ANALYSIS OF BLOOD FLOW IN 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION

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Report submitted in partial fulfillment of the requirements
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Faculty of Mechanical Engineering
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SESI PENGAJIAN: 2011/2012

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Date : 22th JUNE 2012

DEDICATION

I specially dedicate to my beloved parents
and those who have guided
and motivated me for this project

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Alhamdulillah, Thanks to Allah S.W.T, with His willing, gives me the opportunity to complete my final year project "Analysis of blood flow into 3D heart valves under steady state condition". This project fully supervised by Mr.Mohd Azrul Hisham Mohd Adib from Faculty Mechanical Engineering, Universiti Malaysia Pahang (UMP).

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ABSTRACT

Mitral valve (MV) and Aortic valve (AV) are very important in circulatory blood inside the heart which function as a gate for the blood flow. Since the heart valves not working properly, the normal flow of blood inside the heart will be interrupeted. Malfunctioning Mitral Valve and Aortic Valve will cause death if it not immediately detected. The aims of this study are to investigate the blood flow pattern for MV and AV in 3D view which developed by ADINA-FSI application. All the parameter of blood obtained from the previous studies. Both of valve models have been implemented without ventricles and presented as a Newtonian fluid flow in steady condition. The simulation from this study will cover the blood flow pattern in term of velocity, effective stress on the leaflets, strain occurred in the critical area of both heart valves and the nodal pressure pattern flow during the systolic process phase. Moreover, these studies enable to investigate performance characteristics such as effective stress and strain which are very difficult to evaluate experimentally. The findings showed, the velocity and nodal pressure are linear correlation with the blood pressure increased during systole process. Effective stress and strain results showed the critical area region and maximum values with fix Young Modulus 2MPa. The significant simulation results from this study were very useful to give a clear view to the medical practitioners about the pattern of blood flow through MV and AV. Furthermore, these models can be used to investigate heart valve failure and subsequent surgical repair treatment.

ABSTRAK

Injap Mitral dan Injap Aortic merupakan bahagian penting yang berfungsi sebagai pintu untuk proses keluar masuk bagi peredaran darah di dalam jantung. Sekiranya injap ini mengalami kerosakan atau tidak berada dalam keadaan baik, maka peredaran darah akan mengalami gangguan dan seterusnya akan menyebabkan komplikasi pada jantung. Objektif untuk projek ini adalah untuk mengkaji bentuk atau illustrasi peredaran darah dalam 3 dimensi yang melalui injap Mitral dan injap Aortic dengan penggunaan aplikasi perisian ADINA-FSI. Parameter bagi darah diperolehi dari sumber kajian.Kedua-dua injap yang dihasilkan adalah tanpa mengambil kira ventricles dan dikategorikan sebagai bendalir Newtonian serta sistem berada dalam keadaan yang stabil.Simulasi dari projek ini akan menghasilkan keputusan peredaran darah dalam konteks halaju, kesan tekanan yang berlaku pada injap, kesan terikan dan tekanan darah yang mengalir dalam sistem jantung pada proses fasa tekanan tinggi.Tambahan, untuk mengkaji kemampuan bagi kesan tekanan dan terikan adalah merupakan satu perkara yang agak sukar di lakukan atau di terjermahkan melalui eksperimen. Berdasarkan keputusan yang di analisa dari applikasi ADINA-FSI, keputusan menunjukkan halaju dan tekanan darah yang mengalir melalui kedua-dua injap ini adalah berkadar terus dengan tekanan yang di berikan kepada sistem. Signifikasi analisa bagi projek ini, keputusan yang di perolehi dapat digunakan oleh pakar perubatan untuk memberi gambaran jelas kepada mereka berkaitan dengan bentuk peredaran darah yang melalui kedua-dua injap ini berdasarkan parameter. Malah keputusan dari projek ini dapat di gunakan untuk mengkaji tahap kemampuan sistem injap jantung serta di gunakan untuk mengesan sekiranya berlakunya kerosakan pada injap jantung dan rawatan pembolehbaikan di dalam pembedahan.

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LIST OF SYMBOLS

A	Cross sectional area
P	Pressure
E	Young’s modulus elasticity
a	Acceleration
σ	Stress
ε	Strain
m	Mass
V	Velocity
t	Time
$\dot{m}$	Mass flow rate
F	Force
Ma	Mach number
Cp	Specific heat
T	Temperature
v	Kinematic viscosity
μ	Viscosity
ρ	Density
%	Percentage

LIST OF ABBREVIATIONS

ADINA	Automatic Dynamic Incremental Nonlinear Analysis
FSI	Fluid Structure Interaction
3D	3-Dimensional
UMP	University Malaysia Pahang
ALE	Arbitrary Lagrange Euler
FEA	Finite Element Analysis

CHAPTER 1

INTRODUCTION

1.1 BACKGROUND

The heart is a powerful muscle that pumps blood throughout the body by means of a coordinated contraction. It works as a pump to send oxygen-rich blood through all the parts of the body. A human heart beats an average of 100,000 times per day. During that time, it pumps more than 4,300 gallons of blood throughout the entire body.

Basically, the heart consists of four compartments at right and left side which is two atrium and two ventricles. For each atrium and ventricles there have four valves which act to keep the blood flowing in a unidirectional. Repaired or reconstruct process for a heart valve is very complicated step and the percentage of risk to have bad complications was very high. But the risk of complications of heart valves dependent upon a full understanding of normal and abnormal heart valve anatomy and function. Restoration of valve function is performed via valve repair or artificial replacement. Repair is preferred in all cases (if suitable), with improved mortality, life expectancy, reduced risk of stroke and valve related complications.

This project study just focused to the left heart compartments. The mitral and aortic valves are the common valves are the most common sites of heart valve disease, because of their location on the left side of the heart. The left chamber has a greater workload because the left side of heart need to pump the blood to the entire of body, meanwhile the right chambers just work to pump the blood flow only to the lungs. Computational simulation is one of the methods can be applied to give the clearest view of the interaction process blood flow through the MV and AV leaflets structure.

Therefore the fluid structure interaction models of mitral and aortic valve have been generated in three dimensional views to present the process of blood flow activities via both valves. By using the existing anatomy parameters and geometry from previous study, the 3D models of mitral and aortic valve implemented by using the SOLIDWORK application.

1.2 PROBLEM STATEMENT

Blood flow pattern is a fundamental aspect of cardiac function and abnormal function maybe reflected in the flow patterns. The Mitral valve (MV) and Aortic valve (AV) are very important in circulatory blood inside the heart which function as a gate for the blood flow. The Mitral valve (MV) and Aortic valve (AV) are very important in circulatory blood inside heart which function as a gate for the blood flow. Since the heart valves not working properly, the normal flow of blood inside the heart will interrupt. This heart disease will affect the efficiency of heart system. According to the journal "*Simulation of blood flow through the mitral valve of the heart, a fluid structure interaction model*" by Espino, Watkins, Shepherd, Hukins, and Buchan 2006. This paper investigates about the mitral valve failure causes death if it is not corrected surgically. Such surgical repair can be improved by understanding mitral biomechanics. A two-dimensional FSI model of the mitral valve was generated. A simple approximation of the heart geometry was used; the valve dimensions were based on measurements made. Leaflet deformations agreed with results from experiments in the literature and with our previous experimental results. The blood flow pattern of heart circulatory via of both these valves can illustrate clearly in the 3D model. The certain parameters such as density of blood and leaflets of both valves will fix during the simulation process. The purpose of this study was to compute and visualize the blood flow pattern via the mitral valve and aortic valve within the cardiac cycle pressure range.

1.3 OBJECTIVES

At the end of this project,the flow pattern via the mitral valve (MV) and aortic valve (AV) can be determined by using the simulation analysis based on the parameter of blood.

1.4 SCOPES

The scope of study is very important in order to identify the area of focus and boundaries of the research. The scope for this study are as follows:-

i. The model will be limited to three dimensional (3D) model
ii. The blood flow via MV and AV are in a steady state condition
iii. Blood assumes as Newtonian
iv. The blood flow assumed in isothermal condition.

CHAPTER 2

LITERATURE REVIEW

2.1 HUMAN HEART

The human heart is the biological pump that generates the blood to moving throughout of heart circulatory system. Every second the heart contracts and expands without pause and the blood flows only in one direction in the circulatory system. In fact, our heart pumping approximately 2,000 gallons (7580 liters) of blood on average 1000 times

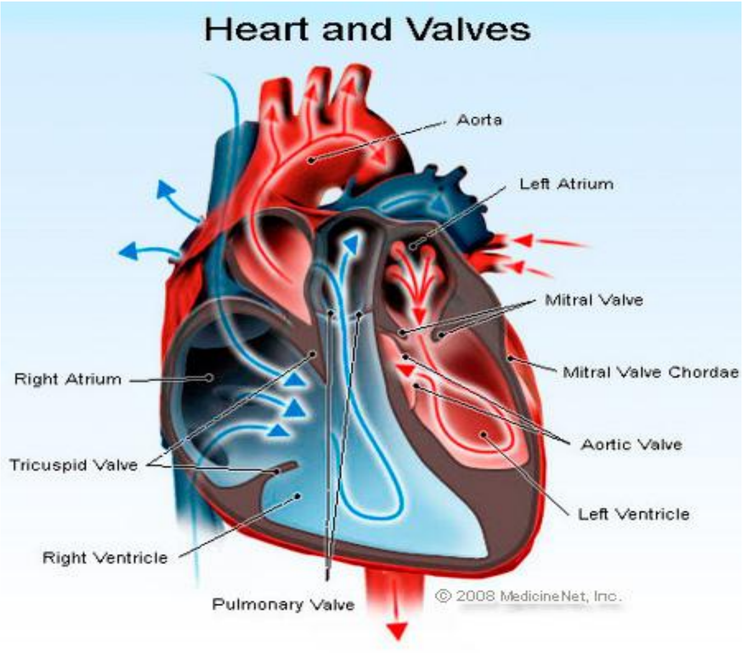


Figure 2.1: Healthy heart cross section

Source : Journal Medicine Institute (2006)

Each valve has thin flaps of muscle tissue, called leaflets or cups that open to let the blood through and close tightly to avoid the backflow occurred during the systole and diastole process. The right atrium receives blood from the great vein (superior and inferior vena cava). It is the content high of carbon dioxide but is relatively depleted of oxygen, which has been absorbed by the body tissues. Tricuspids valve will open to receive the blood from the right atrium. After filling the right ventricle, the blood will pumped via the pulmonary valve directly to the lung organ. In the lungs carbon dioxide is removed from the blood and oxygen will absorb to the blood flows. The oxygenated bloods from the lungs enter the heart halves via Mitral valve (MV) to the left atrium. Finally the blood from the left atrium will pump out through the Aortic valve (AV) to the circulatory system.

The Figure 2.1 shows the direction of blood flow via four heart valves. The blue arrow represents the depleted of oxygen in blood while the red arrow means the high contains of oxygen from the lungs. Cardiac cycle is the sequence of event for the blood completed the circulatory system. Systolic pressure defined as the high pressure when the beat or the contraction phase condition of heart forces blood circulates the whole body. Cardiac output defined as the volume of blood ejected from the left ventricle per minute. Diastolic pressure means the pressure between the heartbeats when the heart in the relax phase condition.

2.2 HEART VALVE DISEASE

Heart valve disease is the condition in which one or more valve doesn't function properly. Each heart valve has thin flaps of muscle tissue (leaflets) act to ensure the blood flow always flow in the right direction. Since the heart valves not working properly, the normal flow of blood inside the heart will interrupt.

2.2.1 Regurgitation

Regurgitation occur when the valve does not close completely and letting blood leak backwards across the valve during diastole (relaxation). This backward flow is referred to as “regurgitates flow”.

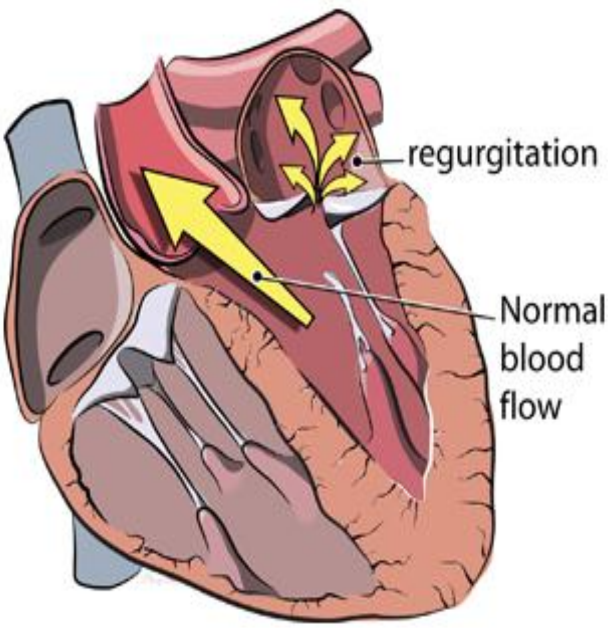


Figure 2.2: Leaky Valve in the mitral valve position

Source : Pick (2011)

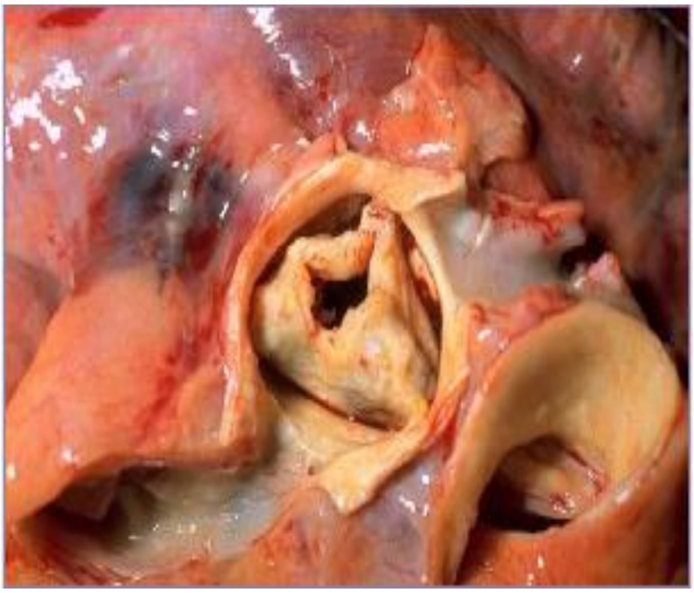


Figure 2.3: View Of Heart Valve Problem Causes By Regurgitation Problem

Source: Retrieved Kardiol Website (2011)

2.2.2 Stenosis Problem

The stenosis problem occurs when the tissue forming the valve leaflets becomes stiffer and the effect problem is the opening gate of a valve will become narrow. So the amount of blood flow via the valve less than a normal condition. However the valve can become so narrow which causes the heart function effectiveness reduced and the body will receive inadequate blood flow.

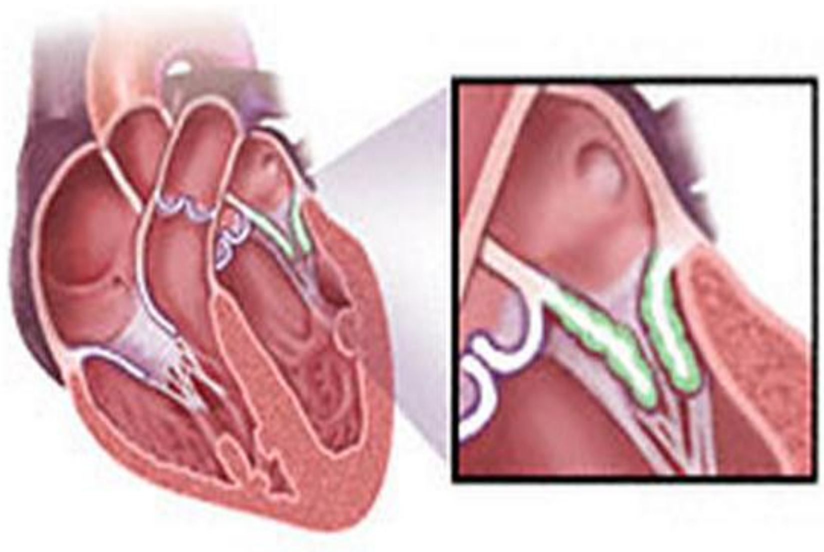


Figure 2.4: The illustrations of gate valves become narrow

Source: Journal American Society of Echocardiography (2011)

2.2.3 Infection Endocarditic

Endocarditic infection is the condition of the bacteria will enter the blood stream and attach to the surface of heart valves. Colonies of microorganism that grow on the endocardium can cause holes in the valve. So this condition will interrupt the effectiveness of heart valve's ability to function well.

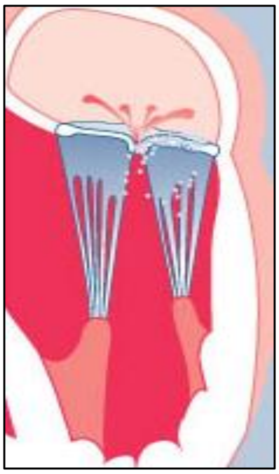


Figure 2.5: The bacteria attach at the top of the heart valve surface

Source : Retrieved from Cleveland Clinic Website (2011)

2.2.4 Rheumatic fever

This problem usually started with an untreated streptococcal bacterial infection. It can affect any tissue in the body, including the joints, the brain, and the skin but most importantly it can scar the heart muscle and particularly the heart valve.

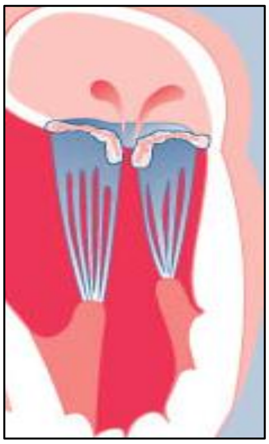


Figure 2.6: Scarring of the heart muscle

Source: Retrieved from Cleveland Clinic Website (2011)

The damaged such as inflammation, scarring problem at the heart, not just caused by bacteria infection but it has also come from the antibody response to fight the bacteria infection.

2.3 MITRAL VALVE (MV)

The mitral valve is located between the left atrium and left ventricle chamber of the heart. Normal mitral valve consists of two leaflets, anterior and posterior. During the systolic process, the mitral valve function is to open the gate to allowing the blood flow. During the diastolic process, the mitral valve gate will close immediately to avoid unidirectional blood flow activities.

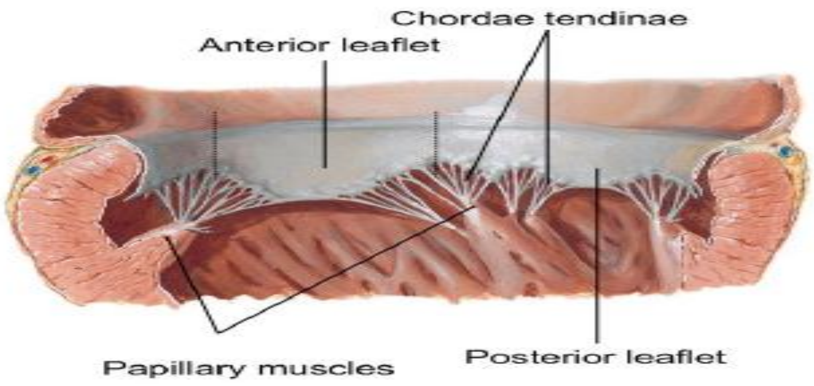
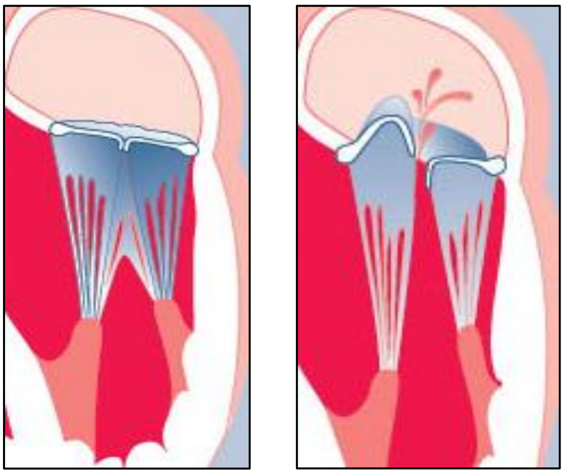


Figure 2.7: Mitral Valve Structure

Source: Thor (1999)

2.3.1 Mitral Valve Prolapsed (MVP)

This disease is also known as click murmur syndrome, floppy-valve syndrome, balloon mitral valve, and Barlow’s syndrome. It is a deformity of the mitral valve that may prevent its leaflets from closing properly.



