that are expected in a bileaflet MHV are thromboresistance, which is the resistance to form blood clot, hemodynamics; the efficient blood flow through the valve sufficient to meet body needs, durability on how long the valve lasts and the quality of life. Blood is modeled as a homogeneous, incompressible Newtonian fluid (5-10), with a specific mass, ρ of the value of $1.06 kg/m^3$ and a constant dynamic viscosity, μ of 3.5×10^{-3} Pa.s. Among the parameters set in measuring the dynamics of bileaflet MHV are: condition of heart rate: 70 beats/minute; systole to diastole ratio : 0.5; cardiac output of 5.01/minute and the maximum tip closing velocity of leaflet set to be between 1.40 and 2.03m/s (Qi, 2003). The gravitational effects are being neglected in the study. The blood flow is assumed to be laminar and time dependent, in which the flow motion through the fully open bileaflet mechanical heart valve model up to peak systole has been described by the principle of momentum conversation, expressed by Navier-Stokes equations and by continuity equation, representing the principle of mass conversation (Grigioni, 2003). The Cartesian coordinates systems are shown below for a transient analysis in the flow motion governing equations:

(x- Momentum):

$$\rho\left(\frac{\partial u}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial u}{\partial y}\right) = -\frac{\partial p}{\partial y} + \mu\nabla^2 v$$

(y – Momentum):

$$\rho\left(\frac{\partial v}{\partial t}+ u\frac{\partial v}{\partial x}+ v\frac{\partial u}{\partial y}\right) = -\frac{\partial p}{\partial y}+ \mu\nabla^2 u$$

(mass):

$$\frac{\partial v}{\partial x}+ \frac{\partial u}{\partial y} = 0$$

whereby u,v, and ρ represents the 3 Cartesian components of initial velocity, final velocity and pressure in each point of the blood (fluid) domain and;

$$\nabla^2 = \frac{\partial^2}{\partial x^2} + \frac{\partial^2}{\partial y^2}$$

The opening and closing cyclic performance of bileaflet mechanical valve need to be studied to obtain the blood flow and the leaflet behavior in a cardiac cycle. In closed state, the leaflets are in mutual contact and a large transvalvular pressure gradient will occur and the tissue of which the heart valves are built has an anisotropic and heterogeneous structure (Loon, Anderson and Vosse, 2006). During the opening phase of the leaflets, the left ventricle contracts at the start of systole which causes an increase in pressure in the left ventricle. The blood flow starts to accelerate due to the resulting positive transvalvular gradient and this eventually induced the valve opening. In an ideal BMHV, the leaflets initially open with a huge increase in angular acceleration but they tend to align with the axial flow and their local linear velocity. This action causes a decrease in both angular velocity and angular acceleration due to pressure moment. The blood flow starts to decelerate at peak systole; when the leaflets are in their fully opened position. The bulk of the flow remains laminar while the formation of small scale turbulence is prevented. During the closing phase, the leaflets start to close and regurgitation occurs as the leaflets need to rotate in a large angle. The angular velocity increases until the leaflets are fully closed. The deceleration of the blood flow is governed by turbulent vortices and small-scale eddies. Regurgitation occurs when the leaflets are in fully closed position due to the small gaps that are present between the valve casing and leaflets and through the gap present in between leaflets (Tullio, Balaras and Verzicoss, 2009).

It had also been noted that most of the previous researchers had been performing the investigation on the bileaflet MHV using CFD regarding the flow field (Kim, 2001). However, there is a key concern on their studies as the researchers seem to have had ignored the effects from the motion of the leaflet valve although this bileaflet MHV performs opening and closing process in a cyclic manner due to the pulsatile flow of the heart beat. Some had approximated this motion into a very much simplified assumptions in which it causes them not to depict clearly the flow field as well as the complete cardiac cyclic process on the leaflet motion((Kim, 2001) and (Sotiropoulos, 2004)). Among the deficiency in Yoganathan's algorithm is that it is currently inadequate and restricted to Reynolds number of two or less simulated flows (Pierrakos, 2002). A FSI model had been presented by (Rui, 2004) to forecast the motion of the leaflets by exerting forces on them due to the flow dynamics in which the 2 dimensional model imitates the closing and bounce back phases of the leaflet during valve closure. The interaction of blood flow and the elastic deformation of the valve leaflets determine the dynamics of the BMHV. As such, FSI analysis for dynamics of bileaflet MHV had been taken into consideration the invented domain method associated with the finite element discretization employing a fixed mesh of motion between the fluid (blood) and both the anterior and posterior leaflets in order to obtain a reliable and accurate simulation results (Rui, 2004). The FSI advancement had been used vastly in previous and current works in which they can be generally classified into two main grid methods; which are moving-grid methods based on Arbitrary – Lagrangian – Eulerian (ALE) approach (Tae-Hyub, 2009), and fixed – grid methods such as particle image velocimetry (PIV) (Manning, Fontaine, Deutsch and Tarbell, 2003) and immersed boundary method (IBM). ALE is the most popular technique in which it incorporates the grid velocities in the momentum and continuity equation in the fluid domain (Redaeli, 2004). Arbitrary Lagrangian is used for the solid (leaflets) with a hyperelastic Neo-Hookean model. The fluid (blood flow) is described by the unsteady Navier-Stokes equation in an Eulerian manner. Meshes are overlapping, but non-conforming and are generated for the solid and fluid domain in which the velocities are coupled along the solid boundary using Lagrange multiplier. The interpolation of the velocities requires the need of coupling due to the domains non-conformities. ALE method, in which it uses mesh which had been conformed to the moving boundary, is one of the most commonly used for FSI problems as this method provide the most accurate results and strong coupling although the mesh moving step could be quite challenging for complicated 3D problems (Redaeli, 2004). This method works very well as long translations; deformations and/or rotation of the solid remain within certain allowable limit. The problems faced in ALE method is that ALE is only applicable for those involved in relatively small structural displacements, elements become misshaped when the limit of rotation, deformation or translation are being violated (Loon, Anderson and Vosse, 2006). Besides, the two main drawbacks of ALE is that the elevated computational price due to mesh handling and local numerical transmission which will take place in regions where remeshing is required (Pierrakos, 2002). Also, the moving step of mesh could be quite expensive and challenging for complicated 3D designs (Tae-Hyub, 2009).

FSI algorithms were developed to predict the leaflets motion. This has a direct consequence of the momentum applied by the blood flow on the leaflets (Nobili et al., 2008). FSI analysis for dynamics of tissue valves are reported using the fictitious domain method allied with the finite element discretization occupying a fixed mesh while the leaflets motions are being integrated (Rui, 2004). FSI focuses on specific phase of the valve dynamics and simplification in the geometry is introduced due to the complexity of the simulation and the consideration on computational cost. Cheng et. al, 2004 had presented a 3D FSI model of transient flow analysis on a BMHV in mitral position during the closing phase. Cheng et.al, 2004 had also performed a two-dimensional (2D) model of BMHV to simulate the rebound phase and closure of leaflets to analyze on the wall shear stress and negative pressure transients at the edge of the leaflets. Alberto Redaeli et.al, 2004 had conducted an analysis of the opening phase of BMHV under low flow rate and validated their results by performing experimental simulation on the leaflets motion. Dumont K. et al., 2005 had presented the simulation of the whole cardiac cycle in laminar condition by assuming a symmetry flow on both sides of the heart. A 2D studies of MHV had been presented by (Cheng, 2004) and (Pelliccioni, 2007) in laminar flow by applying the lattice-Boltzmann method. In this study, two different approaches had been used in which the first is to focus only on the peak and near peack systole phase of the cardiac cycle by neglecting the relationship between blood and leaflets while the latter approach is to consider the interaction of both the solid and fluid domain (Nobili et al., 2008). The numerical simulation of a MHV placed in a physiological aortic root shaped model was introduced by (Grigioni et al., 2003) whereby the indications of the Valsava sinus role on the fluid dynamic downstream at midsystole is taken into consideration (Espino, 2006). Ge et al. (2003) emphasized the simulation on a high Reynolds number setting of a complex hemodynamis of a BMHV by introducing the need of a high grid densities to fully characterize the local aspects of the flow field downstream (Nobili et al., 2008). Numerous investigation that are conducted on the interaction of blood-leaflets through the application of FSI provides a more comprehensive insight on the opening and the closure of valve leaflets.

Most of the researchers who had contributed in studying this area had been providing only macroscopic information,where the scope of assesment of valve performances had been basically based on the measurements such as energy loss, regurgitation and transvalvular pressure drop. Although they offer constructive and functional results to compare among other valve designs available in market, local phenomena such as flow recirculation, disturbed flow patterns, vortices development and stagnation points are left out in their research and they will be described in this paper.

Analysis of the blood flow field around the valve will help in identifying the areas of flow disturbance and areas of stasis close to the valve in which the obtained information from this research can be used in improving the current bileaflet heart valve designs. The FSI technique had always been implemented though the usage of journal files and user-defined functions whereby an external FSI code is written in C++ to perform the transient calculation using only ANSYS FLUENT by previous researchers.

In this study, a FSI model will be developed in which the interaction between the physiological flow of blood with the moving leaflets in BMHV is simulated using FSI. ALE approach will be used to simulate the physical interactions of BMHV, while dynamic mesh method in FLUENT together with Coupling will be used to investigate the interaction between the moving leaflets and the blood flow in an aortic tract in a cardiac cycle.

CHAPTER 3. METHODOLOGY

3.1 INTRODUCTION

This chapter contains the methodology used for this study. In performing a numerical simulation for bileaflet heart valve prosthesis, there are three main phases that need to be solved; the structural (moving leaflets) stage, the flow (blood) stage, and finally the interface between the moving leaflets and the blood flow at the FSI.