(a) (b)

Figure 2.8 : (a) Normal Valve and (b) Mitral Valve Prolapse

Source: Retrieved from Cleveland Clinic Website (2011)

One or more leaflets maybe bulging or the entire valve may be out of its normal position. Depending on the degree of the deformity, the prolapsed can lead to mitral regurgitation.

2.4 AORTIC VALVE (AV)

Aortic Valve (AV) located between the aorta (main artery) and left ventricle. The main function of AV is to ensure when the ventricles contract, the blood flows always in the correct direction. The AV consists of three leaflets which function as a gate in heart system.

The blood flow via MV will fill the left ventricle at the early part of diastole process. After blood fully filling the left ventricle, the MV was closed immediately to avoid backward blood flow. Then the AV will open when the ventricle in a contract condition to allow the blood flow through the AV direct in the circulatory system. The process open and closes for AV and MV will occur repeatedly in the heart beat system without pausing.

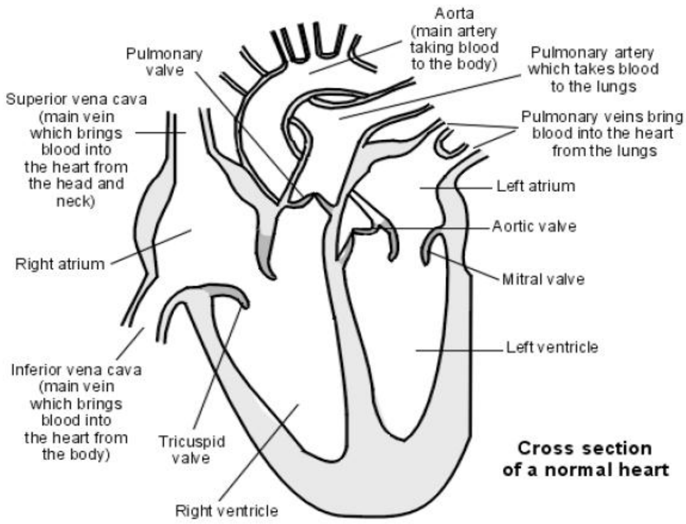


Figure 2.9: Cross section of normal heart

Source: Article NICE Clinical Guideline (2008)

2.4.1 Aortic Valve Stenosis

This problem occurs when the AV becomes narrowed. So when it happens the circulatory system will receive inadequate blood and the left ventricle has to work harder to pump blood into the aorta.

If the problems become severe, it will cause a lot of health problems such as dizziness, faints, chest pain, shortness breath, tiredness and fluid retention in various tissues of the body.

2.4.2 Aortic Valve Regurgitation

Aortic valve regurgitation is according to the problem of valve leaflets that failed to function as normal condition. This problem is referring to the leaky valve and cause the backward blood flow happens during the diastole process. Lists of health problem affected by this disease are such as dizziness, faints, chest pain, shortness breath, tiredness and fluid retention in various tissues of the body.

2.5 FLUID THEORY

2.5.1 Incompressible flow

A fluid flow where variations in density are sufficiently small to be negligible. Flows are generally incompressible either because the fluid is incompressible liquids or cause by the Mach number is low (roughly < 0.3).

$$\frac{\Delta p}{p_0} = \frac{1}{2} M^2 \leq 1 \tag{2.1}$$

Regarding to the scopes area of projects, the blood will be assumed as incompressible flow. The densities of liquids are essentially constant and thus the flow of liquids is typically incompressible. The pressure value , p_0 will be key in as a cardiac cycle of the heart system as low as 5.3 kPa until 23.3 kPa.

2.5.2 Navier Stokes Equation

$$\rho \frac{D\vec{V}}{Dt} = -\vec{\nabla} P + \rho \vec{g} + \mu \nabla^2 \vec{V} \tag{2.2}$$

From the scopes area of projects,the project limit to the Newtonian fluids, where by definition the stress tensor is linearly proportional to the strain rate tensor. Instead the density assumed constant and nearly isothermal flow. A further consequence of the latter assumption is that fluid properties such as dynamic viscosity and kinematic viscosity are constant as well. Notice that, the pressure value comes from the literature review data.In this approximation, the Navier stokes equation reduces to just two terms, pressure and gravity.

2.5.3 Continuity Equation

Basically if the flow is approximated as incompressible, density is not a function of time or space.

$$\frac{\partial u}{\partial x}+\frac{\partial v}{\partial y}+\frac{\partial w}{\partial z}=0 \tag{2.3}$$

Nothing that the duct cross sectional area is constant. That's means, the structure area of model leaflets aortic and mitral valve design are not changed. $A_{initial}=A_{final.}$

2.5.4 Momentum Equation

$$\ell\left(\frac{\partial u}{\partial t}+u\frac{\partial u}{\partial x}+v\frac{\partial u}{\partial y}+w\frac{\partial u}{\partial z}\right)=X-\frac{\partial p}{\partial x}+\mu\left(\frac{\partial^2 u}{\partial x^2}+u\frac{\partial^2 u}{\partial y^2}+v\frac{\partial^2 u}{\partial z^2}\right) \tag{2.4}$$

$$\ell\left(\frac{\partial v}{\partial t}+u\frac{\partial v}{\partial x}+v\frac{\partial v}{\partial y}+w\frac{\partial v}{\partial z}\right)=Y-\frac{\partial p}{\partial x}+\mu\left(\frac{\partial^2 v}{\partial x^2}+u\frac{\partial^2 v}{\partial y^2}+v\frac{\partial^2 v}{\partial z^2}\right) \tag{2.5}$$

$$\ell\left(\frac{\partial z}{\partial t}+u\frac{\partial z}{\partial x}+v\frac{\partial z}{\partial y}+w\frac{\partial z}{\partial z}\right)=X-\frac{\partial p}{\partial x}+\mu\left(\frac{\partial^2 z}{\partial x^2}+u\frac{\partial^2 z}{\partial y^2}+v\frac{\partial^2 z}{\partial z^2}\right) \tag{2.6}$$

This project limited to 3D model for both valve aortic and mitral valve. This equation needed to establish the velocity field through the use of the velocity potential function.

2.5.5 Newtonian Flow

When a fluid is subjected to a shear stress, the fluid continuously changes shape (deformation). If the fluid is Newtonian, the rate of deformation (shear strain) is proportional to the applied shear stress and the constant of proportionality is called viscosity. In general flow, the rate of deformation of a fluid particle is described mathematically by a strain rate tensor and the stress by as stress tensor. Inflows of Newtonian fluids, the stress tensor are proportionality is called viscosity. Most common fluids such as water, oil, gasoline, air, most gas and vapors without particles or large molecules in suspension are Newtonian.

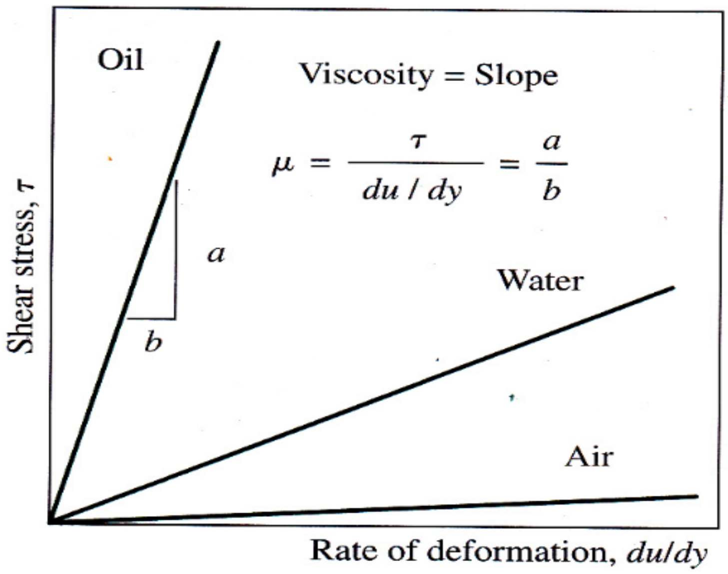


Figure 2.10: The rate of deformation of a Newtonian Fluids

Source : Books Fluid Mechanics Fundamentals and Applications 2nd Edition (2011)

2.5.6 Steady Flow

The terms steady implies no change (constant) in all fluids variables of such as velocity, density, temperature, etc at a point in time. Steady flow conditions can be closely approximated by devices that are intended for continuous operation such as turbines, pumps, boilers, condenser, heat exchangers and cyclic system. During the steady condition, the fluid properties can change from point to point within a device, but at any fixed point they remain constant.

2.6 FLOW PATTERN STUDY

2.6.1 Flow Pattern inside 3D Valve model

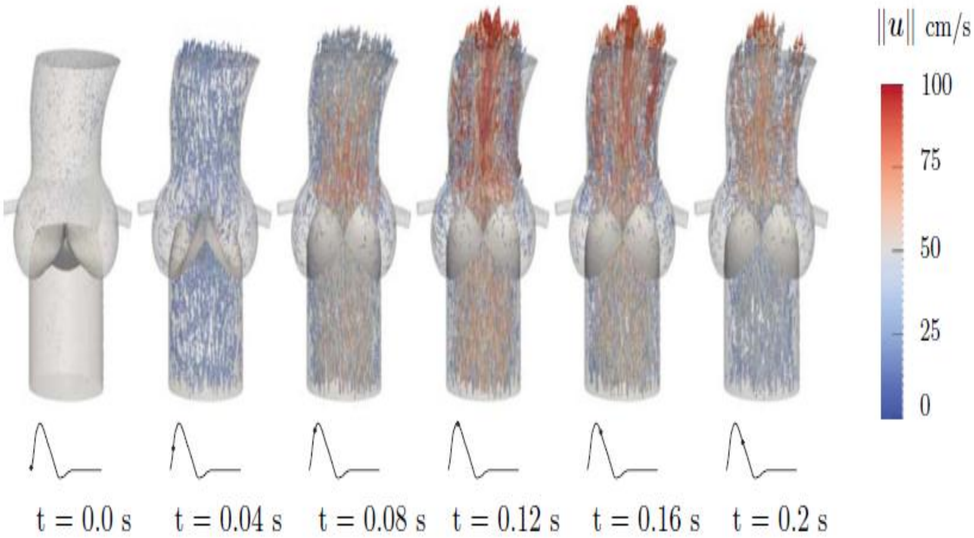


Figure 2.11: Flow Pattern in the 3D view of Aortic Valve simulation

Source: Shadden et.al (2010)

The Figure 2.11 showed the results of flow pattern via the aortic valve at the different certain sequential time. The flow was simulated over two cardiac cycles with a time-step size of $2.5x10^{-4}$ s and data from the last cardiac cycle were used for analysis. Time 0 corresponds to the start of systole the opening of the valve and peak systole occurs after approximately 0.11 seconds.

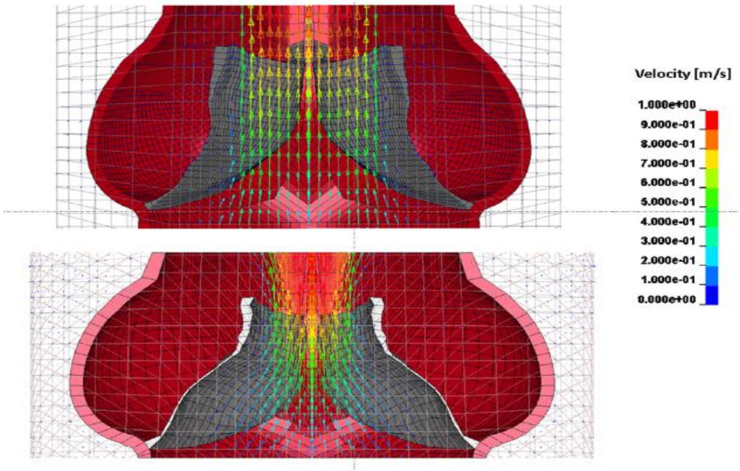


Figure 2.12: Predicted flow fields for aortic valve during mid-systole

Source: Weinberg et.al (2008)

Referring to the Figure 3.20 above, the aortic valve showed the interaction velocity and the movement of aortic valve leaflets. The leaflets effect is reflected by the velocity given to the aortic valve leaflets.

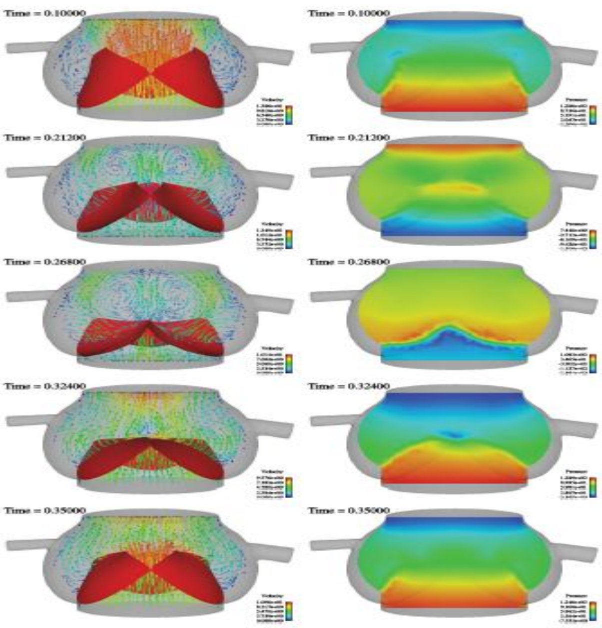


Figure 2.13: Blood velocity vectors (left) and pressure field (right) for 5 different time steps

Source: Astorino et.al (2002)

The Figure 2.13 showed the interaction between the blood and aortic valve model. From the figure above the difference flow pattern velocity can be seen clearly at the difference sequence time. On the left side of blood velocity vectors showed the pressure effect inside the aortic valve.

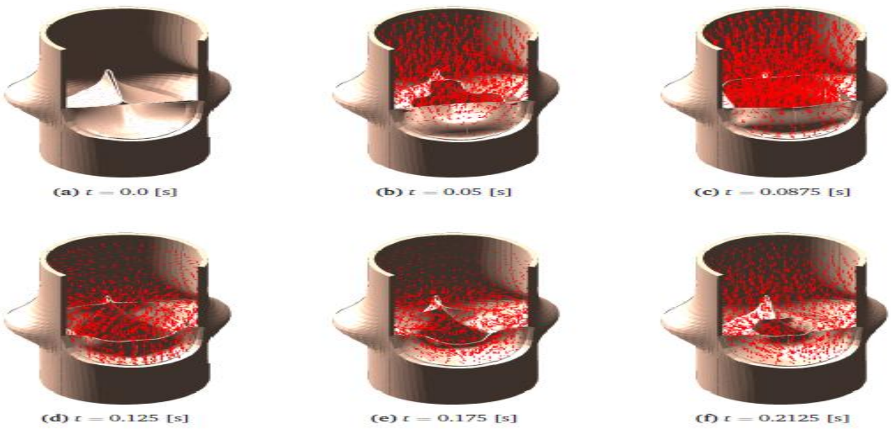


Figure 2.14: Configurations of the stented valve taken at 6 successive points in time during the systolic phase

Source: Hart (2002)

From the Figure 2.14 above, the flow pattern through the aortic valve showed velocity vector for successive points in time. The stress scale is given in kPa and the maximum principle stresses vary from -50 kPa to 60 kPa. The middle frame represents the initial phase of valve closing, which shows significant stresses in the middle of the leaflet. For this configuration the compressive stresses appear on the aortic side, whereas tensile stresses are more dominant on the ventricular side. This phenomenon shows a cyclic loading pattern of the aortic leaflets, which is believed to be important in the analysis of fatigue behavior. The last two frames are taken at the end of the systolic phase, where a rapid closure of the valve occurs.The stresses in the leaflets are significantly increasing in this phase as they have to bear the increasing pressure gradient across the valve. As stated before, the simulation does not include the diastolic phase. Moreover, with the applied material parameters, the leaflet would undergo excessive deformations to balance the diastolic pressure gradient. In reality, the natural valve shows a complex fiber-reinforced composite texture to be able to bear diastolic pressures.

2.6.2 Pressure versus Time

The initial state of the model in the open configuration was assumed to be stress-free. Loads were gradually applied on this model up to diastolic pressures over a period of 0.5 seconds; from this closed, diastolic, stressed configuration, two full cardiac cycles were simulated, for a total simulation time of 1.79 seconds. At time 0.5 seconds the aorta pressure approximate 82mmHg.

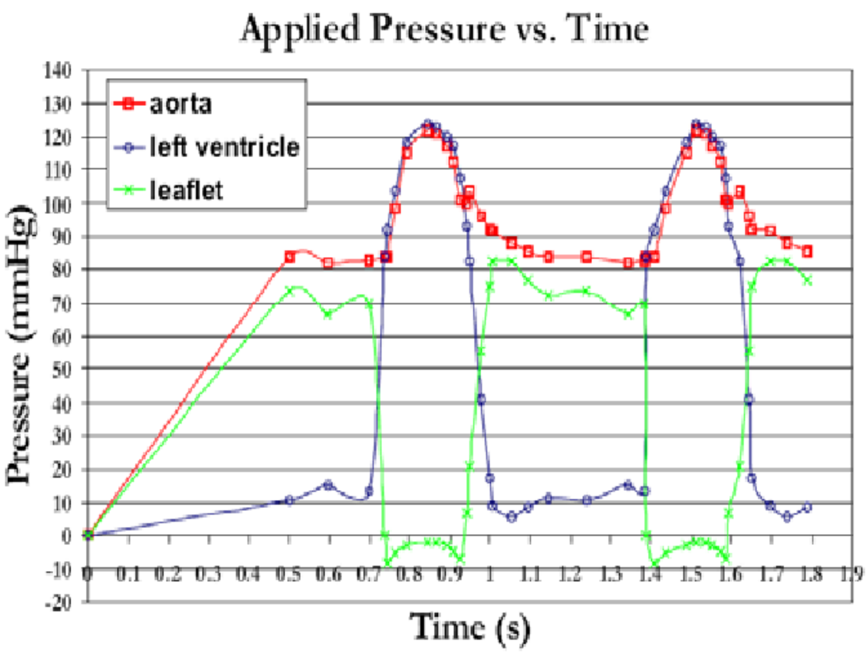


Figure 2.15: Graph Applied Pressure versus time

Source: Ranga et.al (2007)

2.6.3 Stress strain value

As can be seen in Figure 2.16, value of stress is change extremely rapidly during the rapid opening and closing phases. For model aortic valve, the results at 0.25 MPa stress will give value approximate 0.05 strains. The maximum strain value increases as much as 37.5% from the strain equal to 0.05.

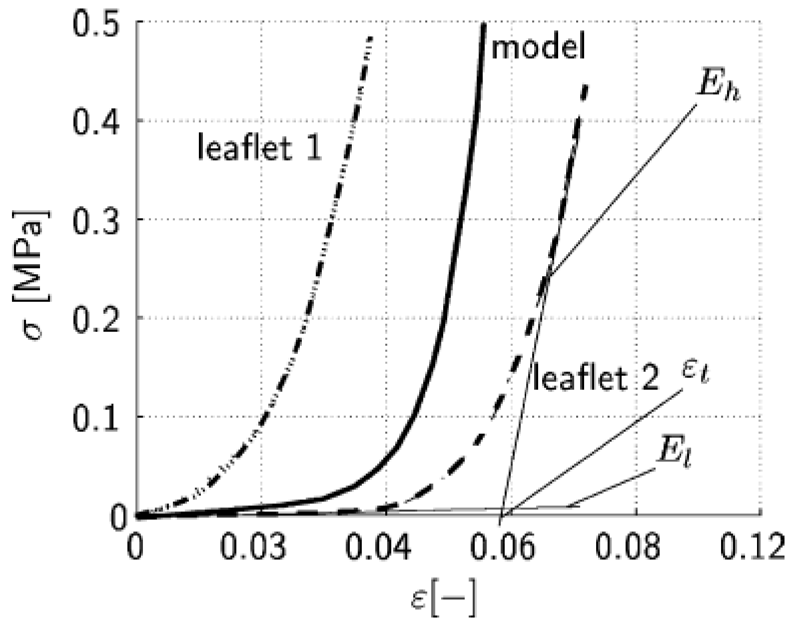


Figure 2.16: Graph Stress versus strain

Source: Hart et.al (2003)

2.6.4 Theory Statement

"The left side hearts have a greater workload because the blood needs to pump through the entire of body". Bender (2000)

2.7. Summary of Literature Review

In this project , the collection of data is most important to ensure the process to ensure the project in the right track. The summary of the literature review that involve in this project can be referred in the Table 2.1.

Table 2.1 : Summary of Literature Review

No.	Author	Summary
1.	Weinberg, E.J., and Mofrad, M.R.K. 2007.	This paper investigates about the development of prosthetic heart valves is primarily based on the time consuming trial-by-error concept involving many animal tests. The integration of computational analyses in aortic valve research is therefore very appealing as it is time and cost efficient. Moreover, these analyses enable to investigate performance characteristics, such as (shear) stresses, vertical flow and wash-out, which are difficult to evaluate experimentally. The mechanics, hemodynamics and kinematics of an aortic valve model have been studied using computational techniques, which incorporate the blood-wave interaction. The results can be used to define design criteria for improved (long-term) functional.
2.	Knierbein, B., Rosarius, N., Unger, A., Reul, H. and Rau, G. 1991.	The computer-supported development of valves for cardiac-assist devices or artificial hearts is shown in relation to plastic technology. A CAD-system is used for the design development, whereas the dimensioning of the critical and highly stressed membranes is examined by FEM-analyses. Economic manufacture is permitted by the combined thermoforming-dip molding technique; the blood-side components are made from biocompatible polyurethane to minimize blood damage.
3.	Hart, J. D., Peters, G.W.M., Schreurs, P.J.G., Baaijens, F.P.T. 2003.	A review of the effect of collagen fibers on the mechanics and hemodynamics of a trileaflet aortic valve contained in a rigid aortic root is investigated in a numerical analysis of the systolic phase. Collagen

		fibers are known to reduce stresses in the leaflets during diastole, but their role during systole has not been investigated in detail yet. It is demonstrated that also during systole these fibers substantially reduce stress in the leaflets and provide smoother opening and closing. Compared to isotropic leaflets, collagen reinforcement reduces the fluttering motion of the leaflets. Due to the exponential stress–strain behavior of collagen, the fibers have little influence on the initial phase of the valve opening, which occurs at low strains, and therefore have little impact on the transvalvular pressure drop.
4.	Ranga, A., R. Mongrain, R., Biadilah Y., and Cartier, R. 2007	This study was to investigate aortic valve is a compliant biological structure experiencing dynamic cyclic loading. The present study’s aim was to integrate three key physiologically important features into a realistic structural simulation of the aortic valve: compliance of the aortic root wall, non-linear material properties of the tissues, and dynamic loading. Compliance of the root dramatically changed the opening and closing shape and dynamics of the leaflets, altered the diastolic and systolic geometries and helped reduce stresses in the valve. Models of the aortic valve such as this one can provide an increasingly.Accurate tool in the investigation valve pathologies, device design and modeling robot-tissue interaction in assisted surgery.
5.	Astorino, M., Gerbeau, F.J., Pantz, O., and Traoré, K.F., 2008.	Present a partitioned procedure for fluid-structure interaction problems in which contacts among different deformable bodies can occur. A typical situation is the movement of a thin valve immersed in an

		incompressible viscous fluid In the proposed strategy the fluid and structure solvers are considered as independent “black-boxes” that exchange forces and displacements; the structure solvers are moreover not supposed to manage contact by themselves.
6.	Shadden, S.C., Astorino, M., and Gerbeau, J.F. *2010.*	The study on improving aortic valve prostheses that are able to adopt the physiological performance and durability of the healthy natural valve requires knowledge of the interaction between the valve and the blood. A short introduction is given on the morphology and functioning of the natural aortic valve system. The mechanical and hem dynamical complications of currently used substitute valves are summarized, expressing the strong need for numerical fluid-structure interaction models that are applicable to predict the functioning of existing and new prosthetic designs or to analyze the performance of diseased valves
7.	Lau, K.D., Zuccarini V.D. , Scambler, P., and Burriesci, G.,2011.	The effect of functional mitral regurgitation has been investigated in an anatomically sized, fluid– structure interaction mitral valve model, where simulated correction has been performed by applying: (1) edge-to-edge repair with annuloplasty and (2) edge-to-edge repair only. Initially defined in an open unstressed/corrected configuration, fluid–structure interaction simulations of diastole have been performed in a rigid ventricular volume.
8.	Bender J.R.,2003.	This book gives a simple review of the heart valves, the surgical repair, how valves work, complications of

		heart, heart valve disease, causes of heart valve disease.
9.	Sacks, M.S., Merryman, W.D., Schmidt, D.E., 2009.	The focus of the present review is on the functional biomechanics of heart valves.Thus, the focus of the presenter view is on functional biomechanics, referring primarily to bio solid as well as several key bio fluids mechanical aspects underlying heart valve physiological function. Specifically, we refer to the mechanical behaviors of the extracellular matrix structural proteins, underlying cellular function, and their integrated relation to the major aspects of valvular hemodynamic function.
10.	Espino, D.M., Watkins, M.A., Shepherd, D.E.T., Hukins, D.W.L., and Buchan K.G., 2006.	The mitral valve failure causes death if it is not corrected surgically. Such surgical repair can be improved by understanding mitral biomechanics. A two-dimensional FSI model of the mitral valve was generated, using an ALE mesh. A simple approximation of the heart geometry was used; the valve dimensions were based on measurements made. Leaflet deformations agreed with results from experiments in the literature and with our previous experimental results. In conclusion, a two-dimensional model of the mitral valve has been developed, and will be further developed to investigate heart valve failure and subsequent surgical repair.
11.	Weinberg, E.J., and Mofrad, M.R.K. 2008.	A set of multi-scale simulations has been created to examine the dynamic behaviour of the human aortic valve (AV) at the cell, tissue, and organ length scales. Each model is fully three-dimensional and includes appropriate nonlinear, anisotropic material models.

		The organ-scale model is a dynamic fluid structure interaction that predicts the motion of the blood, cusps, and aortic root throughout the full cycle of opening and closing. The cell-scale model predicts cellular deformations of individual cells within the cusps. Each simulation is verified against experimental data. The three simulations are linked: deformations from the organ-scale model are applied as boundary conditions to the tissue scale model, and the same is done between the tissue and cell scales. This set of simulations is a major advance in the study of the AV as it allows analysis of transient, three-dimensional behavior of the AV over the range of length scales from cell to organ.
12.	C.J. Carmody, G. Burriesci, I.C. Howard, E.A. Patterson,2004.	A pair of finite element models has been employed to study the interaction of blood flow with the operation of the aortic valve. A three-dimensional model of the left ventricle with applied wall displacements has been used to generate data for the spatially and time-varying blood velocity profile across the aortic aperture. These data have been used as the inlet loading conditions in a three-dimensional model of the aortic valve and its surrounding structures. Both models involve fluid–structure interaction and simulate the cardiac cycle as a dynamic event. Confidence in the models was obtained by comparison with data obtained in a pulse duplicator. The results show a circulatory flow being generated in the ventricle which produces a substantially axial flow through the aortic aperture. The aortic valve behaves in an essentially symmetric way under the action of this flow, so that the pressure difference across the leaflets is approximately uniform. This work supports the use of spatially uniform but temporally variable pressure

		distributions across the leaflets in dry or structural models of the aortic valves. The study is a major advance through its use of truly three-dimensional geometry, spatially non-uniform loading conditions for the valve leaflets and the successful modeling of progressive contact of the leaflets in a fluid environment.
13.	Benjamín González, B.,Benítez,H.,Rufino,K., Fernández, M,. and Waleska Echevarra,W.2008.	Heart valves all are prone to disease and malfunction, and can be replaced by prosthetic heart valves. The two main types of prosthetic heart valves are mechanical and bio prosthetic. The mechanical valve is excellent in terms of durability, but is hindered by its tendency to coagulate the blood. The bio prosthetic valve is less durable and must be replaced periodically. All valve types must be durable, because the body is an extremely hostile environment for a foreign object, including prosthetic heart valves. Today, engineers are researching new designs of prosthetic heart valves. They use the mechanical properties to make an artificial heart valve design. An artificial mitral valve is an option for humans with irreparable valve disease .

CHAPTER 3

METHODOLOGY

3.1 INTRODUCTION

Process simulation was fully simulated by ADINA-FSI application, not just only limited for human inside body analysis, but the coverage of this software is a comprehensive array of multiphasic capabilities including fluid-structure interaction Fluid and structural domains for mitral valve were discretized, respectively by 18825 nodes and 35410 triangular elements. Meanwhile the aortic valve elements are less 6823 number of elements and extra 603 nodes compare to the mitral valve. Simulation process was characterized by a fix density ρ= 1060 kg/m3 and value for an elastic modulus E, = 2 MPa and blood viscosity $\mu = 2.70 \times 10^{-3}$ Pa.s. The maximum blood pressure applied in the heart system via both of these valves equal 240×10^{3} dyns/cm^2 with time step size 50×10^{-3} to reach the peak systole. The simulation process study includes the following steps: first, import the models part of CFD applications, second, key in all the requires parameters, third, mesh generation process, the final step, manipulated setting parameters , in order to define the extent of differences blood flow pattern for both models.

Blood flows through the 3D model for mitral and aortic valves have been implemented by referring the structural parameters and geometry properties obtained from literature study. A 3D model of both valves without ventricle have been developed from Solidwork application technique. The process of simulation is assumed under steady condition flow and blood exhibits as Newtonian flow. Both of 3D simplified models have been simulated due to a time dependent flow and pressure variables are applied in the system during the systolic phase.

3.2 FLOW CHART

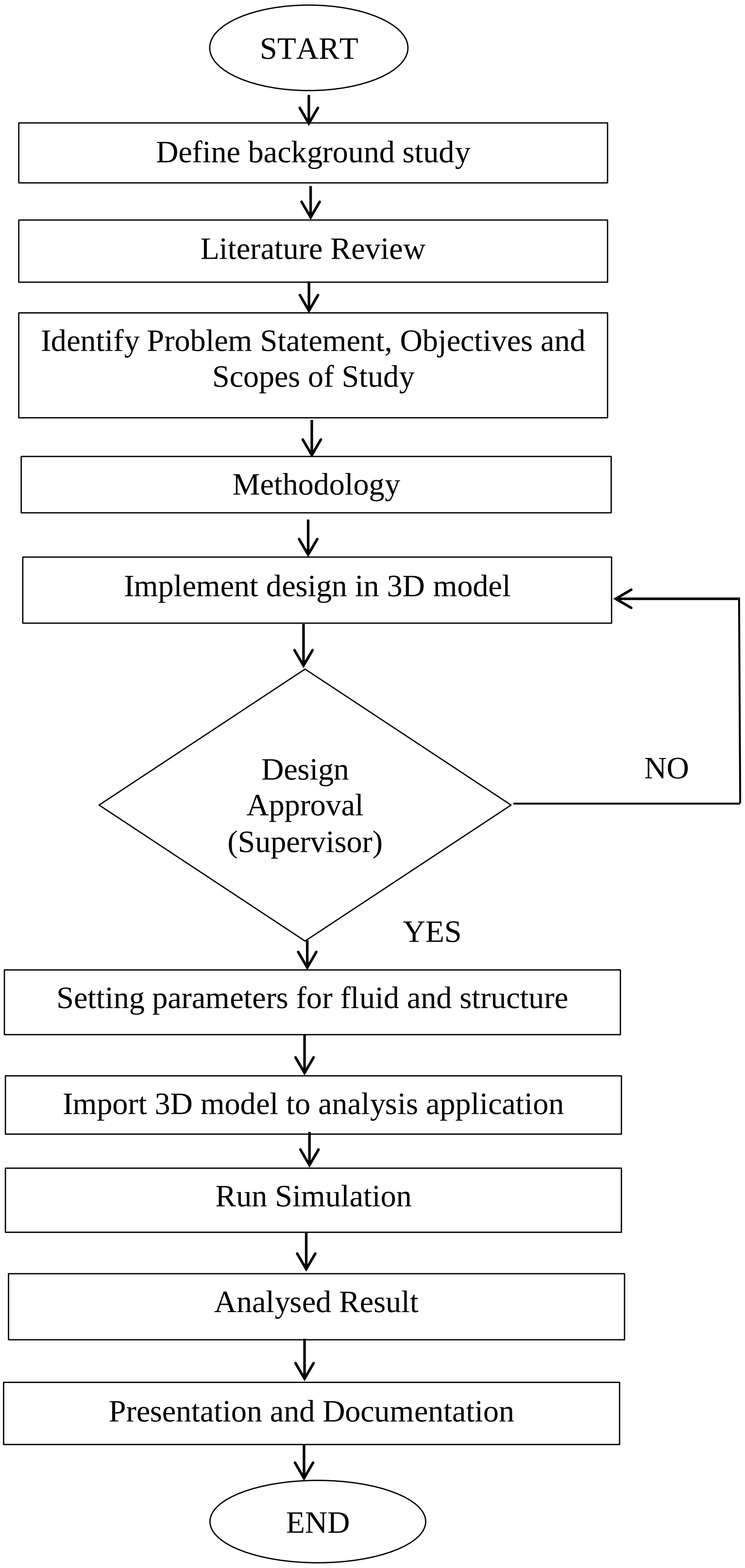


Figure 3.1: Flow Chart Project Study

Research study for project computational analysis of blood pattern in a 3D model of MV and AV started from defined the background study. For the background study, the major problems have been recognized clearly. To ensure the project always on the right track, the objectives need to be followed. Collect the literature review from the previous study for understanding purpose and references. All the literature reviews were involved the blood parameters, geometry design of both models, and validation of project study.

From obtained or reference geometry data, models of both valves have been implemented. The model design has been modeled in 3D view by using the SolidWork application. Next step, the approving design by supervisor need to be passed before the next process can be proceed..The flow process of project study proceeds by importing the valve model to the analysis application software. Analysis application that has been used in this project is ADINA-FSI. Setting the parameters based on the references from previous study.

The meshing process needs to be done completely for both valves before the simulation process able to run. Manipulate the data to obtain the different blood flow pattern through both valves. Then, validation process needs to be conducted by comparing the own results obtained in the previous study. Present the work project and completed the writing process for documentation purpose. The project ended by submitting the work project to the supervisor.

3.3 BLOOD PARAMETERS

Table 3.1: Blood Properties Parameters

Blood Properties	Value
Density (kg/m^3)	1060
Viscosity, Pa.s	2.70×10^{-3}
Cardiac Cycle, kPa	0 – 2.33

Source: Espino (2006)

Fluid parameters of human blood that have been used in this project obtained from the previous study. The blood flow was assumed in steady state condition and the pressure that will be used in this project was referred from the low pressure range to the highest value of pressure during systole phase.

3.4 PRE-SETUP SIMULATION PROCEDURE FOR STRUCTURE PART

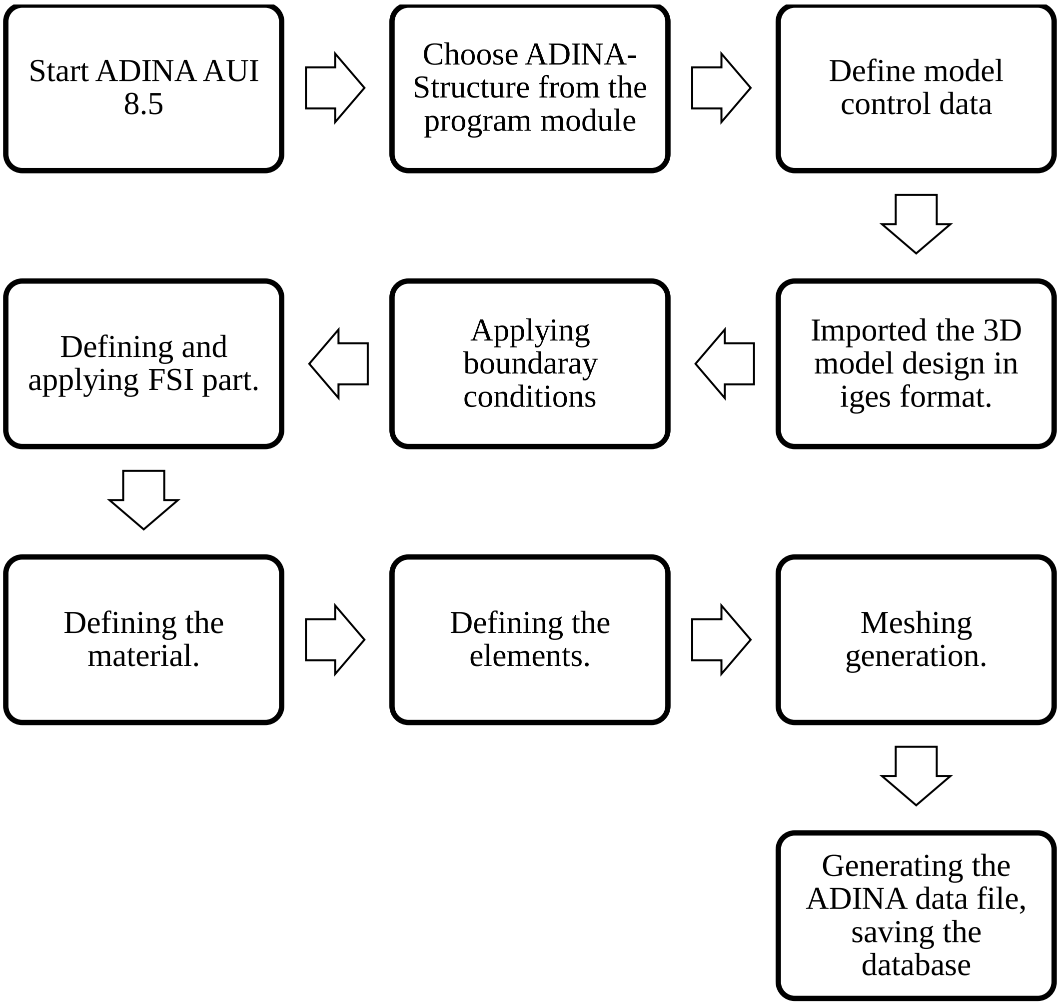


Figure 3.2 : Flowchart for simulation procedure for structure part

In FSI, it involved the combination of interaction between the structure and fluid. The steps for running the simulation in ADINA could be divided into two partitions, which are the structural parts and the fluid parts. From this, in ADINA-FSI, we will import the model design for fluid and structure model separately. In pre-setup for structure part, we initially open the ADINA-FSI software with the program module into ADINA-Structure mode. Then, the setup for model control data such as setup the FSI mode, analysis assumption is assumed and master degrees of freedom will be setup. Next, define the boundary condition for model design and fix wall boundary and the FSI boundary condition was applied. Then, defined the material properties for the leaflet by key in the value of the Young's modulus and Poisson's ratio. Next, performed the elements defining by defining the type of leaflet into 3D solid type. After that, meshing process was continued after the element was defined, by subdivided the structure with division lines and generated the mesh surface. Finally, the drawing is saved into an ADINA data file and it was complete for structure part.