3.2 RESEARCH METHODOLOGY

To start with the structural stage, the design of the MHV in aortic position must first be modeled using Design Modeler in ANSYS. The model consists of a 19mm mitral BMHV of St. Jude Medical, inlet and outlet representing the flow tract of aorta. The BMHV is designed as a rigid casing which has an outer diameter of 19mm and inner orifice diameter of 14.7 mm; consisting of two separate rigid leaflets which can rotate independently around their hinge axes. This valve has two degrees of freedom (DOF) as the leaflet position at each specific time during cardiac cycle is solely determined by its opening angle. The leaflets are designed to be in a fully closed position which lies on the x-z plane where the y coordinates indicate the distance from the leaflet surface on the atrial side of the valve. They are 0.50 mm thick in which they are enabled to rotate between 10° and 85 and are made from pyrolytic carbon which has a density of 2116 kg/m^3. The valve geometry is simplified whereby no hinges are modeled for FSI calculation purposes (Chandran, 1998). Moreover, the leaflet size had been reduced to 98% of its original measurement to avoid problems on contact region between the valve housing and the moving leaflets and at the same time preserving an acceptable level of skewness (Dumont, 2005) as shown in Figure 3.1 .

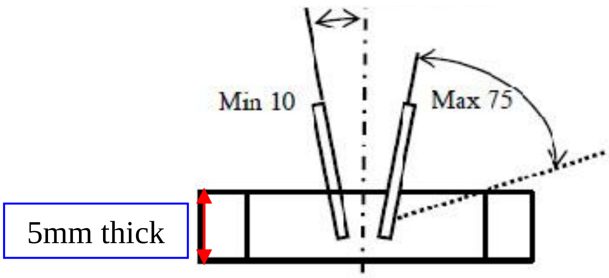


Figure 3.1 : Bileaflet aortic valve model (Reproduced from (Nebras and Al-Timemy, 2009))

The inlet and outlet of the aortic tract has an expanding tube with 1.3 as the diameter ratio. The inlet of the aortic tract is 22mm in diameter. This results to an outlet diameter of 28.6mm (Nebras and Al-Timemy, 2009). Both tracts are set to be 50mm in length as shown in Figure 3.2.

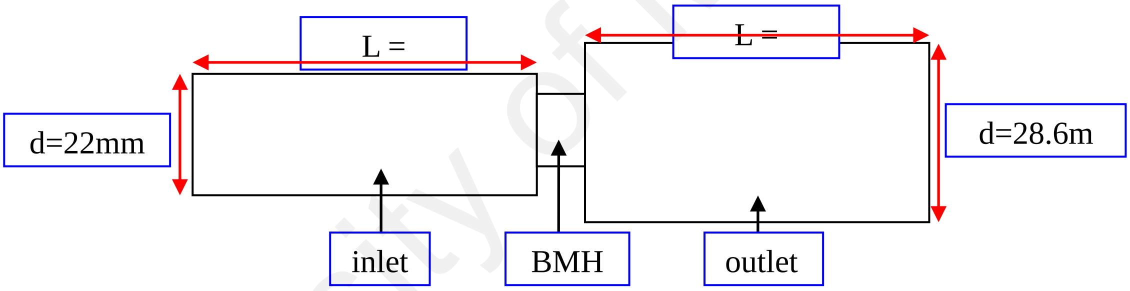


Figure 3.2 : Model of the aortic tract

The design is remeshed in order to maintain the dimensions of the cells to be as near as possible with the original value and collectively to minimize and preserve the level of skewness. On top of that, a high grid resolution is chosen at areas close to the leaflet surface to enable more accurate flow calculation to be conducted during the simulation. However, remeshing refinement is done only at the region of interest and is avoided at the zones which are not taking part in leaflet motions to save the computational time and cost (Chen, 2006).

In the transient structural phase, the leaflets are set to be fluid structure interface and their rotation velocity is set as 1.48 rad/s for both leaflets but in opposite directions. As for the hinge, a fixed support is set at the edge of the leaflet. Directional deformation is expected to be solved in the solution output. Once the results are obtained, they are transferred to the setup of Fluent. For the blood flow, the model has transient time and the solver type used is pressure based with absolute velocity. A direct numerical simulation is used to solve the fluid phase where no turbulence or averaging modeling is applied. The determination of the whether the flow is laminar or turbulent is based on the Reynolds number. Laminar flow occurs at low Reynolds number. Blood is a suspension of particles and not entirely consists of fluid. Blood viscosity decreases when the diameter of blood vessel decreases as the cells tend to move to the central part of blood vessel. Viscoelastic effects are also affected by the red blood cell elasticity in which these effects are crucial in small vessels. Only large vessel is considered in this study and as such the fluid is assumed to be Newtonian (Espino, 2006).

The fluid; blood as such is assumed to be laminar, incompressible, and Newtonian with a density of 1056 kg/m3 and viscosity of 0.0035 kg/ms, representing human blood properties at 37°C (Chandran, 1998). This simulation uses mass conservation and time-dependent Navier-Stokes equation in order to incorporate a moving cell technique of a moving cell to simulate the leaflet motion in one cardiac cycle.

$$\frac{\partial \rho}{\partial t} + \frac{\partial}{\partial x_j}\left(\rho u_j\right) = 0$$

An ALE mesh is used to allow the performance of FSI simulation for the moving leaflets to determine the fluid dynamics (blood) and the structural (leaflets) deformations (Espino, 2006). The aortic walls are assumed to be rigid with no slip conditions are set on the blood wall (zero velocity). At inlet, the inflow boundary conditions are applied a velocity profile for the axial velocity and zero transverse velocity to simulate the opening of valve during diastole and the closing of the valve during systole. An inflow velocity is applied at the atrium during diastole of the heart as shown in Figure 3.3 and a pressure condition is applied at the apex of the heart to apply the relevant ventricular pressures. During systole activities, the outflow boundary condition is set at the aortic outlet with the pressure is set to be 0mm Hg and ventricular pressure is applied to the apex of the heart.

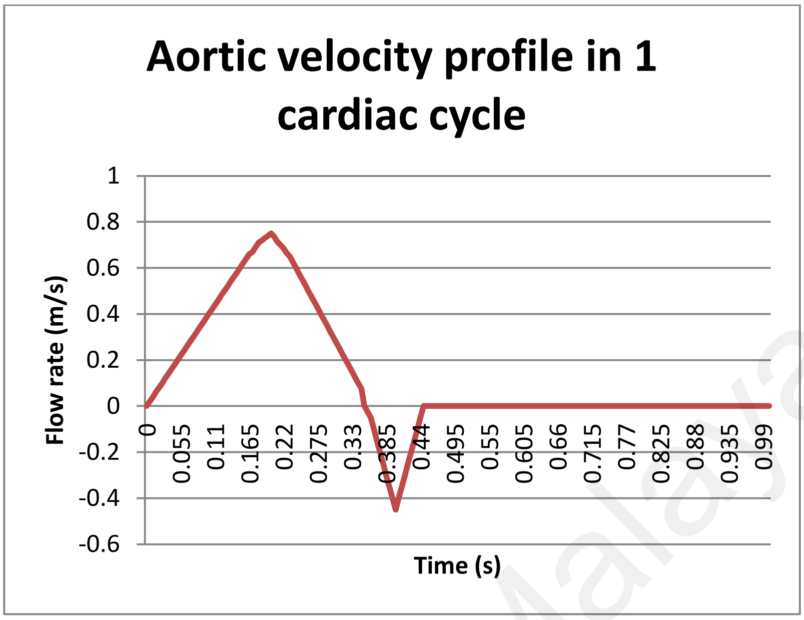


Figure 3.3 : Aortic velocity profile in 1 complete cardiac cycle

In modeling the interaction between the moving leaflets and the blood flow, Lagrange multipliers are used to exert forces on the deforming leaflets due to the blood flow and velocity constraints are applied to the blood flow on the boundary of the leaflet and blood in order to ensure the synchronous activity in the velocity of the moving leaflet and blood (Cheng, 2004 and Hart et al., 2003). The blood flow is assumed to be at rest with the valve present in the closed position at the beginning of the calculation when time is 0 s. The cardiac cycle which completes at 0.8 s has a physical time step of 0.005 ms with 160 time steps. In order to remove the dependencies of the opening phase from initial condition and the invariance occurs in the guarantee cycle, three cardiac cycles are simulated and this thesis presents the results from the third cycle. As for the last stage, where the coupling takes place between the moving leaflets and blood flow, continuous interactions between the heart valve leaflet rigidity and blood pressure becomes relevant in FSI. Information on the position of contacting boundaries along the leaflet boundary is obtained through the integration of coupling variables (Hisham, 2011). The valve closure is simulated using the contact equations between the boundaries of two mitral leaflets with the contact surface of the inner orifice diameter of BMHV and the contacting boundaries are separated into small sections consisting of nodes at each end of small fragment where the distance between the nodes of one or whole boundaries with the leaflets are determined by equation below:

$$F = \begin{cases} Pe^{-gk/P}, & g > 0 \\ P - gk, & g \leq 0 \end{cases}$$

Where P denotes the applied blood flow pressure, k is the contact stiffness; g is the distance between two contact points and F is the resulting contact force attained at each node (Espino, 2006)

The leaflet rotation in accordance with space conservation law is:

$$\frac{d^2\theta}{dt^2} = \frac{M}{I_o}$$

where θ(t) is the leaflet opening angle and the indication of the leaflet position at any instant *t; M* is the total momentum acted on the leaflet from the external forces inducing the leaflet motion and I_o is the moment of inertia of the leaflet about the pivot. The simulation of FSI begins at early systole when the valve is in the fully closed position. At this state, the left ventricular pressure will be higher than the aortic valve. Flowchart in Figure 6, depicting the iterative FSI algorithm is created to enable the leaflet position in each time step is in equilibrium position with the force generated by the blood flow pressure (Palmieri, 2007). As the velocity of the leaflet varies during the opening and closing of valve, the flow chart shown in Figure 3.4 calculates the time step size automatically. This enables the overall angular displacement of the leaflet to be compatible with the remeshing requirements in each time step (Palmieri, 2007). Before an iteration of FSI calculation is complete, the leaflet opening angle, θ is checked at each timestep whether the maximum opening angle, θ_{max} had been reached. Once the maximum angle is achieved, steady iterations are executed until the pressure moment reverses causing a restart of FSI iterations in the closure phase (Hashim, 2010). The time increment Δt is controlled in the range of 0.005s to 1s to associate with the variance of the leaflet opening angle. The time increment, $\Delta t_{n+1}$for the next time step is based on the differences obtained on angle calculated with the previous time step $\Delta\theta_n = \theta_n - \theta_{n-1}$and this permits one to keep the deviation of the open angle to be less than the consecutive time steps as summarized in the flow chart of Figure 3.4.