3.5 PRE-SETUP SIMULATION PROCEDURE FOR FLUID PART

In the pre-preparation process in fluid simulation, the steps involved were nearly same step as the setups involved in structure. The steps were started with open the ADINA-CFD module and defined the model control data. Then FSI analysis is chosen, flow assumption was assumed, time steps and time function is determined. Next, model geometry built into SolidWork applications before imported into the ADINA applications.. The next step was proceeded by defining the material properties such as density and viscosity of fluid. After this, it proceeded by defining the boundary condition with no-slip boundary condition and FSI boundary condition.

A loading that functions as the normal traction was setup for applied the pressure difference. The value of normal traction was referred from the literature review of the cardiac cycle. Then, defined the element group by adding the type of 3D fluid. Meshing was continued after defining the element group by subdividing model fluid design.. Triangular cells with zero skewness is the type of mesh was applied due to the numbers of nodes for simple design of model can provide better quality solutions for complicated design.. It reduced false diffusion when the mesh is aligned with the flow. Meshing was performed after a model has completely selected. Finally, after the meshing step, the drawing could be saved into ADINA-CFD database file.

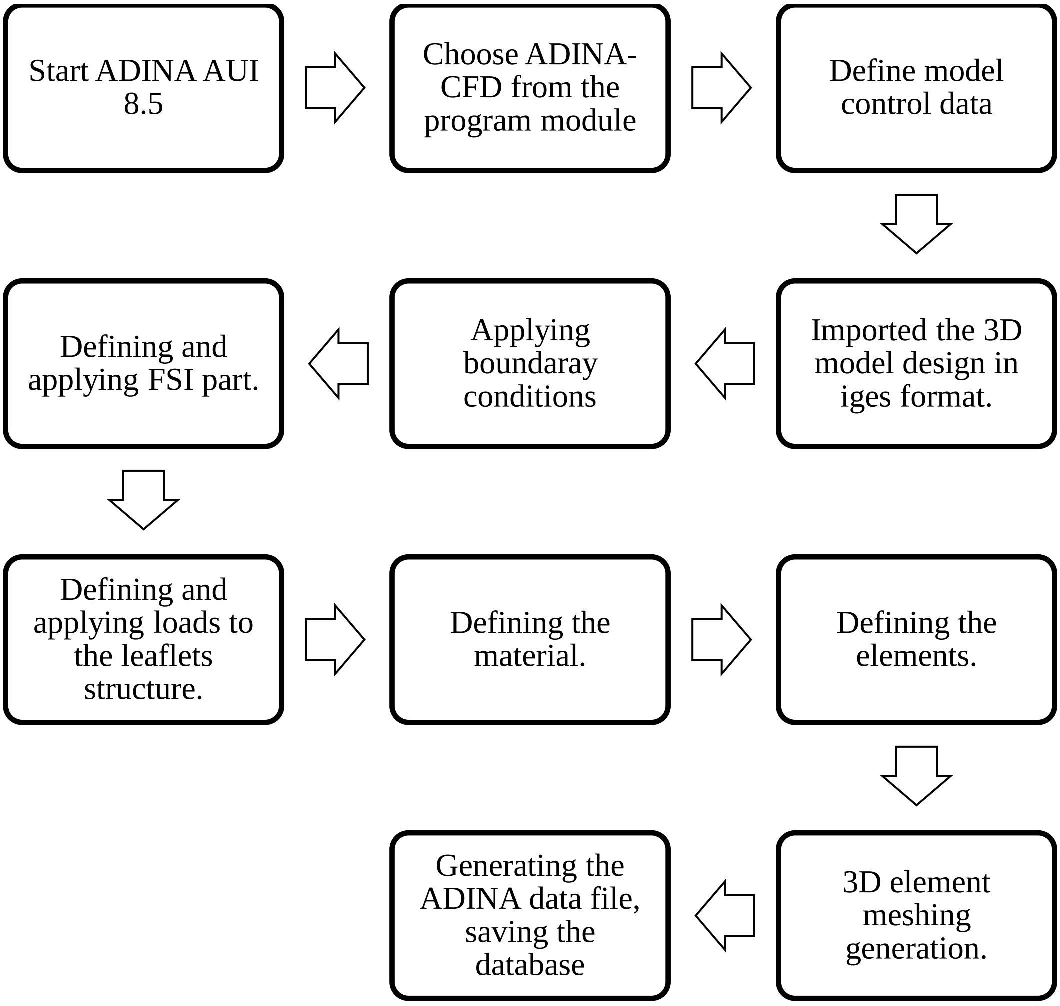


Figure 3.3 : Flowchart for simulation procedure for fluid part

3.6 MITRAL VALVE 3D MODEL

The three dimensional of heart valve models constructed using SolidWorks applications In this project, fluid model and structure model need to be imported from SolidWork application in iges format. The fluid model is entered as an ADINA-F model and the solid model is entered as an ADINA model. However the analysis itself is performed as a fully coupled analysis using ADINA-FSI. Invoke the ADINA Structures from the program module drop down list.

Pressure constraints were applied to the fluid and structure boundary. Interaction between leaflets structural deformation and fluid dynamics are determined simultaneously.

3.6.1 Geometry and Parameters Mitral Valve

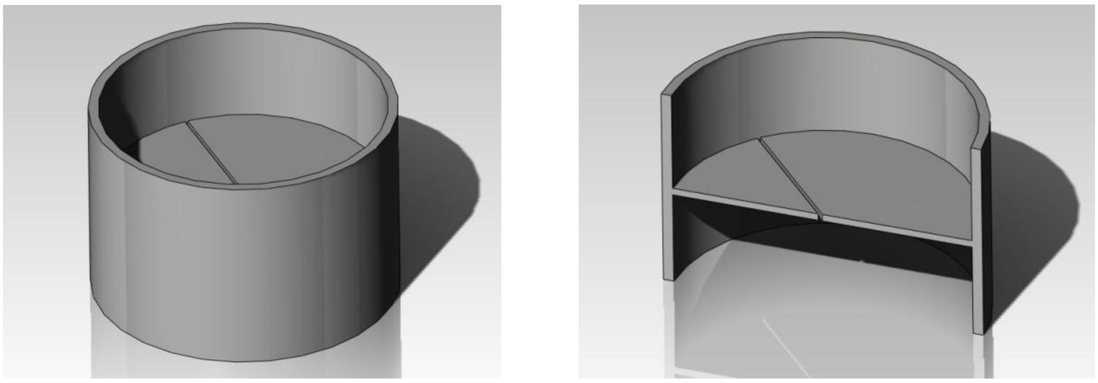
The Mitral Valve consists of two flexible leaflets and the geometric dimensions of the valve are described by a set of value parameters such in Table 3.1. The mitral valve model has been implemented in this research study is without ventricle. This simplified of models have assigned to the 1mm wall thickness. A region in the structure where contact was likely to occur were identified from past experience, e.g. Kunzelman (2007) and Espino (2006).

Table 3.2: Mitral Valve Properties Parameters

Parameter	Dimension
Normal Mitral Valve Thickness , mm	0.7
Mitral Valve Leaflet density, kg/m^3	1.06×10^3
Leaflet Poisson's ratio	0.49
Normal Diameter Mitral Valve, mm	30
Normal Height Mitral Valve, mm	20
Young Modulus,MPa	2

Source: Espino (2006)

Mitral Valve 3D model design has been built without ventricle. The geometry of the mitral valve was referred from the literature review and previous data from previous researcher. Mitral Valve 3D model consisted of two leaflets as shown in Figure 3.4 with the thickness is 0.7 mm. The Mitral Valve leaflets part has been designed flatted with horizontal axis. In this research project, two parts need to be designed which ease structures part and fluid part for analysis purpose.



(a) (b)

(c)

Figure 3.4: (a) Mitral valve structure design, (b) Mitral valve sectional view , (c) Mitral valve fluid design

The structure design was totally followed the dimensions and specifications from the previous study. However the fluid design has been implemented by using the inner space size of the solid design. The mitral valve fluid design purpose as the medium for fluid movement illustrates during the simulation process. The full dimensions of the Mitral Valve model can be referred in appendix C1-C6.

3.6.2 Mitral Valve Structure Procedure

Figure A1-1 until A1-8 shown in appendix A1 represents the mitral valve structure procedure. After the solid model successfully imported from SolidWork application, the next step in analysis solid model is by defining the model control data. Enter the heading for analysis and choose the degree of freedom. Uncheck the X-Rotation, Y- Rotation, Z- Rotation axis such in Figure A1-2.

In the meshing process created the element density Figure A1-3 for the model need to be determined. The size of the elements for mesh model has been determined by choosing the type of length density. The smaller size of element meshes gives more clear of illustration simulation and accurate result. For this project, the triangular elements have been selected as mesh elements. To ensure the definition of boundary condition process on a right track, the next process needs to in order step. Select the proper boundary conditions for the mitral valve by fixed the outer body of the Mitral Valve wall such as in Figure A1-4

Then define the fluid structure-interaction (FSI) for solid structure such illustrated in Figure A1-5. Choose the correct faces for FSI because when wrong FSI selected, it will ruin the simulation process. Next, define the element group and choose 3D solid type. The isotropic linear elastic material has been chosen while the properties of Young Modulus and poison ratio value are referred from past studies. After all the data already key in, save the project in data solution for ADINA and save the database file.

3.6.3 Mitral Valve Fluid Procedure

The process to create the mitral valve into ADINA-F is almost same with the solid procedures. The differences between of this procedure are the material type that's been used, the loads apply to the model and time setting for fluid simulation progress. Normal traction has been applied at the inlet channel of model or the starting point for fluid flow, Define the Normal Traction dialog box, add Normal traction 1 and set the magnitude 1 Figure A1-7 For setting up the time step and time function, it's important for ADINA-F because it's showing the time process for simulation. In this project, the simulation will be run in 1 second with 20 numbers of steps.

After all the procedure followed, save the project in data solution for ADIN-FA and save the database file. To run the ADINA-FSI, choose the solid part (ADINA) and fluid model (ADINA-F) together and click start. When ADINA-FSI finishes, closed all open dialog boxes and choose Post Processing from the Program Module drop down list. Select the properties (Velocity/Effective Stress/Strain/Nodal Pressure) for simulated process. The clear interaction between structure and fluid can be illustrated clearly and data results were obtained from the application. Record, tabled, make a graph and analyze the result obtained from the simulation.

3.7 AORTIC VALVE 3D MODEL

The AV model has been implemented without ventricle and the dimensions of AV fully referred from past studies. The view model reference design of AV as showed in Figure 3.5.

3.7.1 Geometry and Parameters Aortic Valve

The specifications and dimensions of the aortic valve model listed in Table 2. The basic design of aortic valve consisting of three leaflets at 120° angles. The modeling and simulation of the interaction between the blood and the valve are challenging due to the complex valve dynamics, possible contact among the valve leaflets, intense velocity and pressure gradients . For FSI purpose, the model will be divided into two parts which ease for structure model and fluid model. Structure and fluid model were shown in Figure 3.6 (a),(b) and (c).

Table 3.3: Parameters of anatomy aortic valve

Parameter	Dimension
Normal Aortic Valve Thickness, mm	0.3
Aortic Valve Diameter,mm	23.55
Shell Thickness,mm	1

Source: Knierbien et. al (1991)

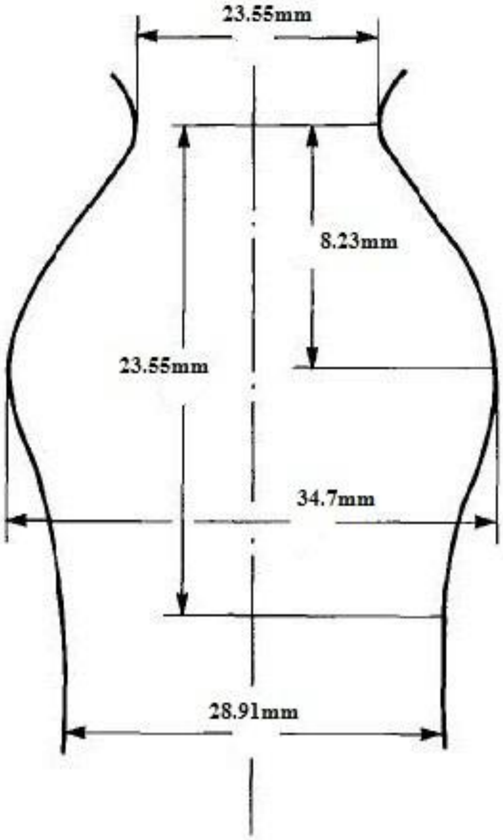
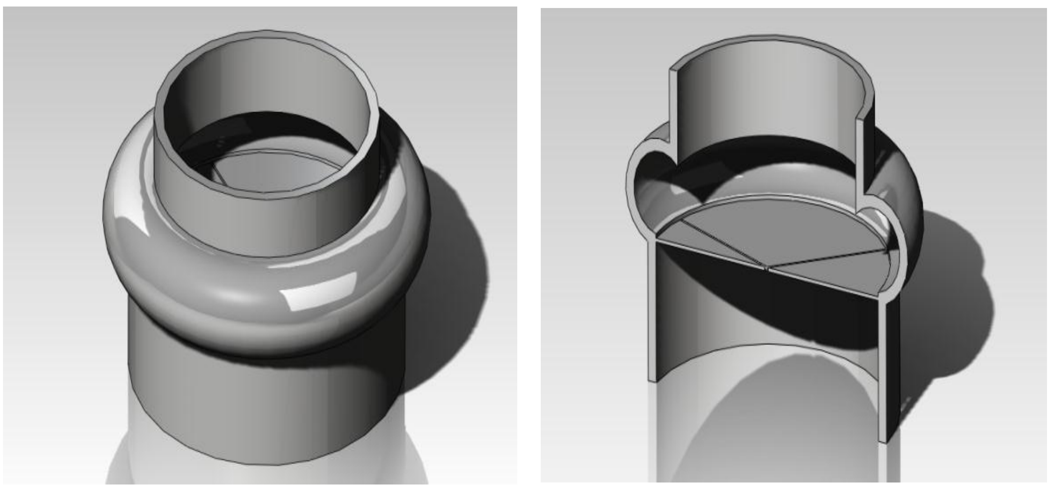
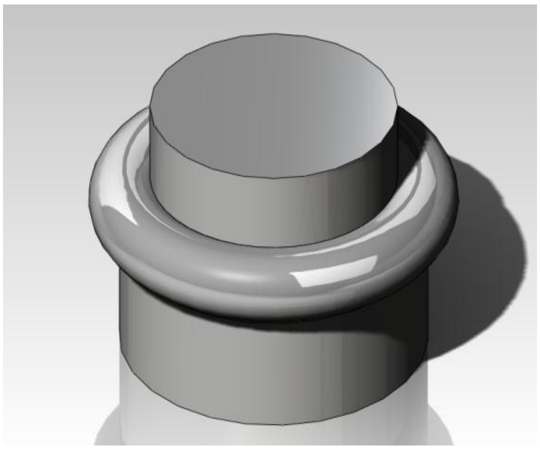


Figure 3.5: Geometric parameters of the aortic valve housing model

Source: Knierbien et.al (1991)



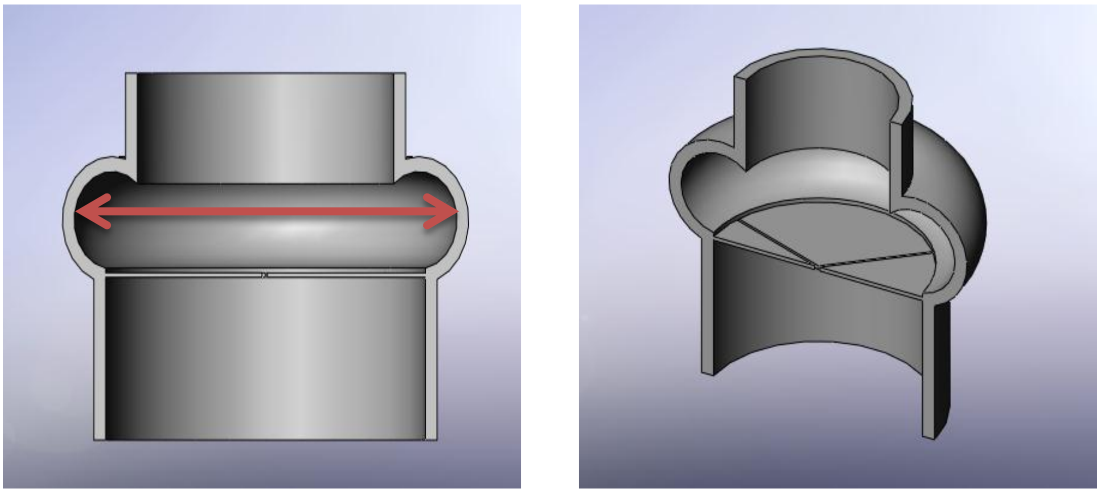
(a) (b)



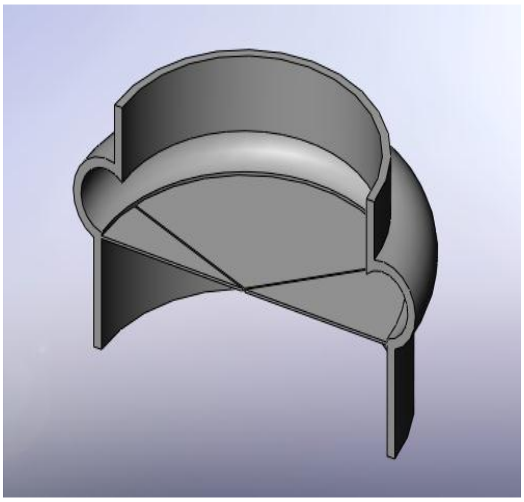
(c)

Figure 3.6: (a) Aortic valve structure design, (b) Aortic valve sectional view ,(c) Aortic valve fluid design,

The Aortic Valve model has been implemented by applications SolidWorks. For this model, the different size of AV sinus has been constructed. The curve at the AV as showed in Figure 3.7(a) called sinus of the valve. The smaller size of AV sinus that has been implement have a 10 % smaller than the original size of AV reference and in larger size of the sinus, the AV has been constructed 10 % more large.



(a) (b)



(c)

Figure 3.7 : (a) Sinus of Valve, (b) Small size of AV sinuous design, (c) Large size of AV sinuous design

3.7.2 Aortic Valve Structure Procedure

Figure A2-1 until A2-2 shown in appendix represents the procedure for aortic valve structure. The procedures to create the simulation analysis of aortic valve structure are totally same with the procedures of mitral valve structure. The difference for both structure model steps is selecting the FSI boundary condition such in Figure A2-1. .Key in the parameters is involved in the studies then save the project in data solution for ADINA and save the database file.

3.7.3 Aortic Valve Fluid Procedure

ADINA-F procedure for aortic valve model is almost same with the solid model procedure. The differences between of this procedure are the material type that's been used, the loads apply to the model and time setting for fluid simulation progress. Normal traction has been applied at the inlet channel of model or the starting point for fluid flow, Define the Normal Traction dialog box, add Normal traction 1 and set the magnitude 1 as represented in A2-2.For setting up the time step and time function, it’s important for ADINA-F because it’s showing the time process for simulation. In this project, the simulation will be run in 1 second with 20 numbers of steps.

After all the procedure followed, save the project in data solution for ADIN-FA and save the database file. To run the ADINA-FSI, choose the solid part (ADINA) and fluid model (ADINA-F) together and click start. When ADINA-FSI finishes, closed all open dialog boxes and choose Post Processing from the Program Module drop down list. Select the properties for velocity,Effective Stress,Strain and Nodal Pressure for simulated process. The clear interaction between structure and fluid can be illustrated clearly and data results were obtained from the application. Record, tabled, make a graph and analyze the result obtained from the simulation.

CHAPTER 4

RESULTS AND DISCUSSION

4.1 INTRODUCTION

Blood flows through the 3D model for mitral and aortic valves have been implemented by referring the structural parameters and geometry properties obtained from literature study. A 3D model of both valves without ventricle have been developed from Solidwork sapplication technique. The process of simulation is assumed under steady condition flow and blood exhibits as Newtonian flow. Both of 3D simplified models have been simulated due to a time dependent flow and pressure variables are applied in the system during the systolic phase. Process simulation was fully simulated by ADINA-FSI application, not just only limited for human inside body analysis, but the coverage of this software is a comprehensive array of multiphasic capabilities including fluid-structure interaction Fluid and structural domains for mitral valve were discretized, respectively by 18825 nodes and 35410 triangular elements. Meanwhile the aortic valve elements are less 6823 number of elements and extra 603 nodes compare to the mitral valve. Simulation process was characterized by a fix density $\rho = 1060\ kg/m^3$ an elastic modulus E, = 2 MPa and $\mu = 2.70 \times 10^{-3}$ Pa.s. for blood viscosity .The maximum blood pressure applied in the heart system via both of these valves equal $240x10^3\ dyns/cm^2$ with time step size $50x10^{-3}$ to reach the peak systole. The simulation process study includes the following steps: first, import the models part to CFD applications, second, key in all the requires parameters, third, mesh generation process, the final step, manipulated setting parameters , in order to define the extent of differences blood flow pattern for both models.

4.2 VALIDATION OF RESULT

4.2.1 Flow Pattern inside 3D Valve model

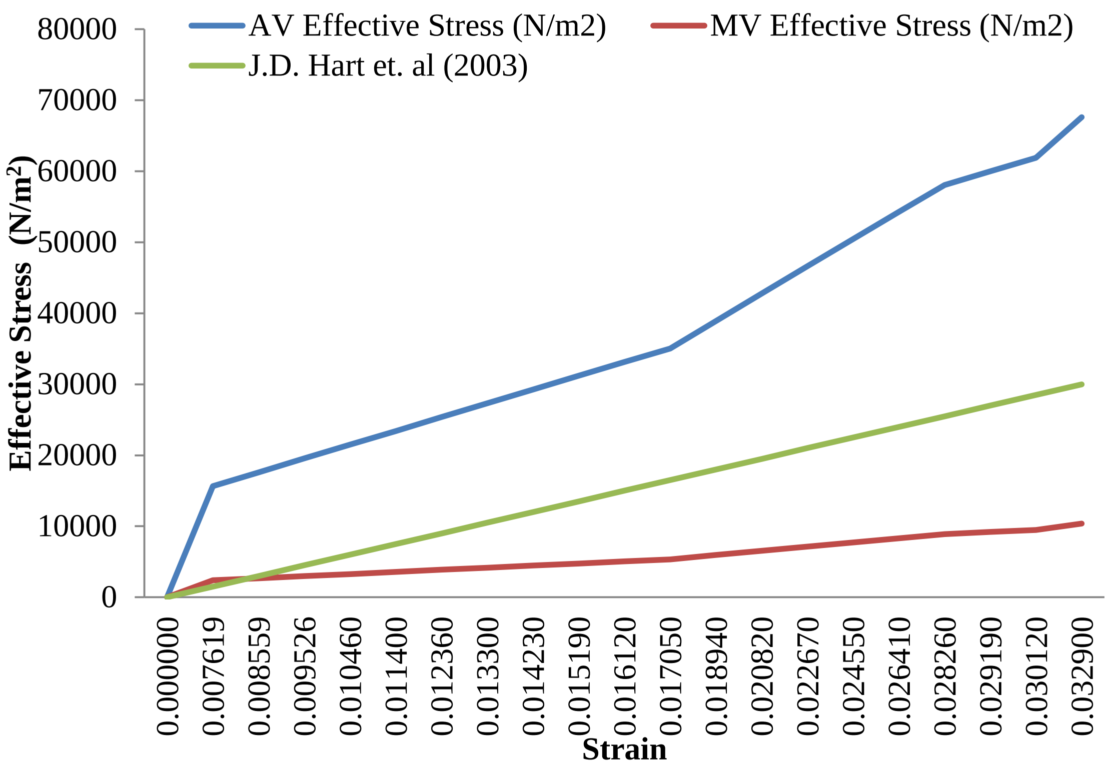


Figure 4.1: Graph Effective Stress versus strain value comparison for both valves

According to previous study by Hart et. al (2003) from the Journal Collagen fibers reduce stresses and stabilize motion of aortic valve leaflets during systole. The effective stress versus strain within time 1 second showed as in the graph above. Referring to the analysis results, the trend of effective stress for both model AV and MV is increase positively and proportionally with the increasing value of strain.

4.2.2 Pressure versus Time

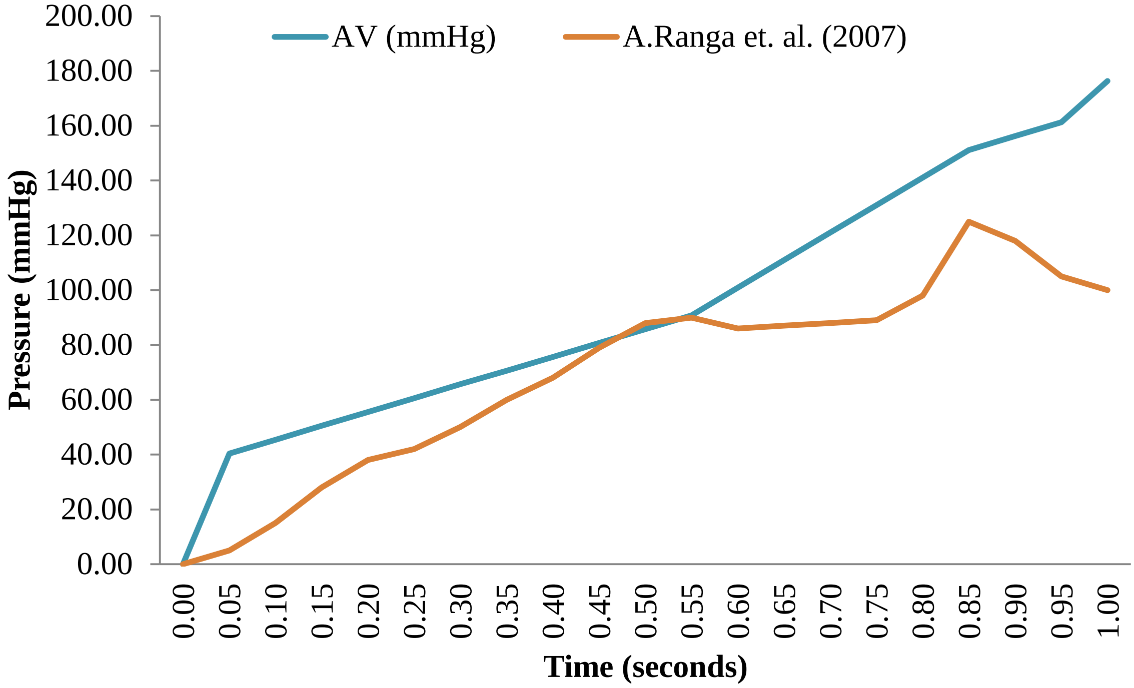


Figure 4.2: Graph Comparison aortic valve model pressure with previous study results

From the journal “A compliant dynamic FEA model of the aortic valve” by Ranga et al (2007) showed the pressure value acting on the aortic valve system. According to the results,at time stage 0.5 seconds the results simulation pressure approximate same with the A.Ranga results. From the analysis, this level showed the normal pressure act in the human circulatory system (80 mmHg- 120 mmHg.)

4.2.3 Theory Statement

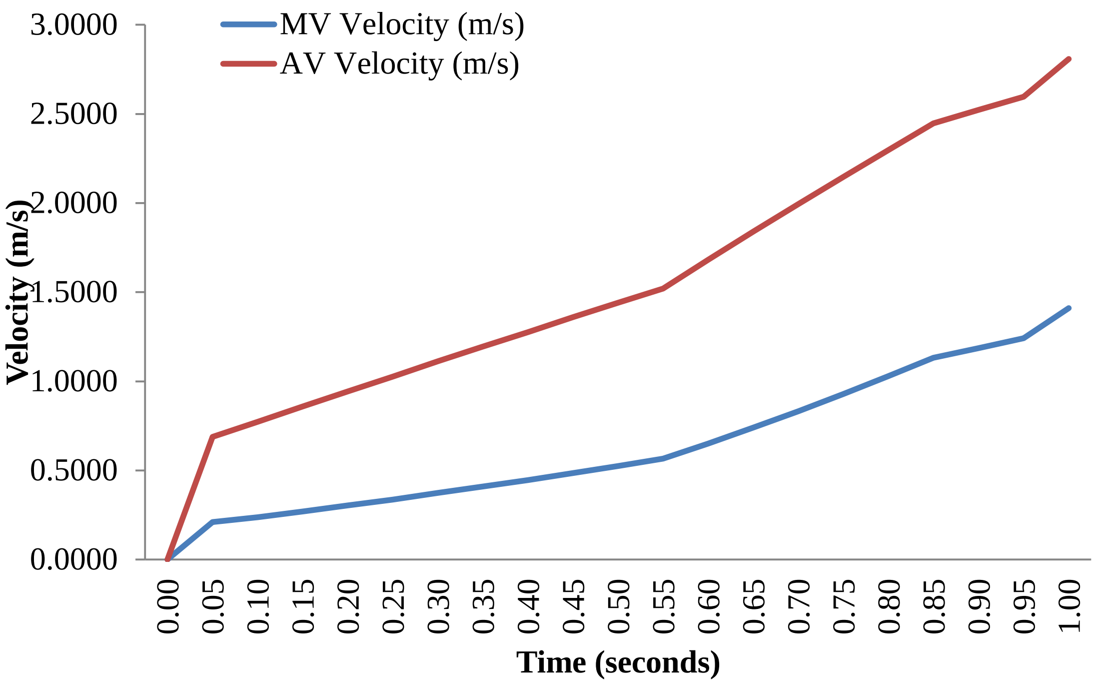
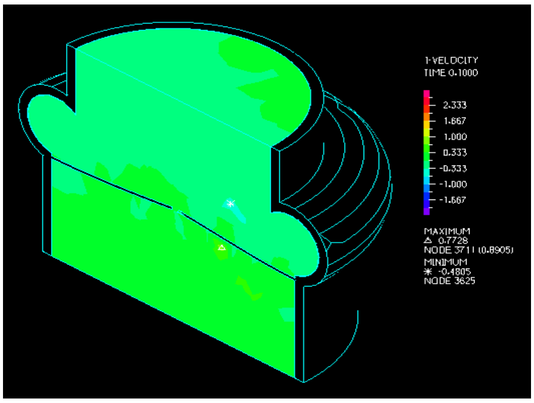


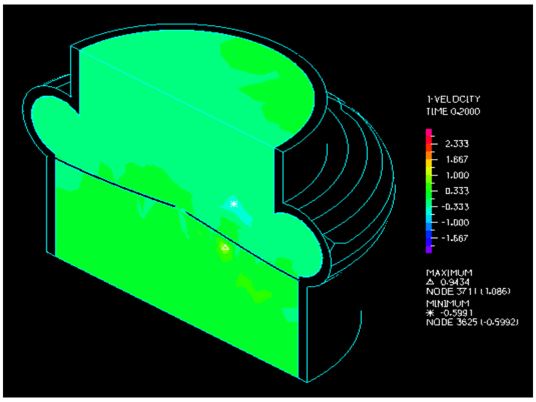
Figure 4.3: Graph Comparison velocity for Aortic and Mitral model

From the statement from J.F. Bender from his books Heart Valve Disease. Yale University School of Medicine Heart Book page 167, “The left side hearts have a greater workload because the blood needs to pump through the entire of body”. From this simulation results, it showed the AV velocity recorded higher velocity compared to the MV velocity.

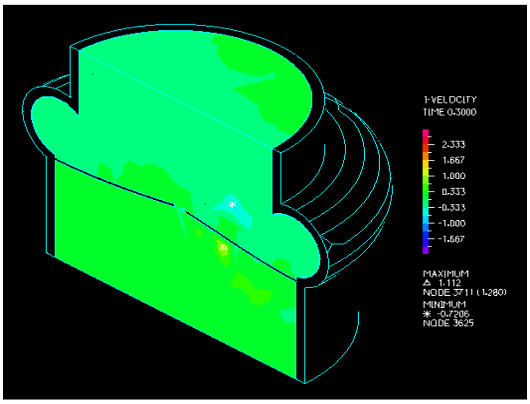
4.3 RESULTS FOR AORTIC VALVE

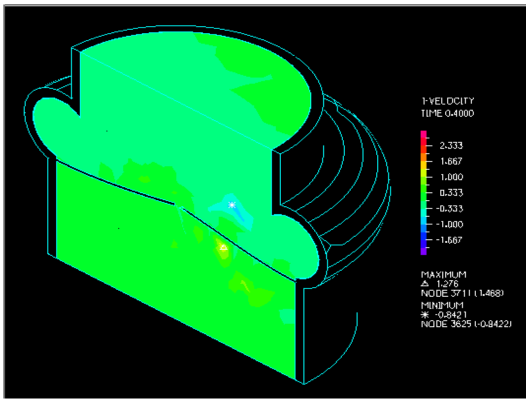
4.3.1 Velocity Analysis (Normal Size of Sinus)



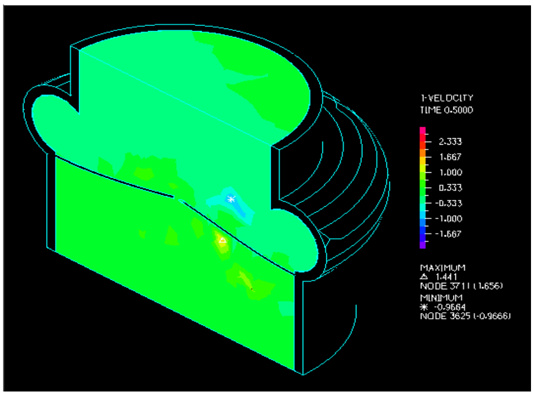


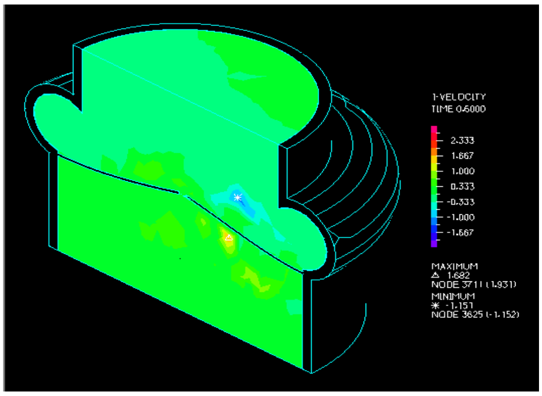
(a) (b)



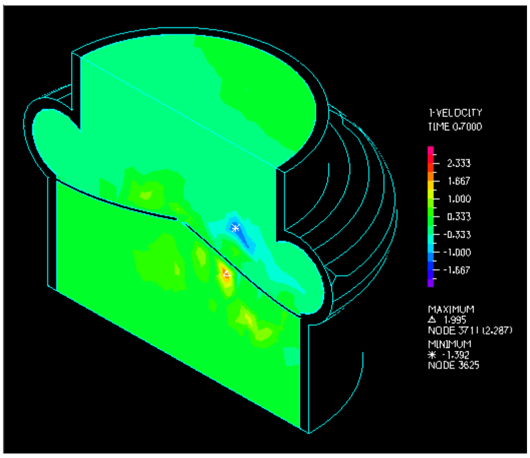


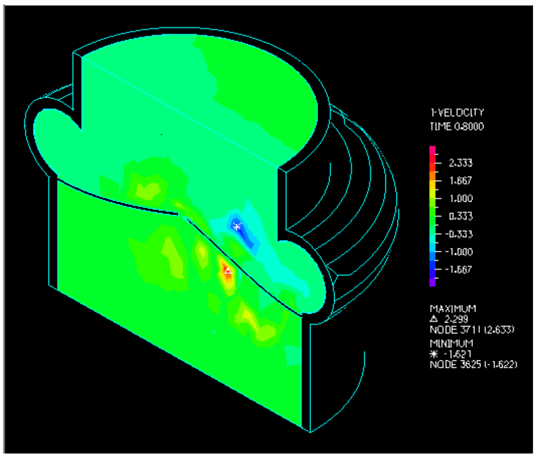
(c) (d)



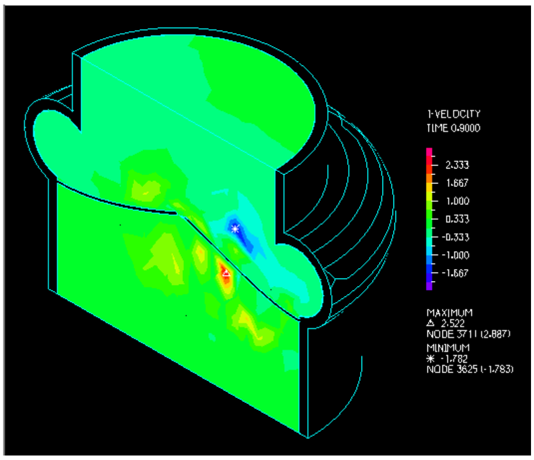


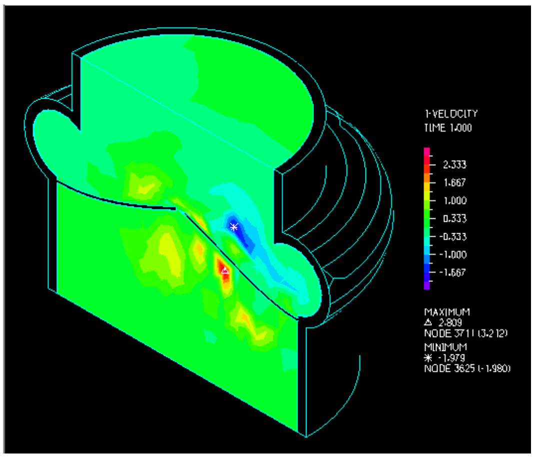
(e) (f)





(g) (h)





(i) (j)

Figure 4.4: Velocity analysis for normal size of AV sinus at time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j 1.00 Sec

At the initial condition of velocity flow pattern in the aortic valve, no changes or movement happens to the leaflets. During this time, no pressure flow via this valve, but when the pressure entered the valve region at time 1 second with pressure value 5.99 kPa, the velocity recorded 0.7728 m/s. By referring the sequence of figures, the deformation of leaflets is linear proportion to the pressure given into the aortic valve. Next, the changes in velocity flow pattern in this simulation were represented by the colors .The different colors represented different value of velocity. The increasing of pressure as much as 47.1 %, the velocity recorded 1.441 m/s at time 0.5 seconds because at this moment, the pressure of cardiac cycle influence the rate of velocity in the system. In this condition, the opening of the leaflets is bigger than before. The maximum velocity obtained 2.809 m/s at time 1 second, with increasing of pressure value starting from initial condition as much as 18.66 kPa.