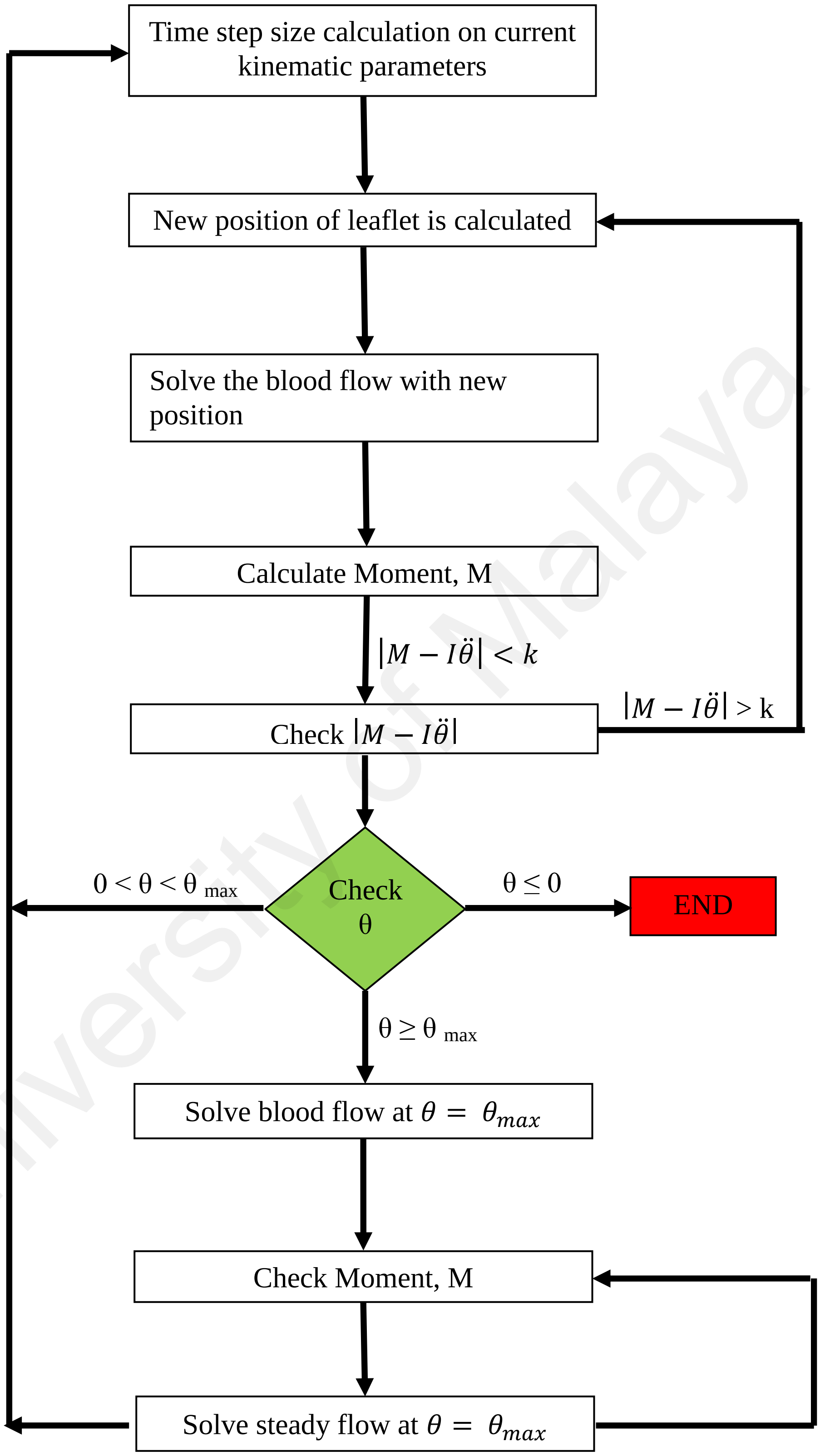


Figure 3.4 : FSI Implementation

CHAPTER 4. RESULTS

4.1 DESIGN OF THE AORTIC TRACT WITH THE LEAFLETS

Figure 4.1 shows the bileaflet MHV which had been designed using SJM model as the reference (standard A101, size 19 mm) together with the aortic tract

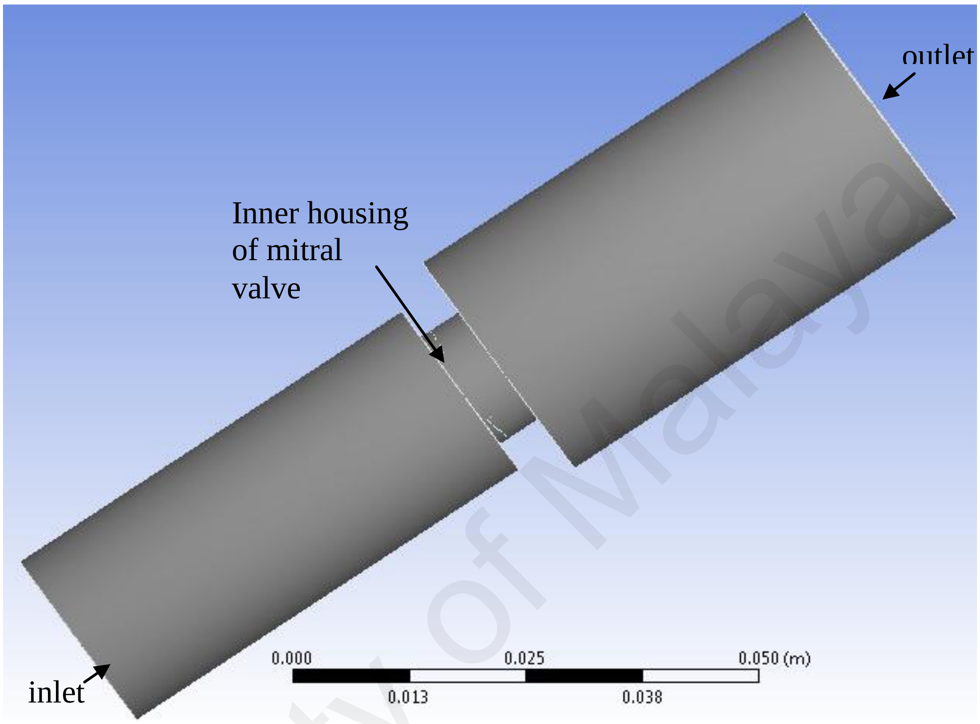


Figure 4.1 : Designed BMH valve with aortic tract

Figure 4.2 shows the initial position of the leaflet.

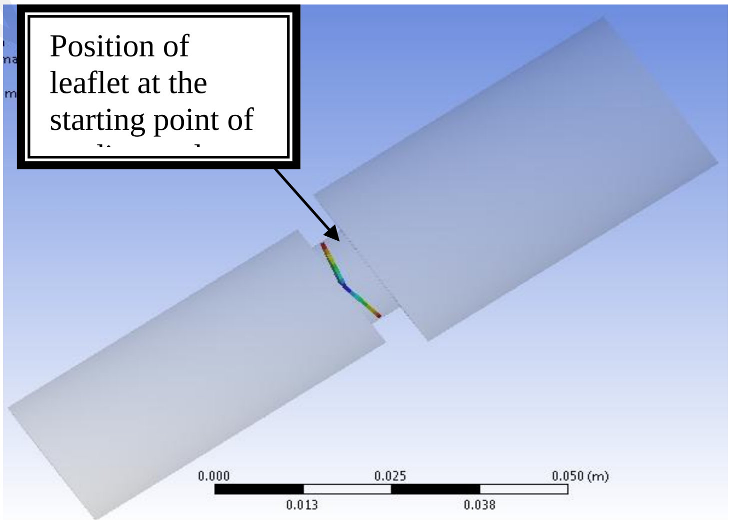


Figure 4.2 : Leaflets initial position

4.2 DESIGN OF THE SIMULATION

Figure 4.3 depicts the simulation design that had taken place to perform FSI simulation between the blood flow and valve leaflet deformation against time without writing C++ codes.

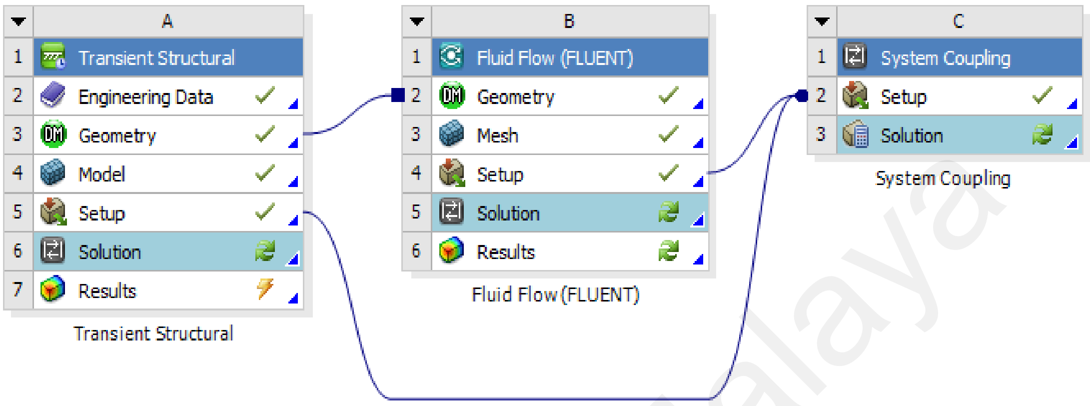
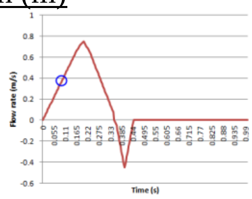
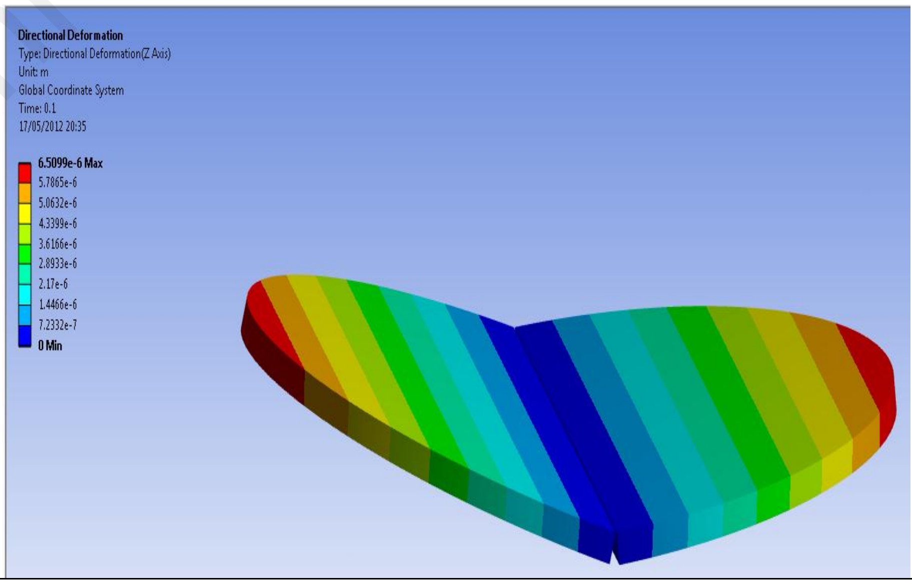


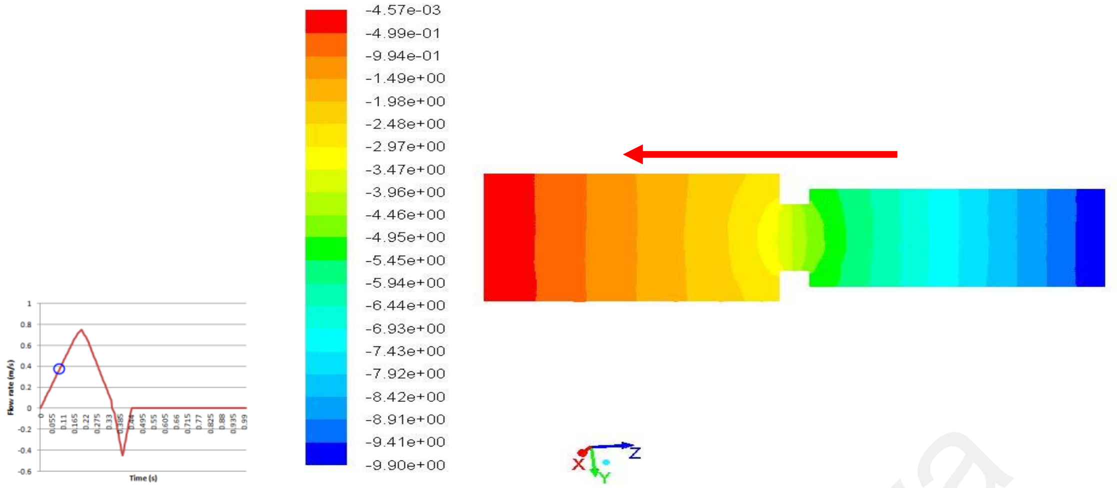
Figure 4.3 : Simulation design

Table 4.1 below shows the displacement that is observed on the leaflets with respect to time together with the variation in the velocity and pressure in 1 cardiac cycle.

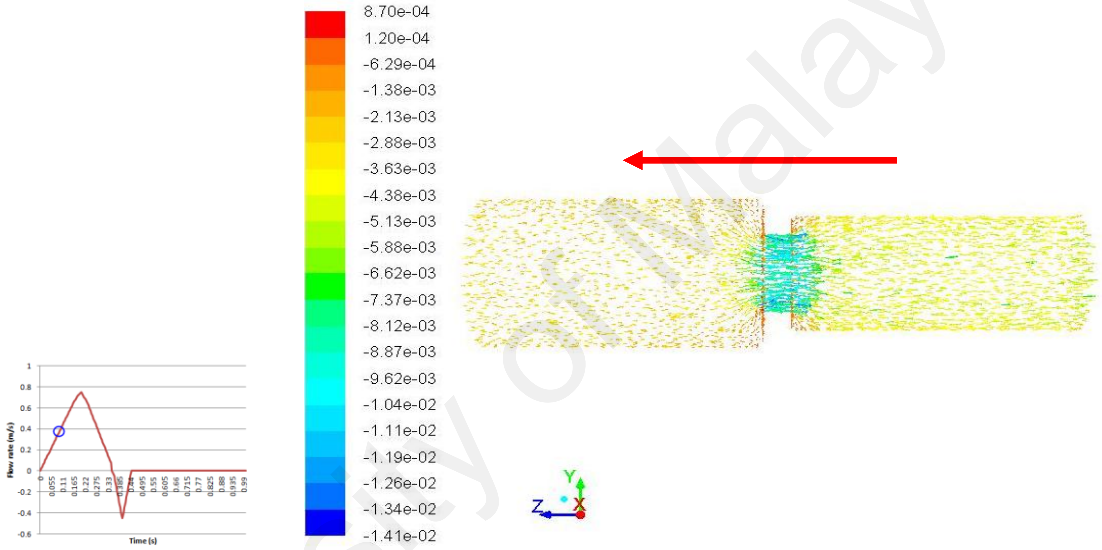
Table 4.1 : Variation of leaflets, velocity and pressure with time

Time(t) = 0.1s
Leaflet movement and deformation (m)  

Static Pressure of the blood flow (Pa)

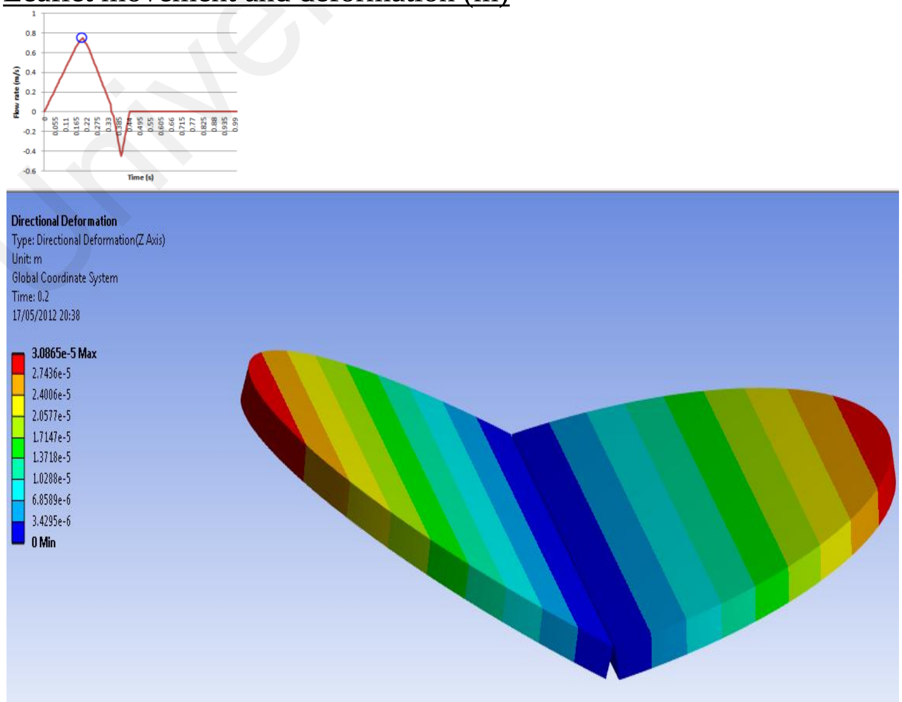


z-Velocity of the blood flow (m/s)

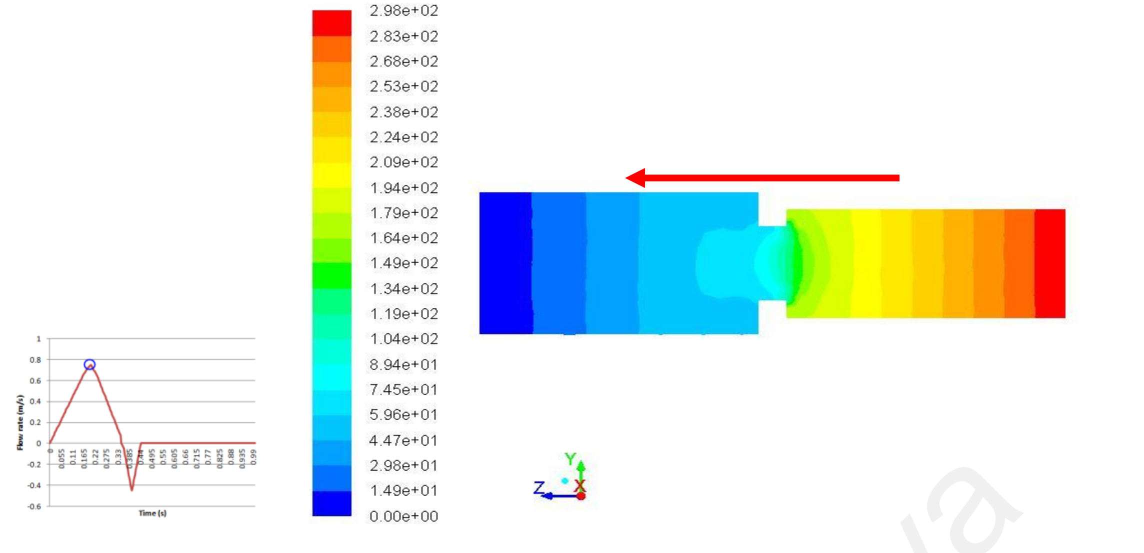


Time(t) = 0.2s

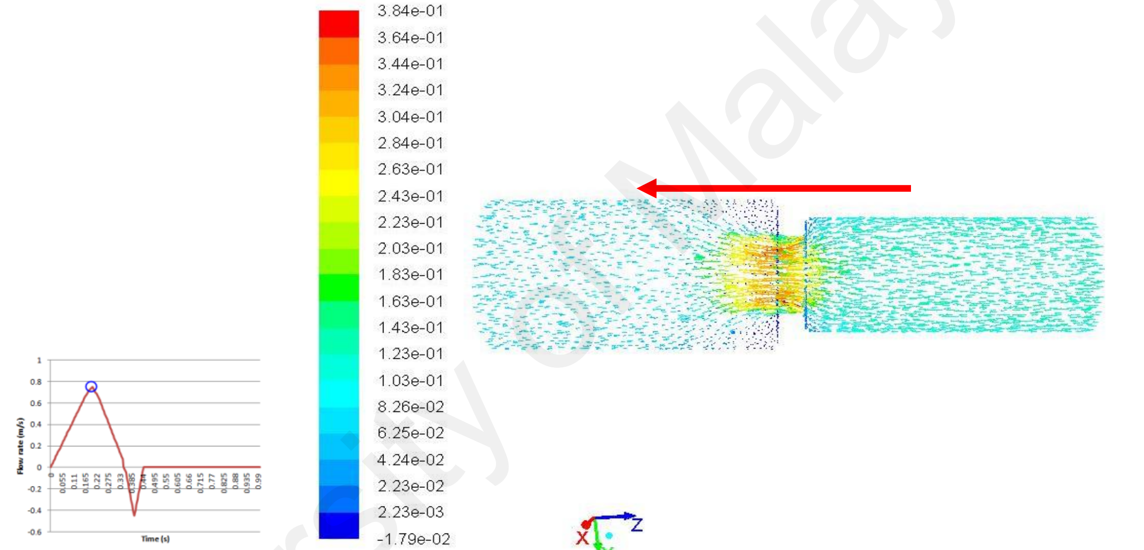
Leaflet movement and deformation (m)