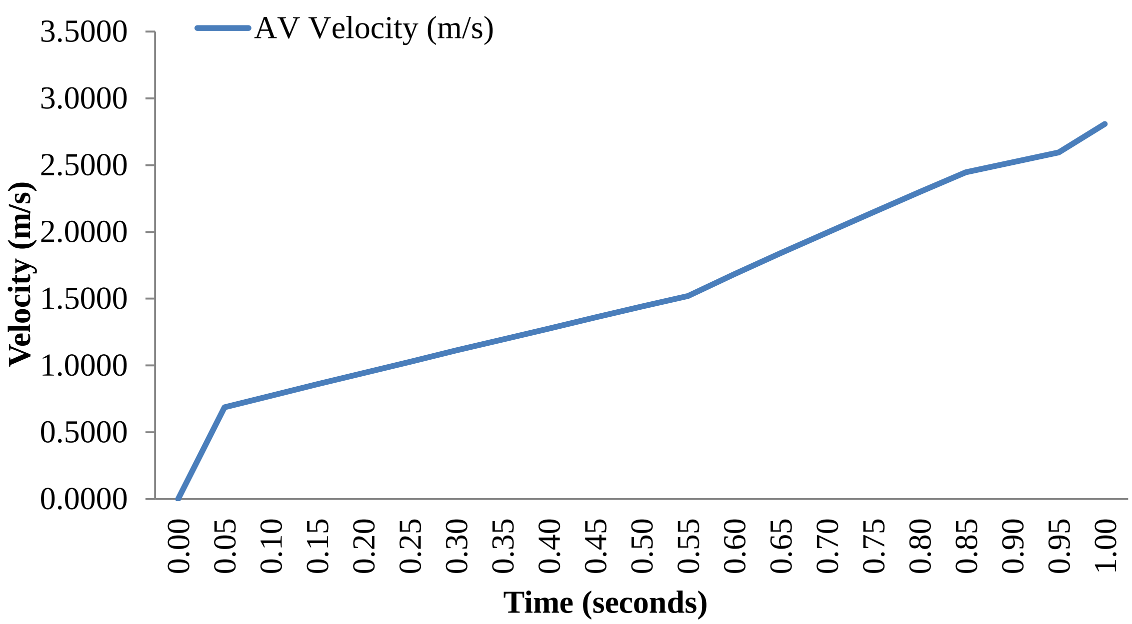
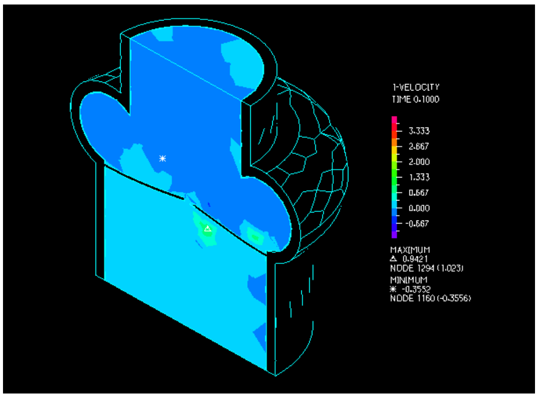


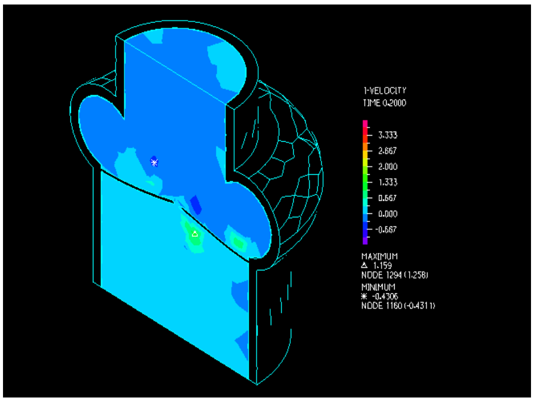
Figure 4.5: Graph Maximum velocity versus time for aortic valve

The graph is a linear correlation within time 1 second. The lowest velocity recorded when the pressure give into the aortic valve is equal to 0.6881m/s. From graph analysis, the velocity value is increased when pressure acting inside the aortic valve increased during the systole phase. The velocity increased as much as 52.22 percent with different time 0.45 seconds.The maximum velocity is equal 2.8090 m/s when the working pressure is 23.99 kPa.The detailed of velocity value within time 1 second can be referred in Table 4.1 in appendix D1.

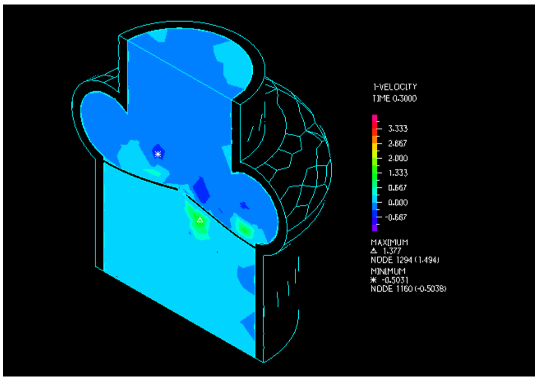
4.3.2 Velocity Analysis (Small Size of Sinus)



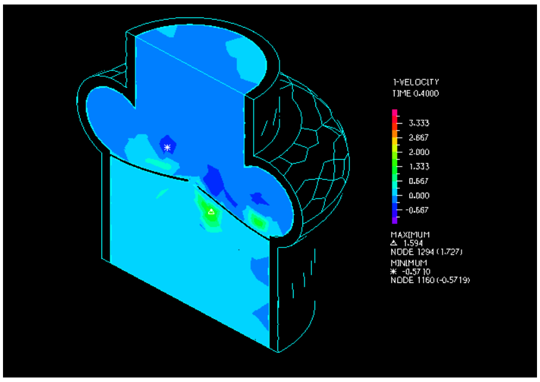
(a)



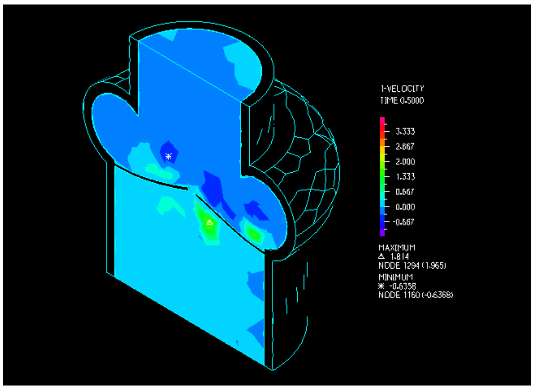
(b)



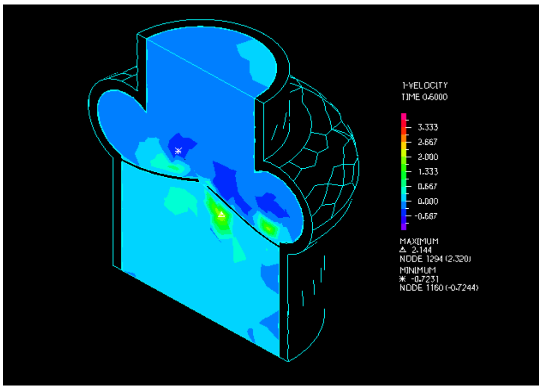
(c)



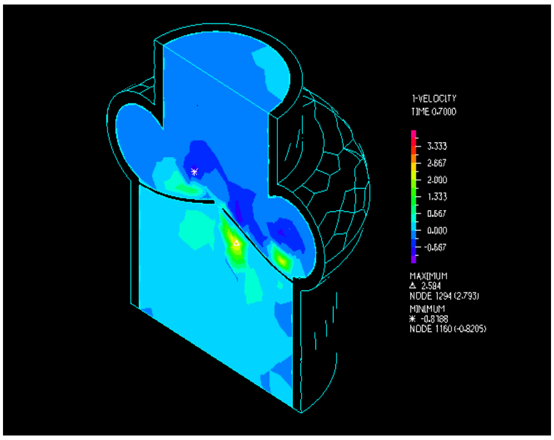
(d)

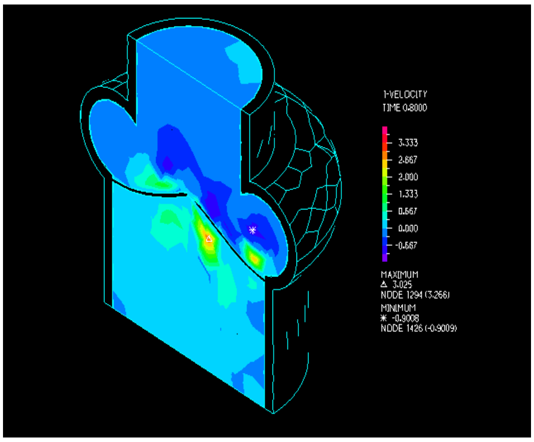


(e)

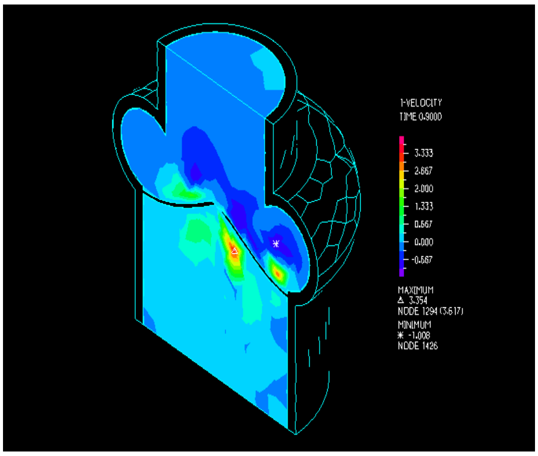


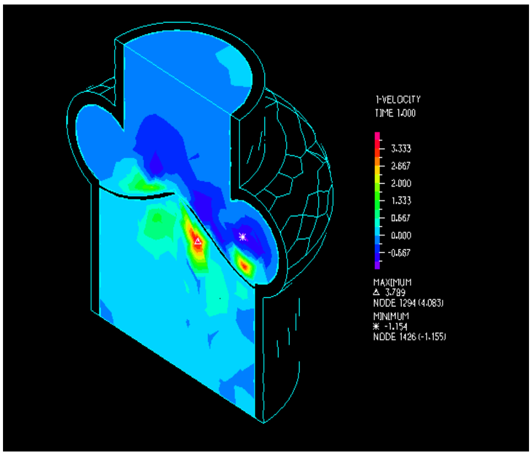
(f)





(g) (h)





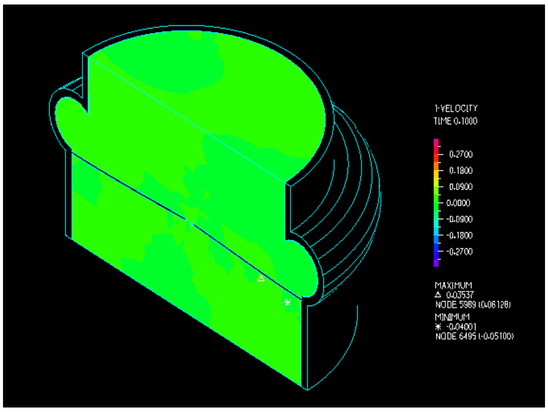
(i) (j)

Figure 4.6: Velocity analysis for small size of AV sinus at time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j 1.00 Sec

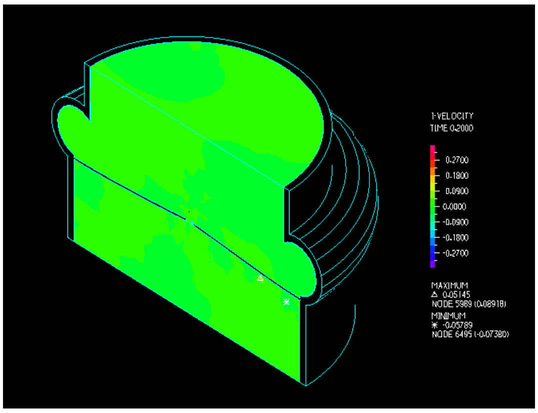
Regarding to the illustration of velocity inside the aortic valve with the small size of the sinus, the initial velocity was recorded 0.02027 m/s. The initial stage measured from opening state of leaflets due to the initial pressure act through the leaflets. As can see in the illustration, the leaflets deformation were proportionally with the working pressure. At 0.5 seconds time stage, the velocity has increased as much as

0.15393 m/s from the initial velocity. The maximum pressure acting inside the aortic valve gives high value of velocity 0.3605 m/s.

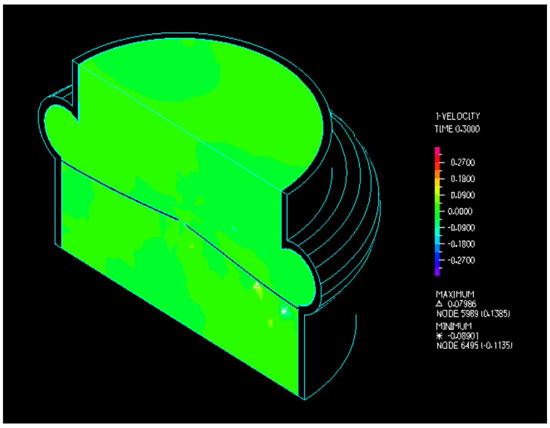
4.3.3 Velocity Analysis (Large Size of Sinus)



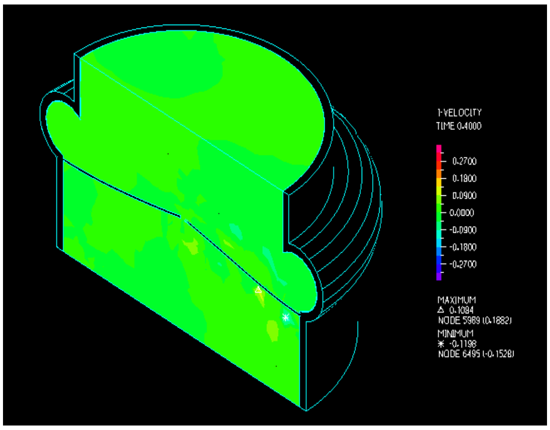
(a)



(b)



(c)



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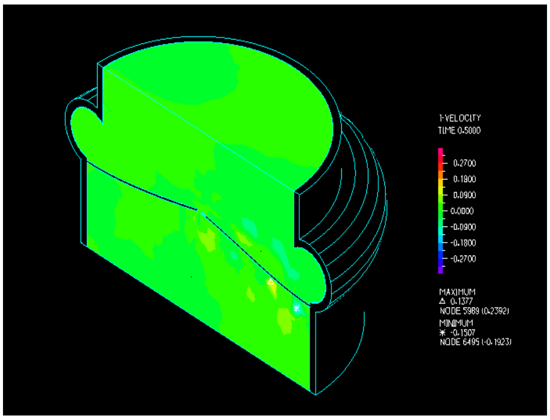

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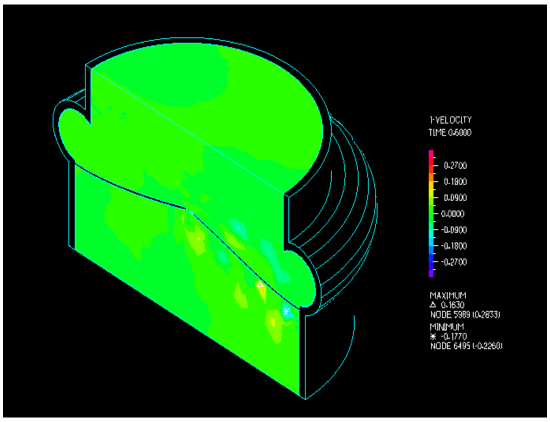
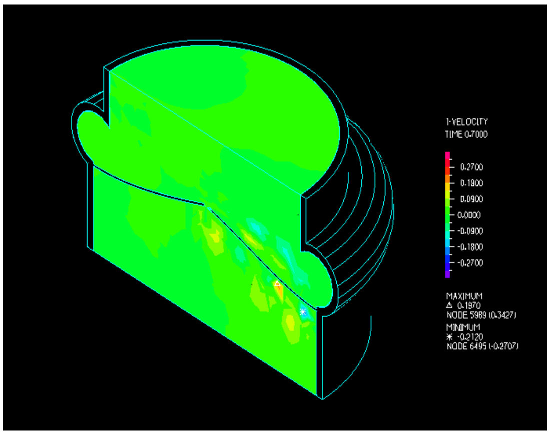


(f)



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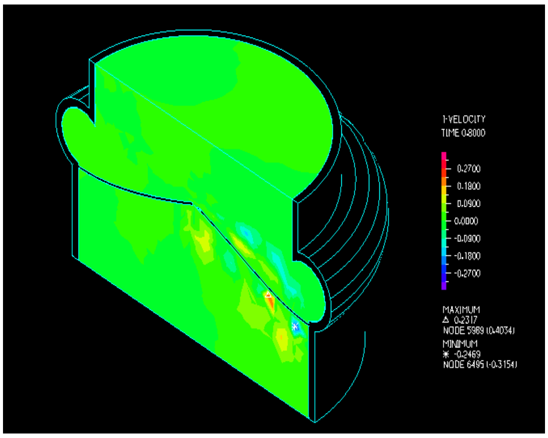
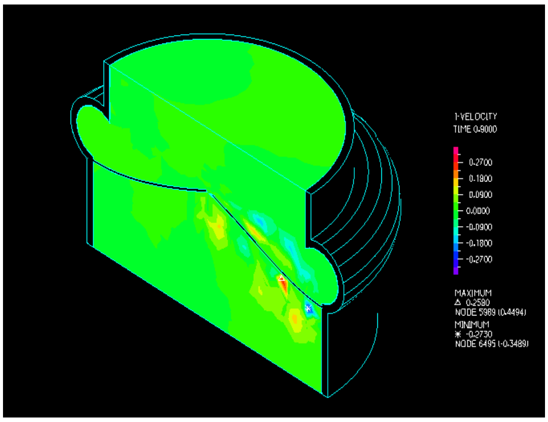
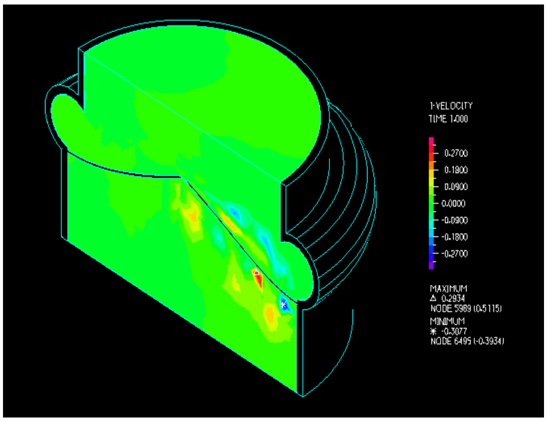


(h)



(i)



(j)

Figure 4.7: Velocity analysis for large size of AV sinus at time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j 1.00 Sec

From the sequences of simulation velocity, the pattern of velocity is clearly illustrated. The minimum velocity recorded at the pressure range below than 40 mm Hg is 0.01564 m/s. The velocity in this system has increased as much the increasing of pressure. The differences of this velocity at time 0.5 seconds at the initial stage is 0.12206 m/s. The maximum value for the velocity in the time range 1 second is increased as much as 53.51 percent from the opening leaflets deformation.