Static Pressure of the blood flow (Pa)

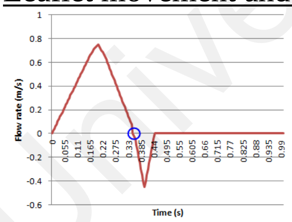


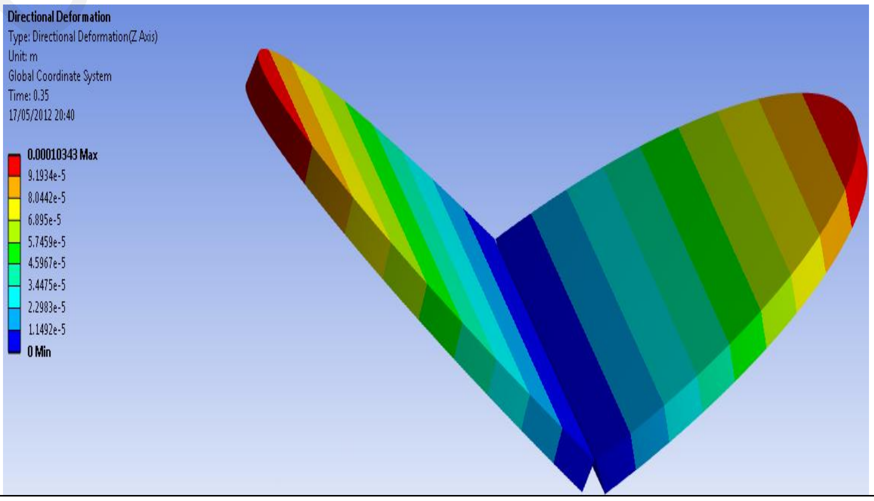
z-Velocity of the blood flow (m/s)



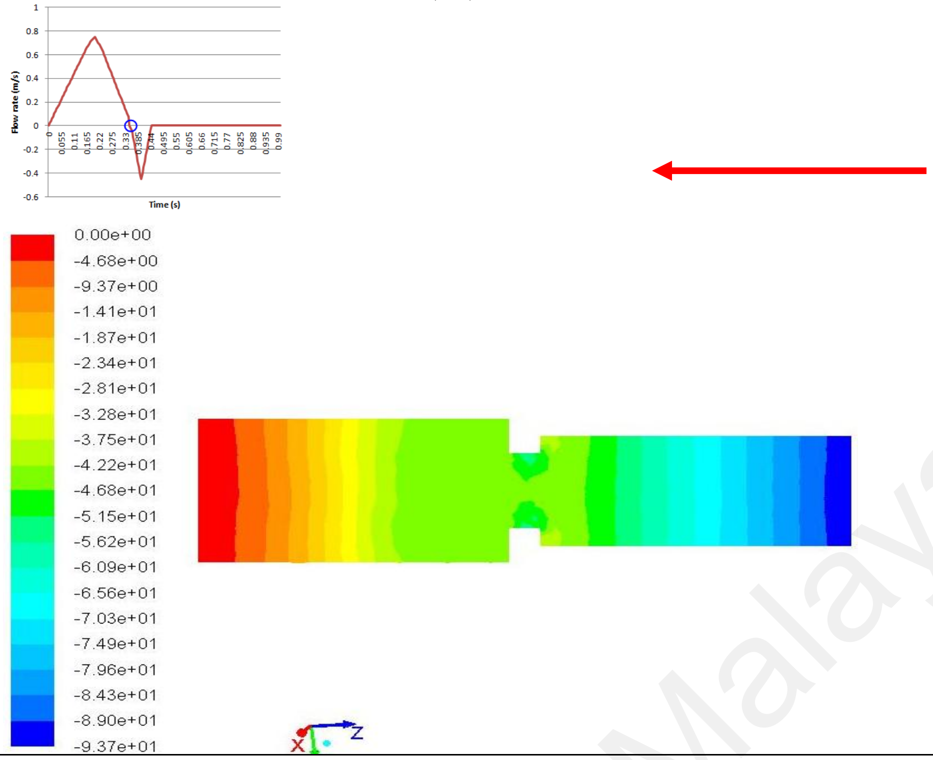
Time(t) = 0.35s

Leaflet movement and deformation (m)

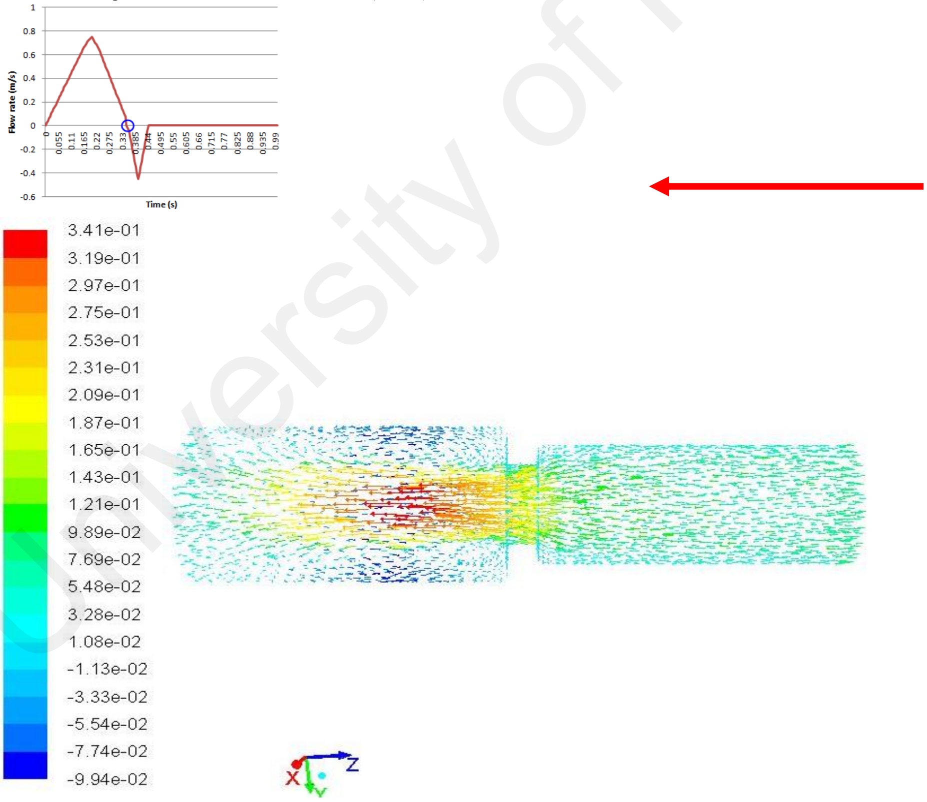




Static Pressure of the blood flow (Pa)

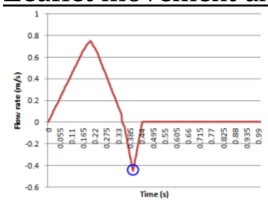


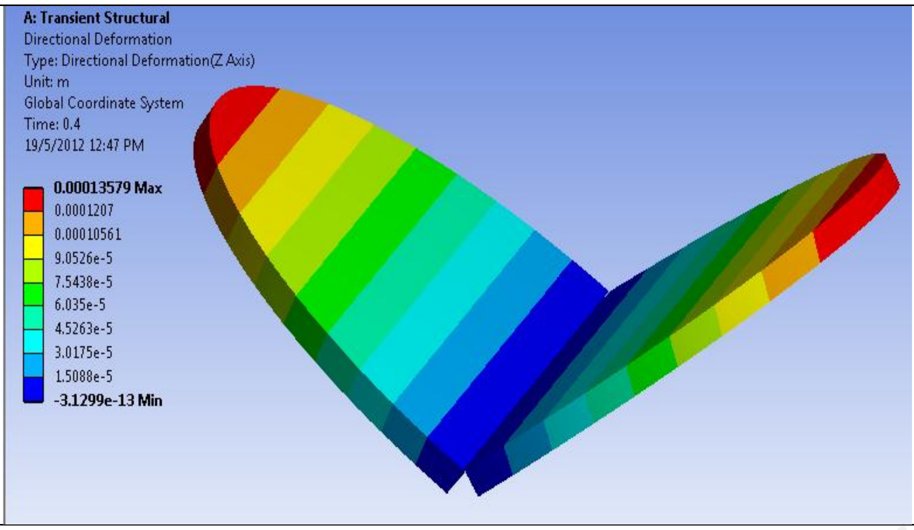
z-Velocity of the blood flow (m/s)