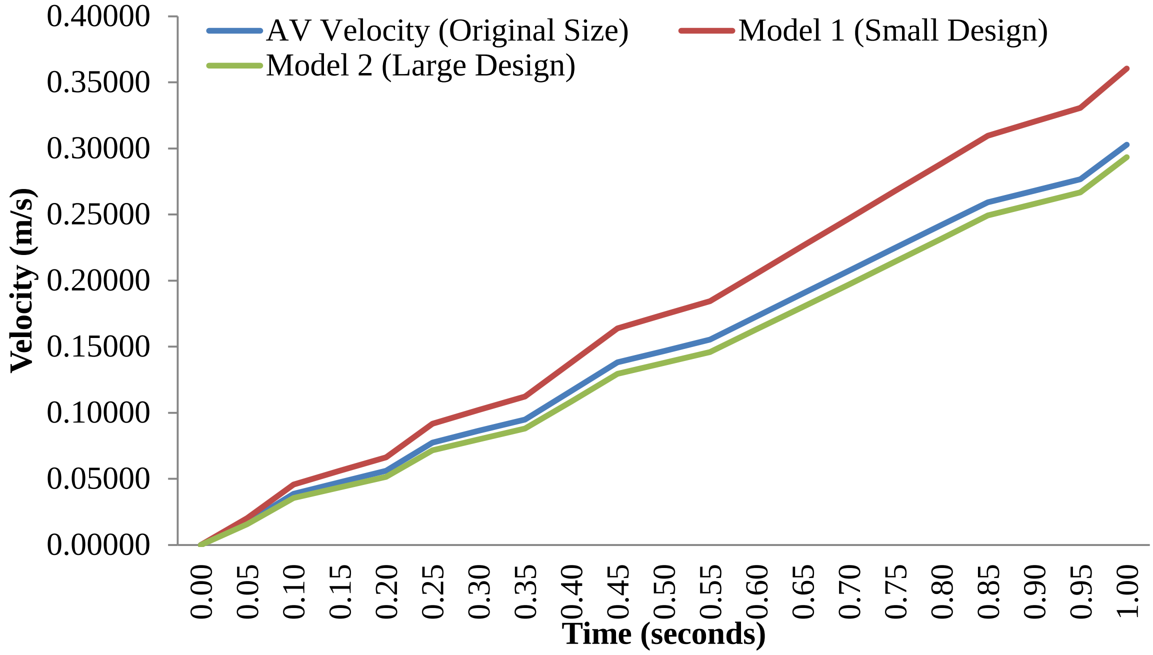
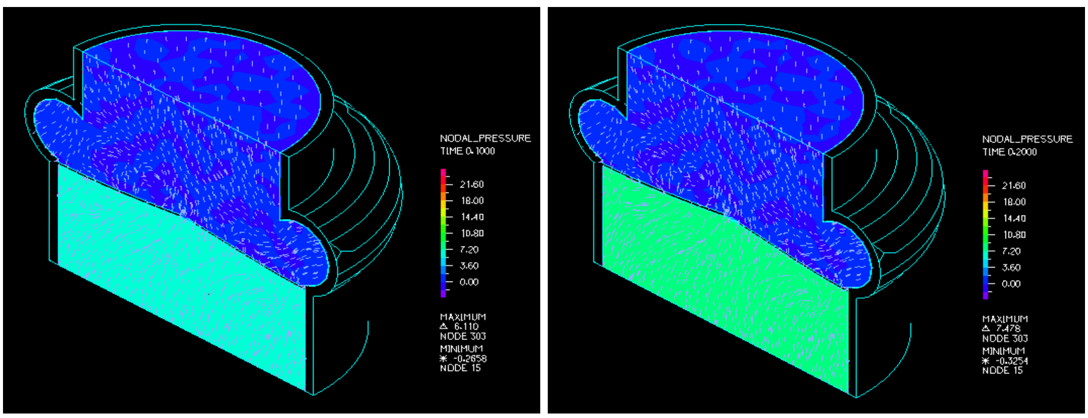


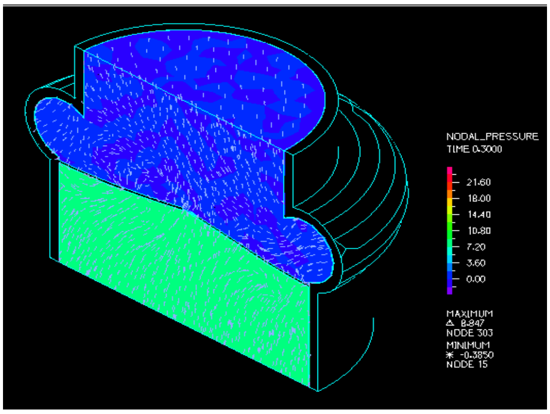
Figure 4.8: Graph Comparison velocity with different size

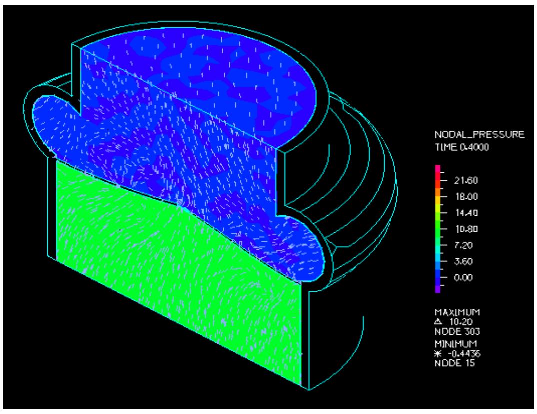
According to the graph, all of the models show the positive increased of graph trend. At time 0.5 seconds, normal design of the sinus is 6.13 percent higher than large model design of sinus but slower as much as 0.0275 m/s at the same level of working pressure. From the results, the model 1 with the small design of sinus recorded the highest velocity compare to the normal size and large design. The maximum value comparison showed the large design of the sinus was recorded lower velocity than small design as much 0.0365 m/s.

4.3.4 Nodal Pressure Analysis (Normal Size of Sinus)

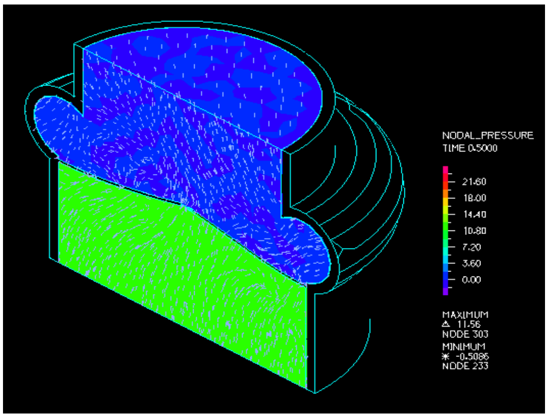


(a) (b)


NODAL_PRESSURE
TIME 0.3000
21.60
18.00
14.40
10.80
7.20
3.60
0.00
MAXIMUM
△ 8.847
NODE 303
MINIMUM
* -0.3850
NODE 15


NODAL_PRESSURE
TIME 0.4000
21.60
18.00
14.40
10.80
7.20
3.60
0.00
MAXIMUM
△ 10.20
NODE 303
MINIMUM
* -0.4436
NODE 15

(c) (d)


NODAL_PRESSURE
TIME 0.5000
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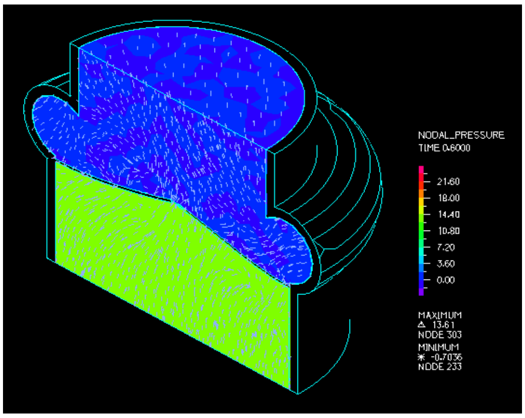
18.00
14.40
10.80
7.20
3.60
0.00
MAXIMUM
△ 11.56
NODE 303
MINIMUM
* -0.5086
NODE 233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14.40
10.80
7.20
3.60
0.00
MAXIMUM
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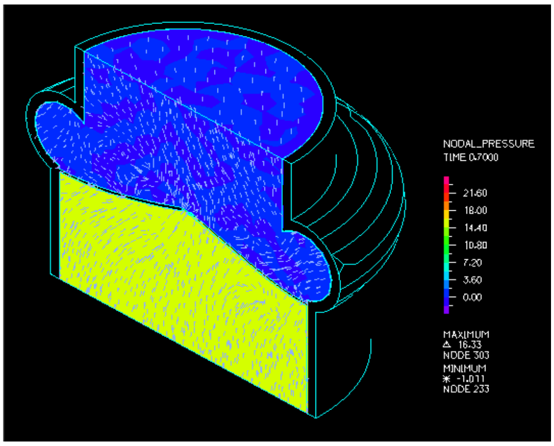
NODE 303
MINIMUM
* -0.7036
NODE 233

(e) (f)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14.40
10.80
7.20
3.60
0.00
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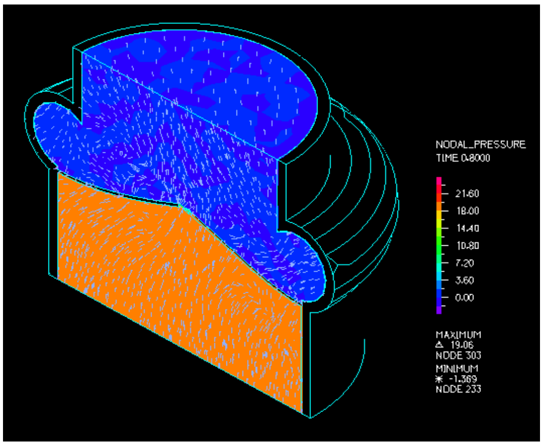
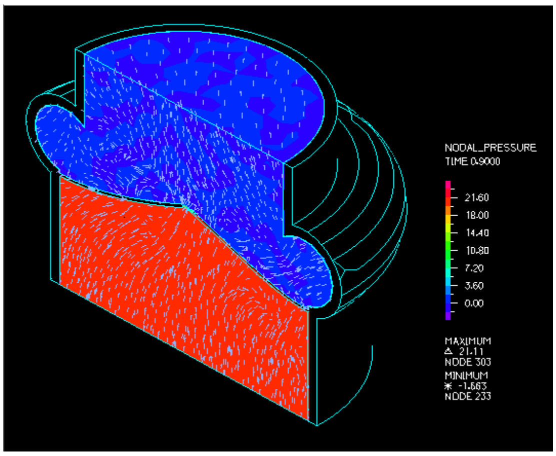
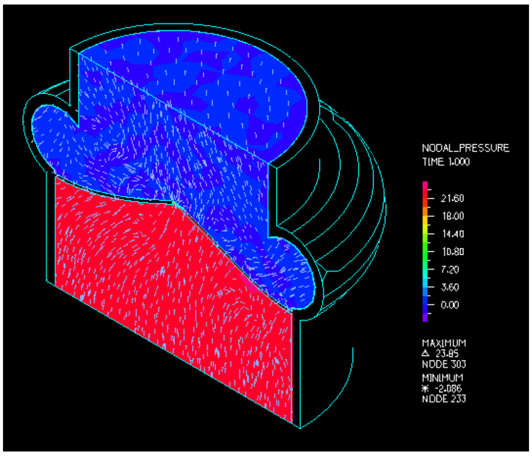
MINIMUM
* -1.011
NODE 233


NODAL_PRESSURE
TIME 0.8000
21.60
18.00
14.40
10.80
7.20
3.60
0.00
MAXIMUM
△ 19.06
NODE 303
MINIMUM
* -1.369
NODE 233

(g) (h)



(i)



(j)

Figure 4.9: Nodal Pressure analysis for normal size of AV sinus at time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j 1.00 Sec

Illustrations showed the vector of blood flow and magnitude changes which represented by a color. On time.0.1 second, the nodal pressure magnitude is equal 6.0493 kN/m^{2,} which increased 11 percent from the initial state. The vectors performed the direction of flow in the system during the systole process. The initial state of the model showed the vector flow via the opening leaflet valve when the loads applied in the system. Loads 40 mmHg to 175 mmHg were gradually applied to this model. The results showed clear illustrations of fluid dynamics and interaction of the structure within time 1 second. The nodal pressure recorded 11.4327 kN/m^{2} after 0.5 second with the value of pressure acting is 85 mm Hg. The nodal pressure depends on how much the pressure applied to the heart system and the ability of leaflets to allow the blood through it. The maximum states at load pressure 175 mm Hg applied to the fluid and structure model of this analysis are the vector number become more compare than before condition. The value of an increased percentage of nodal pressure value from initial state to the peak condition is 77.09 percent.

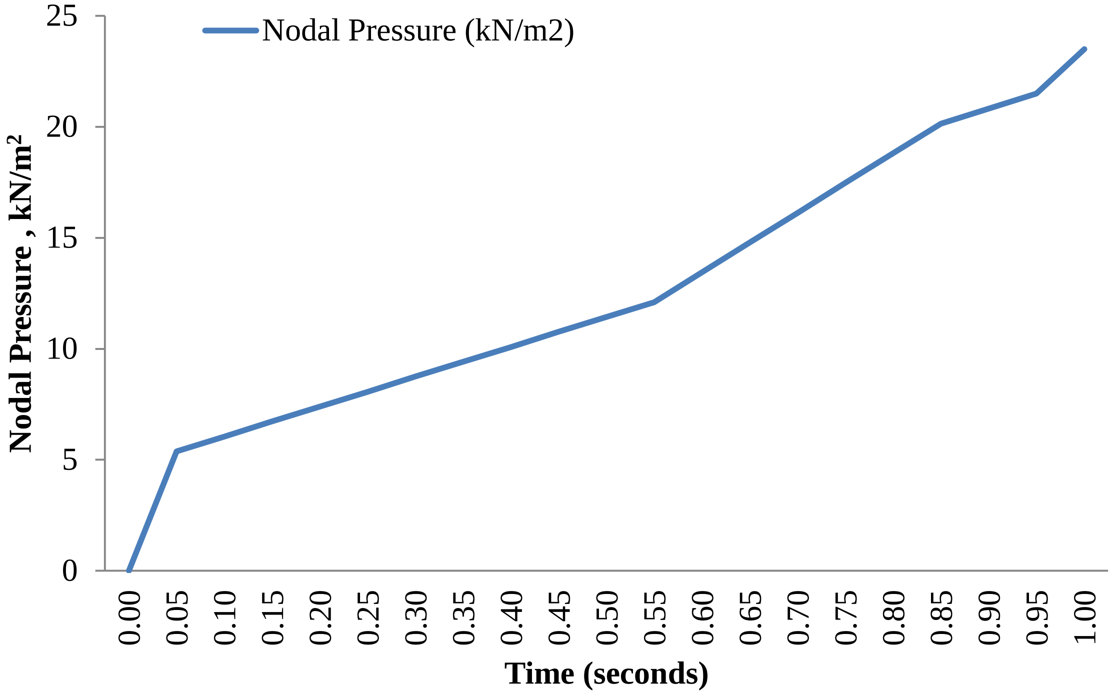
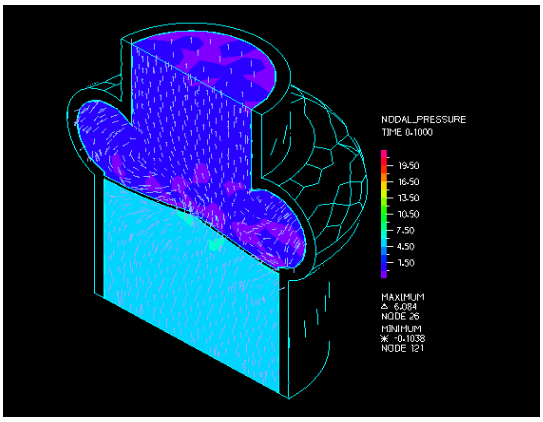


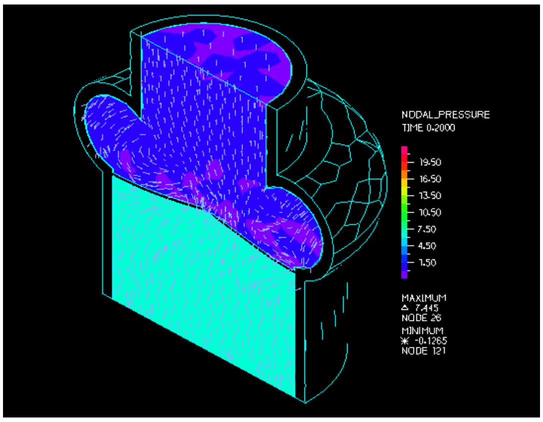
Figure 4.10: Graph Nodal Pressure versus time for Aortic Valve

The graph is a linear correlation within time 1 second. The initial state of the model showed the vector flow via the opening leaflets (at time 0.05 seconds) valve when the loads applied in the system. Loads pressure was applied as much as 40 mmHg to 175 mmHg to the aortic valve model. The initial nodal pressure obtained, when the pressure give into the aortic valve is equal to 5.3833 kN/m^2.

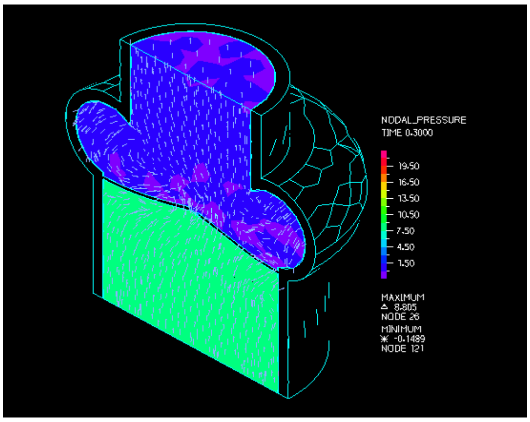
From graph analysis, the nodal pressure value is increased when pressure acting inside the aortic valve also added within time 1 second during the systole phase. At time 0.5 second, the nodal pressure increase as much as 6.0494 kN/m^2 started from the opening phase of leaflets. While the maximum of nodal pressure for aortic valve recorded in 1 second is 23.5026 kN/m^2 with working pressure 175 mmHg.The detailed of nodal pressure value within time 1 second can be referred in Table 4.5 in appendix D1.