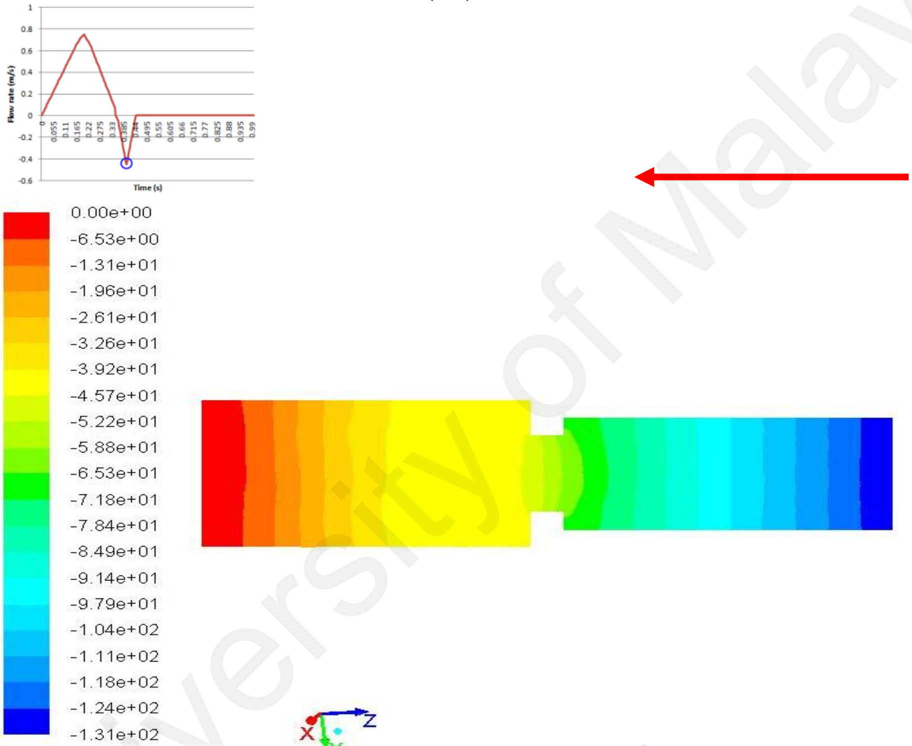
Time(t) = 0.4s

Leaflet movement and deformation (m)

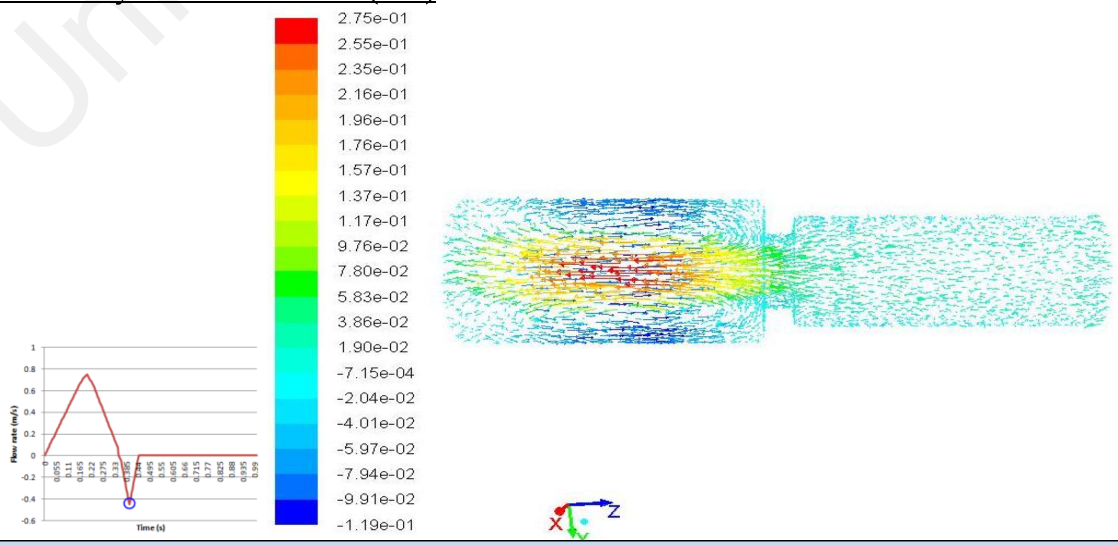




Static Pressure of the blood flow (Pa)

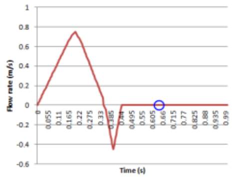


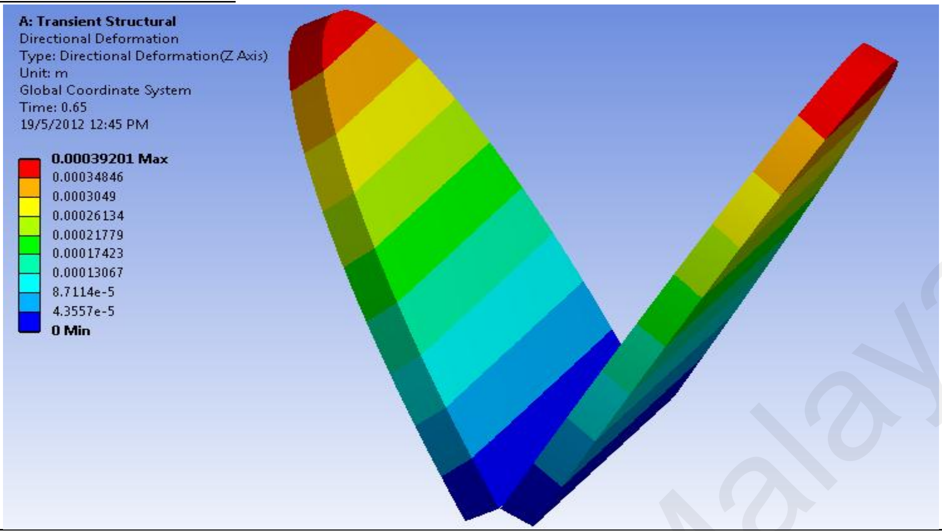
z-Velocity of the blood flow (m/s)



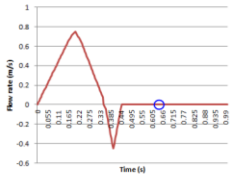
Time(t) = 0.65s

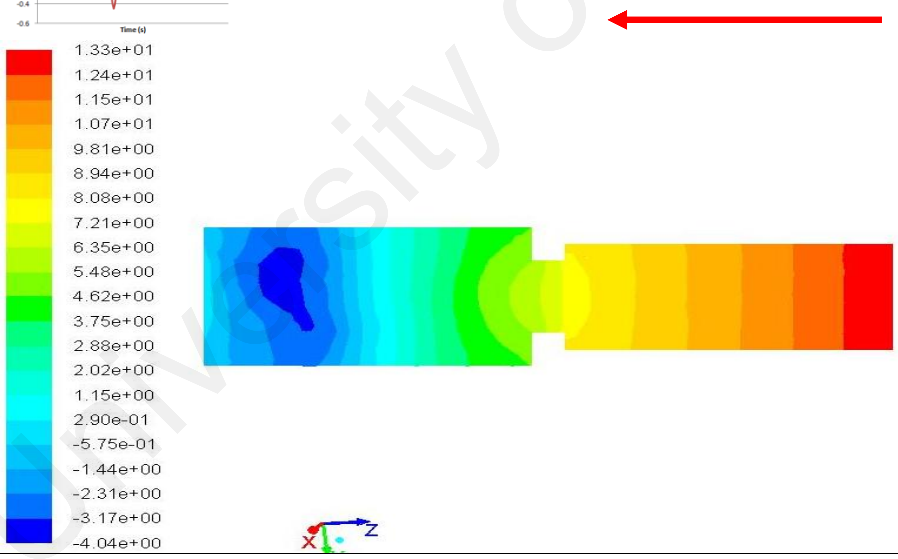
Leaflet movement and deformation (m)



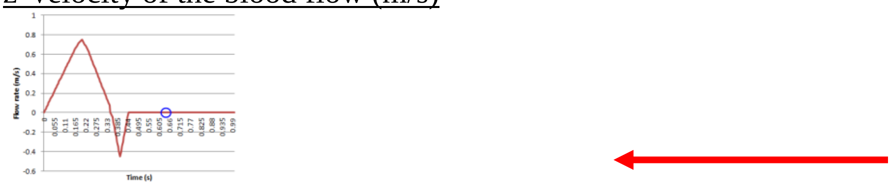


Static Pressure of the blood flow (Pa)





z-Velocity of the blood flow (m/s)



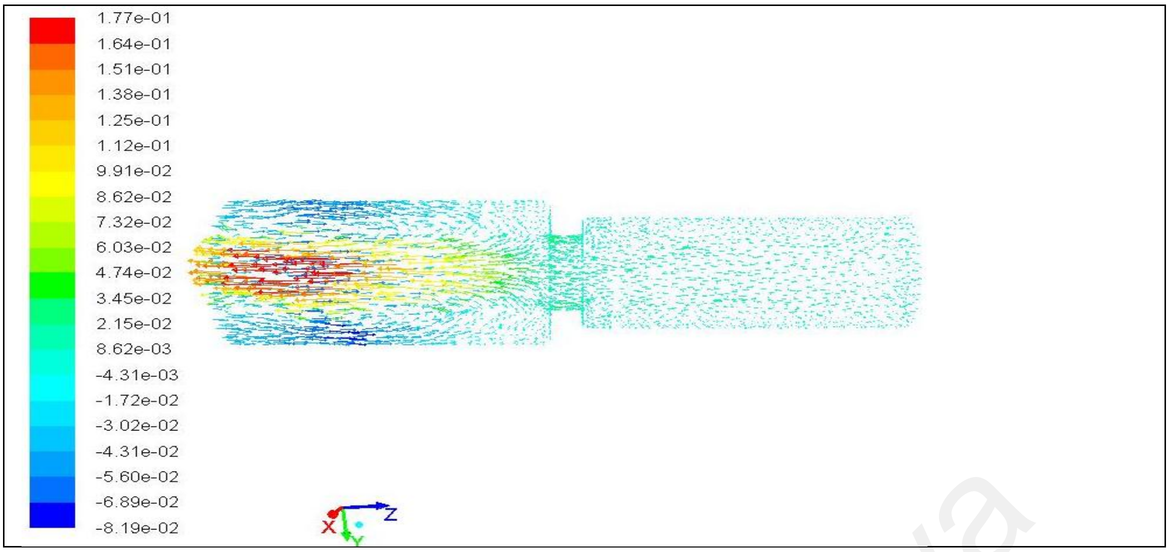


Figure 4.4 shows the bileaflets directional deformation against 1 cardiac cycle.

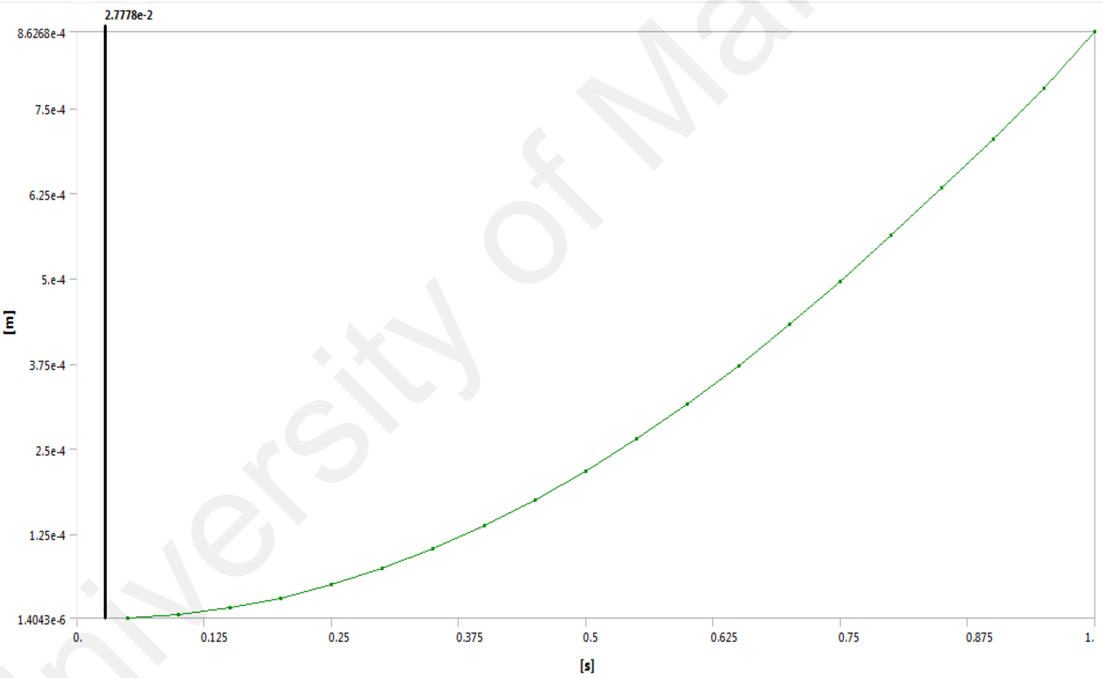


Figure 4.4 : Leaflets deformation against time

Figure 4.5 shows the convergence forces exerted on the leaflet during the entire time step.

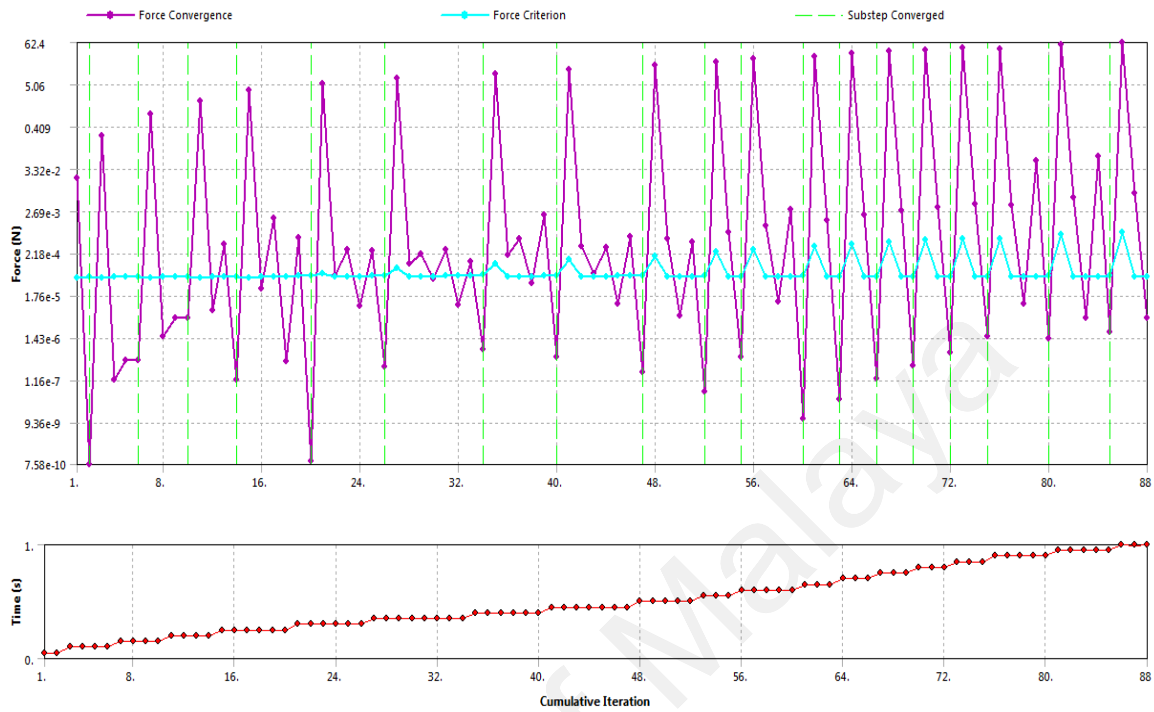


Figure 4.5 : Convergence force on the leaflet

Figure 4.6 shows the convergence of the leaflet displacement in the entire time step.

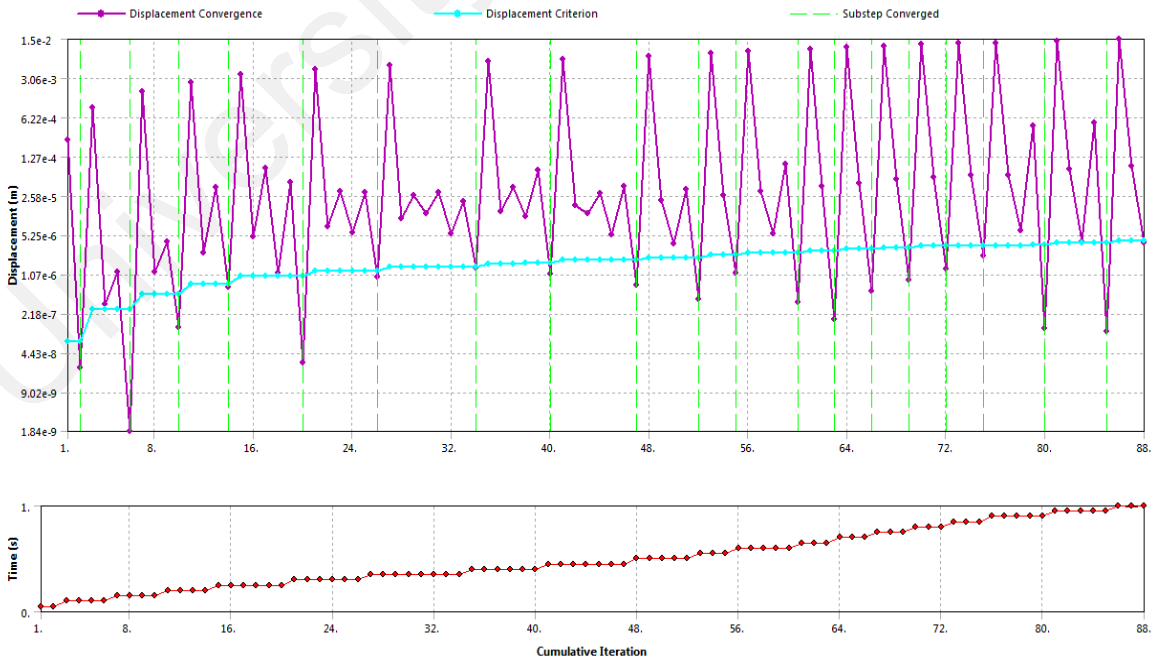


Figure 4.6 : Displacement convergence of leaflet

Figure 4.7 shows the maximum increment of the DOF of the leaflets in the time step.

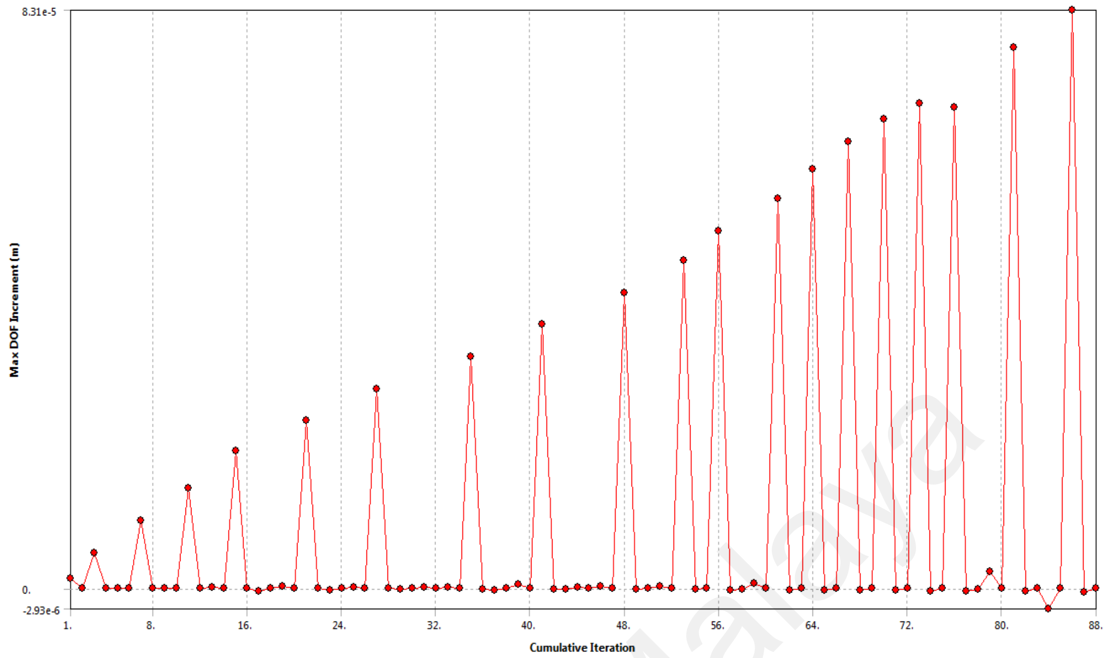


Figure 4.7 : Maximum DOF of leaflet

CHAPTER 5. DISCUSSION

The results shown above are partial results that were obtained in the first and second stage of simulation; which is during individual leaflet and blood flow simulation. The final results of coupling could not be obtained during the submission of this thesis due to time constraint. The expected results with the completion of final stage of simulation, which is coupling include the energy loss during 1 cardiac cycle, regurgitation and disturbed blood flow pattern when the valve is in fully closed position, shear stress during the movement of the leaflets, development of vortices and stagnation points. Besides, analyses are expected to be performed also to measure forward flow pressure drop, root-mean-square forward flow rate, closing, leakage and total regurgitant volume, and forward flow, closing, leakage and total energy loss during the entire cardiac cycle. It is reported that noticeably large shear stress found at the tip of leaflets and during the contact region between the valve housing and leaflets at the closing phase (Cheng, 2004). The characteristic of the opening and closing of valve leaflets is dependent upon the dynamic interaction between the leaflets and the blood flow. This interaction is greatly influenced by the leaflet thickness and the geometry of the leaflet itself. The pressure in the blood flow rises from free stream pressure to stagnation pressure as the blood flow particle streams towards the edge of the leaflets (Hsu, 2001) . The blood fluid experiences a net pressure force opposite to its direction of motion as the pressure increases in the direction of flow. The result also shows that negative pressure develops quickly at the tip of the atria side. Pressure becomes more negative during the progressive leaflet closure and reaches the lowest pressure when the leaflet is in the closed position (Lai, 2002). The dynamics of blood flow during the closing stage of leaflet is highly dependent with the velocity of the leaflet at the particular instant of time. A reduction of 50% is noticed to the magnitude of negative pressure during the last four degrees of leaflet motion as the tip of the velocity decreases by a factor of three (Lai, 2002). The contour plots of the axial velocity shown at the several selected times illustrates that higher values of axial velocity is associated with the jet through the central orifice in the valve region. The flow is distributed evenly at the downstream of the valve plane during the deceleration phase. During the acceleration phase of the valve leaflets, the blood flow emerges with a higher velocity from the central orifice. The magnitude of vorticity increases in the valve region as the valve opens and with the increase in the blood flow rate. The closing phase of the leaflets starts when the flow is completely inverted. The magnitude of vorticity increases in the valve region with the increase in the blood flow and during the opening of the valve. The maximum values of vorticity are obtained when the blood flow reaches the surface of the leaflets, near the valve ring and in the transitional areas between the three jets. A progressive narrowing of central orifice jet can be predicted during systole succession due to the progressive shed and downstream movement noticed at the trailing edges of the leaflets that progressively shed and move downstream. An area of low velocity is seen to develop immediately at the downstream of the central orifice jet. Small discrete, swirling vortices are also noticed at this area which signifies that mixing is taking place on the edge of the central orifice jet (Hsu, 2001). No vortices were observed during the opening stage of the valve and the maximum velocity is noted at the central region of the aortic tract through the contour plotted in the results section which is at 0.2 s. The pressure throughout the cardiac cycle remained almost constant in the axial plan except near the leaflet region (Sotiropoulos, 2004). Besides, numerical studies are also able to provide the indication regarding blood flow velocity distribution within the housing of the valve. This region was inaccessible to the traditional experimental method previously where the flow field description was near the walls of the housing only. The results obtained are analyzed from a Lagrangian point of view to explain the complex structure of the pulsatile blood flow through the valve.

The maximum pressure drop is expected to occur at edge of the leaflets and the inner housing when the valve is in fully opened position with the velocity flow rate is at its maximum (Yoganathan, 2005). Transvalvular pressure drop is measured by the difference of the pressure obtained at the inner and outer surfaces of the leaflets in the middle of the leading and trailing edges. Yoganathan (2005) had also quoted that the flow through the St Jude valve is an unobstructed and a centralized one where optimal effective valve area for a given orifice area can be achieved. As for the pressure loss discussed earlier, the energy loss can be determined through the product of the forward blood flow rate (Q) and pressure drop (ΔP) with the addition of energy loss due to the leakage near the hinges ($\Delta E_{leakage}$);

$$\Delta E_{energy,loss} = \int_{0}^{T_s} Q.(\Delta P)dt + \Delta E_{leakage}$$

where T_S is defined as systolic ejection period. The influence of valve performance during each cardiac cycle can be characterized in terms of energy loss. The energy loss of heart work corresponds directly to the loss of pressure experienced across the heart valve (Hellevik, 2006).

The effective orifice areas (EOAs) on the other hand can be determined by performing a linear regression analysis on the root mean square of the forward blood flow rate (Q_{rms}) and pressure drop (ΔP) from the cardiac outputs. The regression performed is to determine the best fit of the area of the 90 data points from three cardiac cycles.

$$Q_{rms}^2 = m\Delta P + b$$

Once the Q_{rms} is obtained, the EOA can be calculated from the equation below:

$$EOA = \frac{Q_{rms}}{51.6\sqrt{\Delta P}}$$

CHAPTER 6. CONCLUSION AND RECOMMENDATION

Although numerical simulation computational techniques are influential and powerful tools in assessing the design of heart valves virtually, there are still limitations which causes the result achieved do not exactly represent the actual valve design. As an example, to perform FSI, a gap is created between the housing and the closed valve leaflets whereby the size of the leaflets were reduced to 98% from their actual size. This gap is needed as computing FSI necessitates the need of physical separation of the fluid domain (blood flow) from the structural domain (leaflets). This difference maybe small but this dimension eventually underestimates the shear stress that occurs between the leaflets and the housing. The shear layers which enclose the hinges and leaflets due to high gradient of velocity flow expose the platelets to elevated shear stresses leading them to the entrapment and consequent damage to the red blood cells in the shed vortices of the hinges (Sebastiaan Annerel, 2010). Regions of stagnation, separation of blood flow and recirculation had been found to encourage the increase of thrombi formation and the deposition of damaged blood cells (Dasi, 2009). Besides, (Dumont, 2005) had pointed out that the yielding of central gap had lead to an overestimation of the leakage flow through the gaps which causes most of the researchers to avoid analyzing the leakage flow in their research work. Besides, scaling the leaflets to 0.98 from its actual size causes the leaflets to have a 4% reduction on the leaflet total area. As such, it might affect the results of the dynamics of the scaled leaflets between the numerical and experimental results. To simulate a full dynamic behavior of valve leaflets in a flow tract, a FSI model is required as the dynamic motion is essential in the simulation of repairing the valve such as the edge to edge technique where the dynamics of valve and leaflets are affected by the EOA and the restricted motion of the leaflets due to the presence of hinges. Furthermore, the role of valve motion on the flow rate and its effect to the blood circulatory system can be administered by simulating the valve in a normal and pathological state as the mitral valve leaflets regulates the flow of blood in the heart (Lau, 2010). A high-resolution FSI algorithm is needed which includes the detailed structural analysis of leaflets valves and the finite particle dynamics of blood flow to simulate the motion of red blood cells and platelets to understand better on the functional mechanics of the valve (Sotiropoulos, 2004). On top of that, the design of the aortic tract need to be more closer to the anatomical aortic root shape in future studies as the dynamics of blood flow is extremely sensitive in this area to the features present in the vascular geometry. The design should be in a Valsalva sinuses at the beginning of outlet tract in future work where it is located right after the mitral valve prosthesis. This design is a need as studies had predicted that the presence of Valsava sinuses enables a greater forward flow and backflow of about 2% of the forward flow in the presence of this geometry (Tullio, 2009) as shown in Figure 6.1.

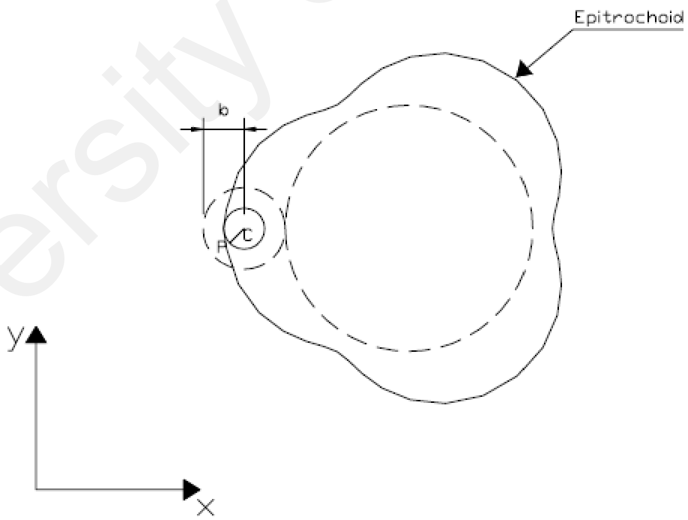


Figure 6.1 : Epitrochoidal representation of a cross section of the aortic root (Grigioni, 2003)

There are other reasons motivating this design as well. This geometry imitates closely the tract of the artery pulmonary. Besides, local phenomena such as damage of red blood cells and activation of platelets occur at the near the valve leaflets, housing and hinges due to the high level of shear stress. On top of that, the BMHV simulation should be performed with less assumptions and simplifications done to the model to imitate the exact BMHV that will be placed in the aortic tract in the open position. The number of simulated cycle flows need to be increased to get a more converged statistical results and a more realistic modeling should be designed for the hinge of the leaflets at the housing as flow at the region is often omitted in the research where it contradicts with the experimental results whereby the highest viscous stresses and hemolysis occurs at this region and this area should not be omitted from a numerical simulation research (Dumont, 2005). The closure of valve phase implicates the initiation of thrombosis activities and the deformation that occurs at the aortic walls can change into turbulent flow which should be investigated in future works as well. ALE based models provide a more accurate results in current research, however, these models have two main drawbacks; where simulation is demanding in terms of costs and mesh handling requirements and the need of diffusion for local numerical in regions where remeshing is required to be performed (Sotiropoulos, 2004). As such, to increase the adequate accuracy level in numerical simulation, one should consider the complex actual scenario and physics involved such as improvement to the FSI algorithm to enable complex designs to be accepted for higher spatial and temporal resolution and turbulence presence at the valve housing during the closing stage with reasonable computation times and cost.

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