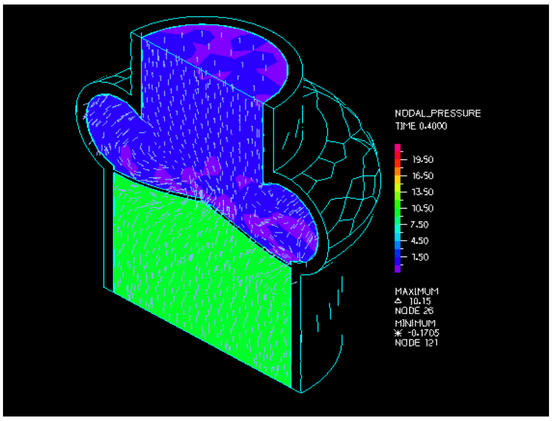
4.3.5 Nodal Pressure Analysis (Small Size of Sinus)



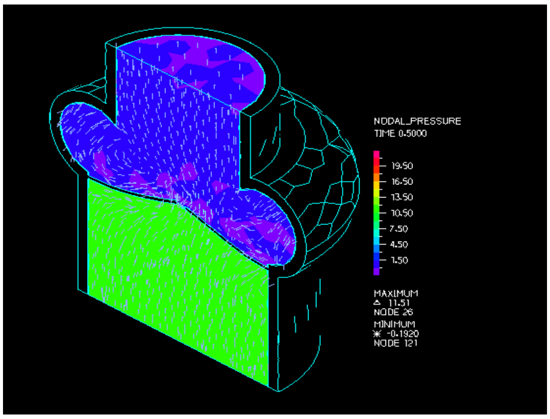


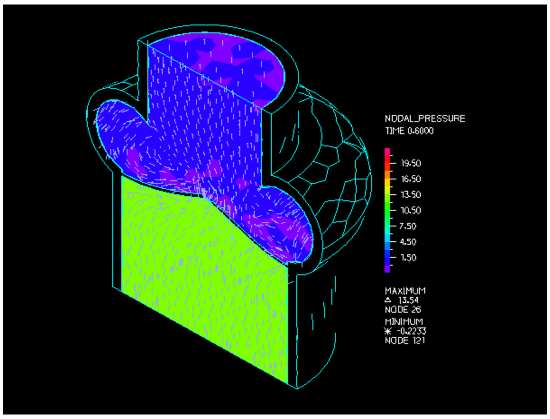
(a) (b)



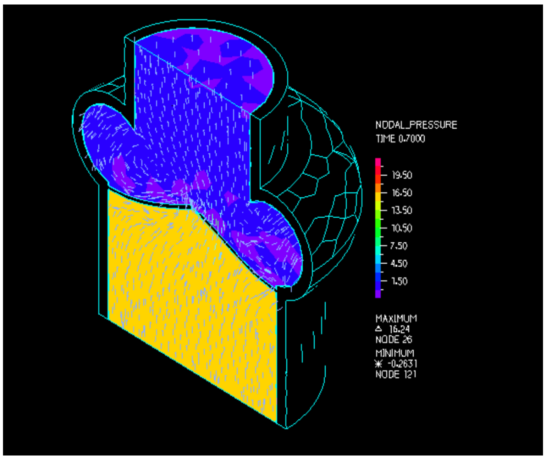


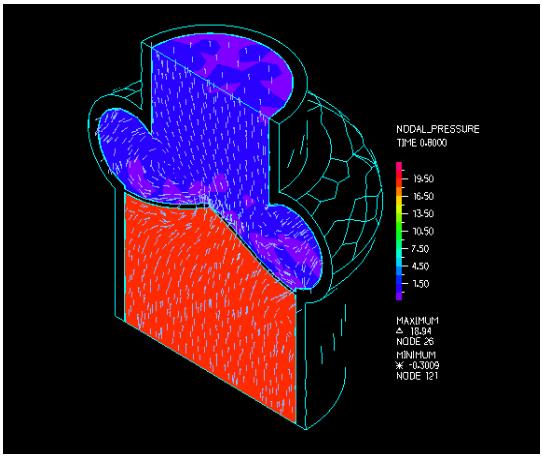
(c) (d)





(e) (f)





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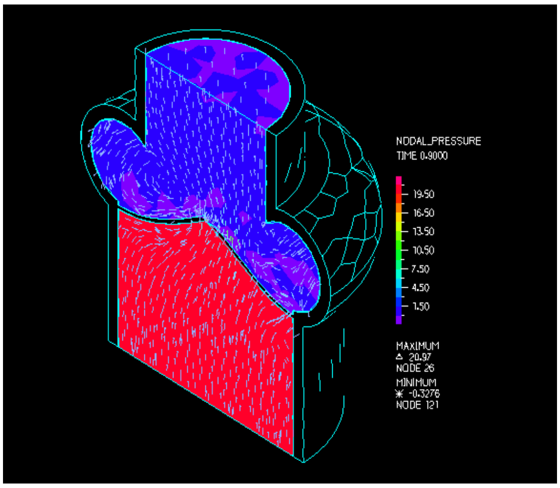
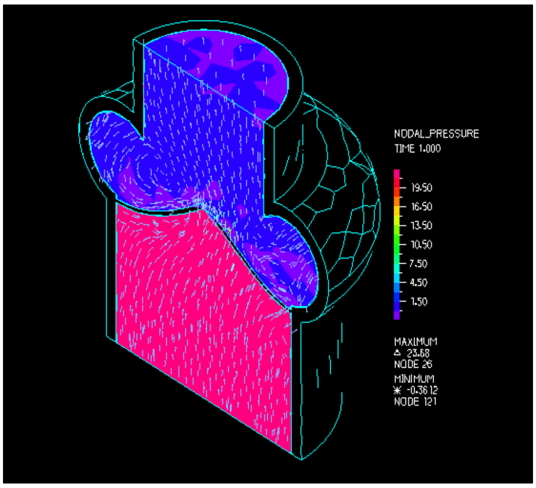


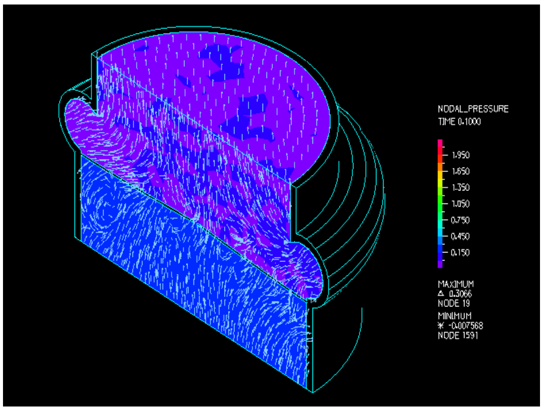


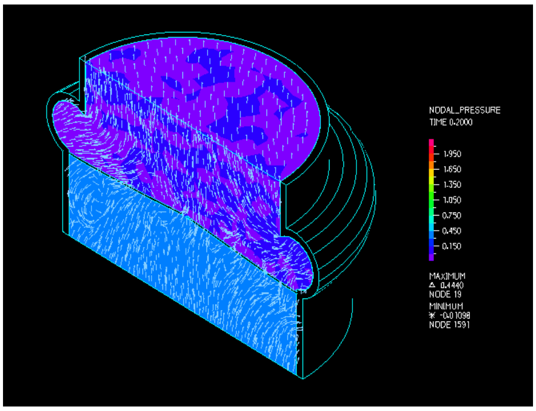
(i) (j)

Figure 4.11: Nodal Pressure analysis for small size of AV sinus at time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j) 1.00 Sec

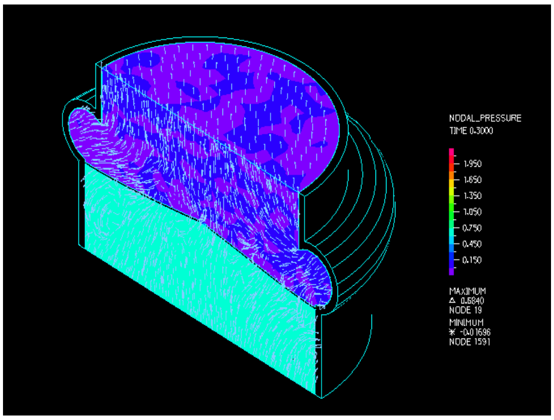
The illustration figures show the nodal pressure for small design of sinus. The leaflet is deformed linearly with the pressure acting. In this comparison, the working pressure state is below than 40 mmHg. The differences colors show the difference value of pressure within in the time range 1 second. The movement of vectors can be seen clearly in the aortic valve region. The maximum velocity achieved at a time setting 1 second.

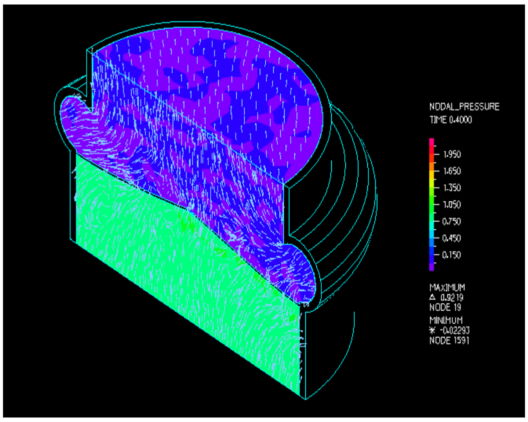
4.3.6 Nodal Pressure Analysis (Large Size of Sinus)



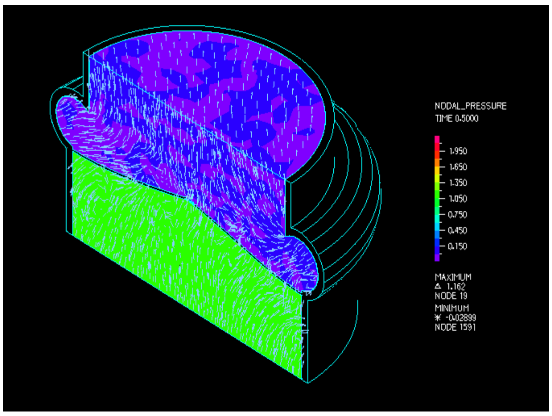


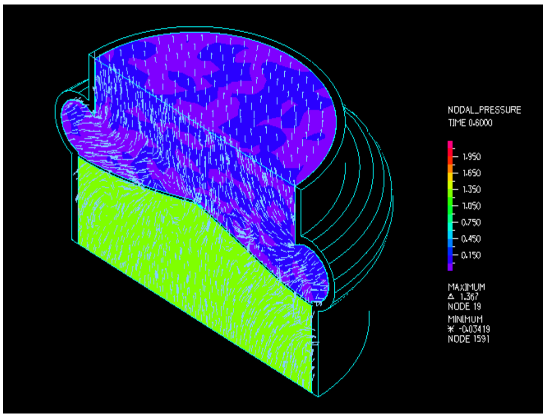
(a) (b)



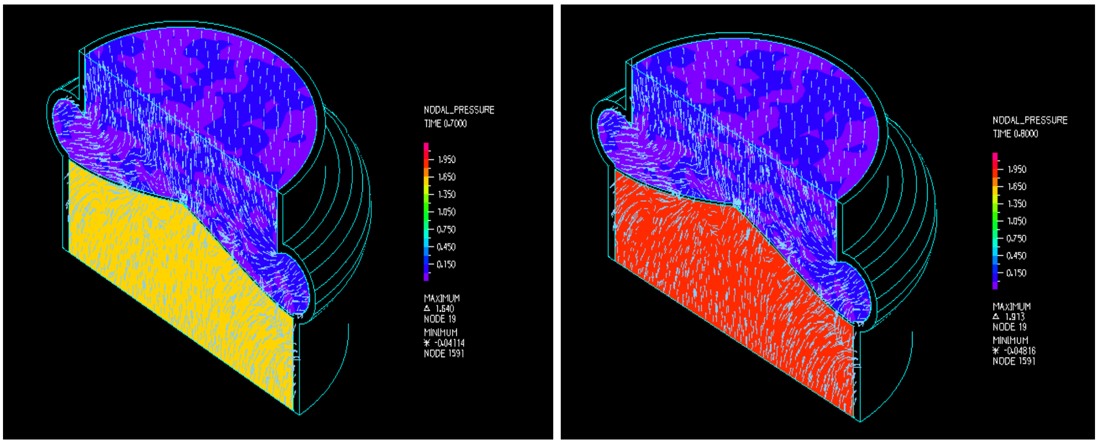


(c) (d)

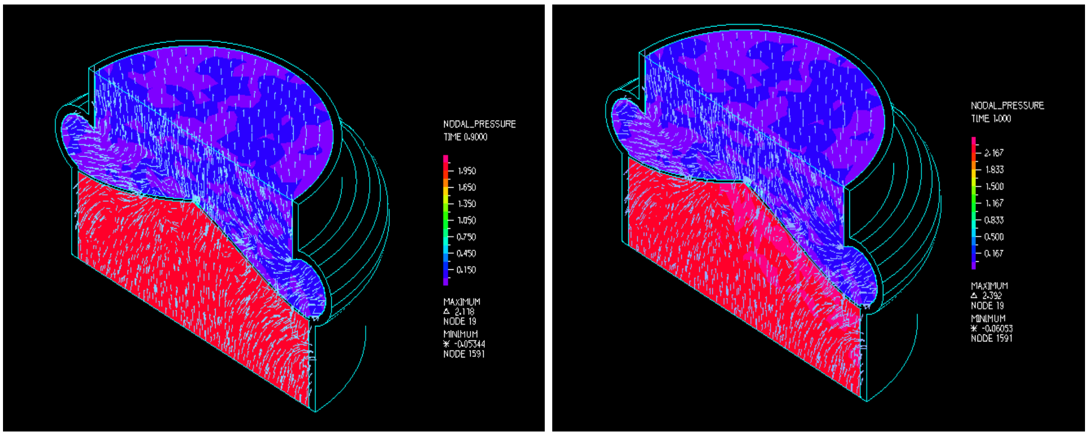




(e) (f)



(g) (h)

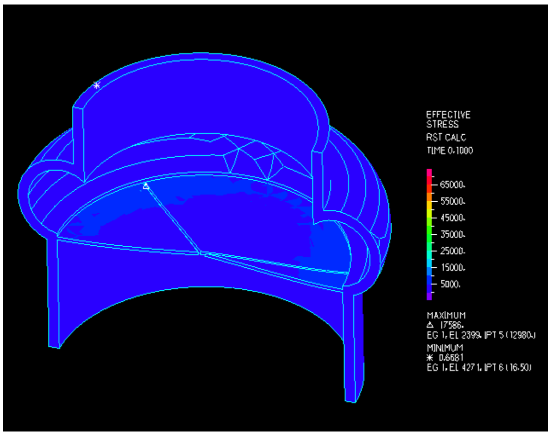


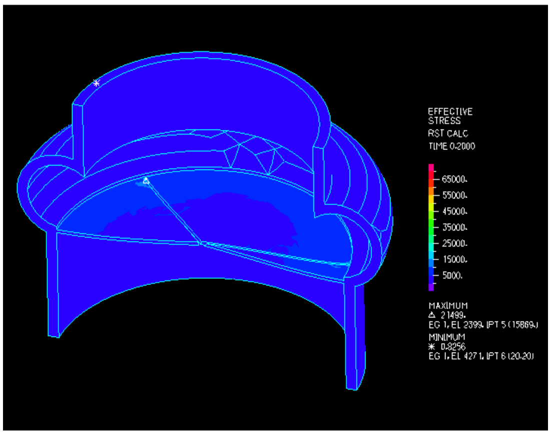
(i) (j)

Figure 4.12: Nodal Pressure analysis for large size of AV sinus at time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j) 1.00 Sec

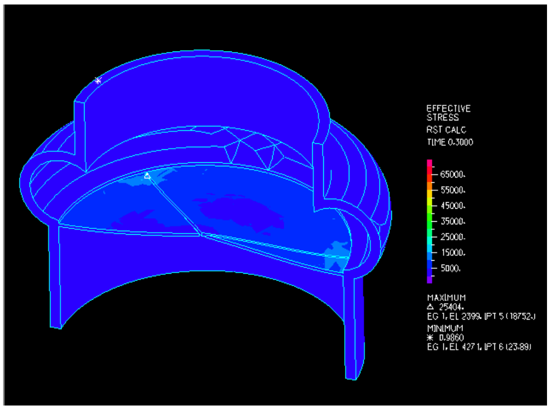
According to the sequences of illustration, the dynamic motion of leaflets is proportion to the working pressure. The vectors showed the movement of flow pattern in the system. At the initial stages, the leaflets are starting to deform due to the pressure action. Then, the leaflets deformation has been increased linearly as much as flow pushed the leaflets move upwards. From the results, the nodal pressure for large design is higher compared to the reference and small design of sinus model. It causes by the area of sinuous design. More larger design of sinus gives the less nodal pressure in the system.

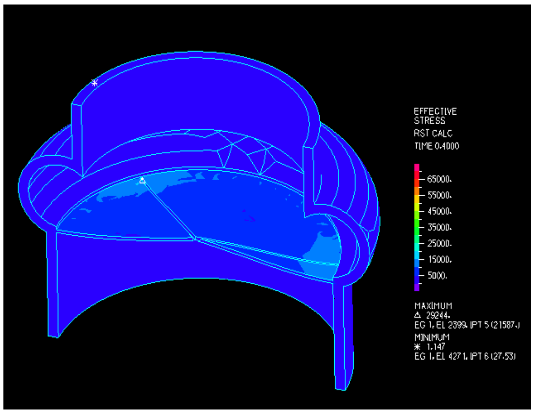
4.3.7 Effective Stress Analysis (Normal Size of Sinus)



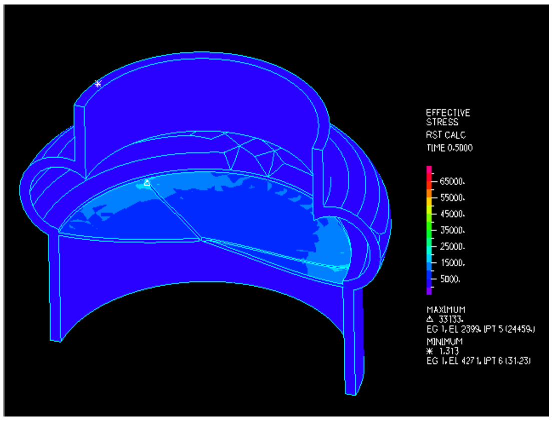


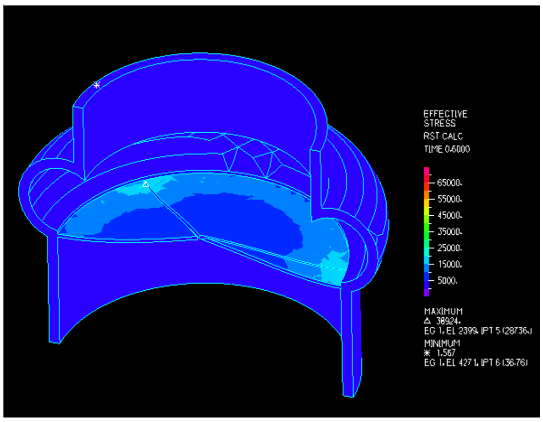
(a) (b)



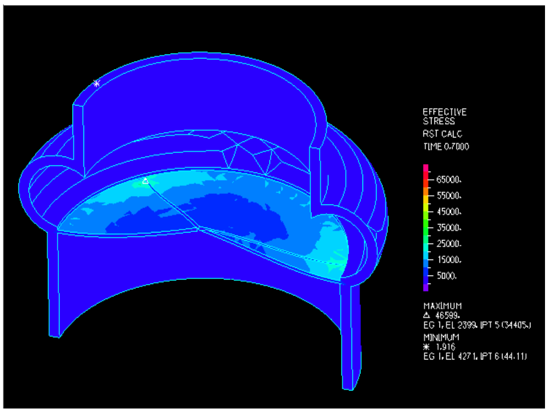


(c) (d)

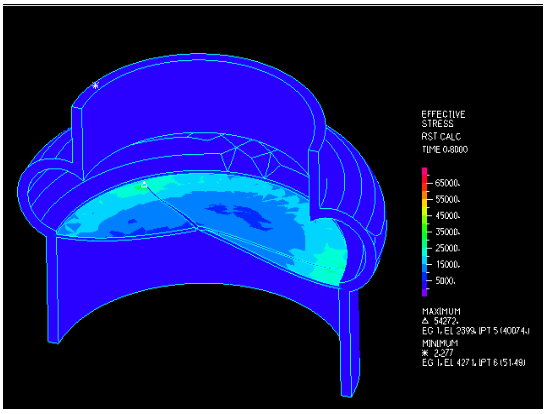




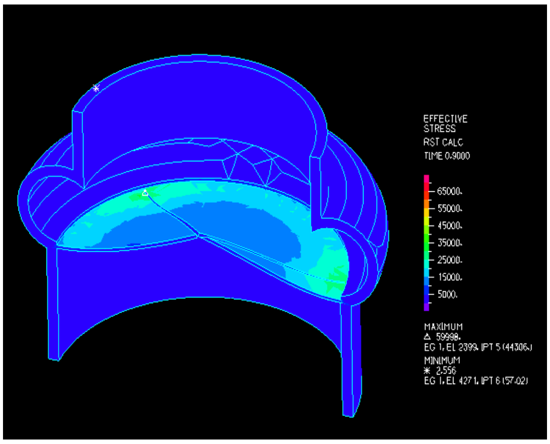
(e) (f)



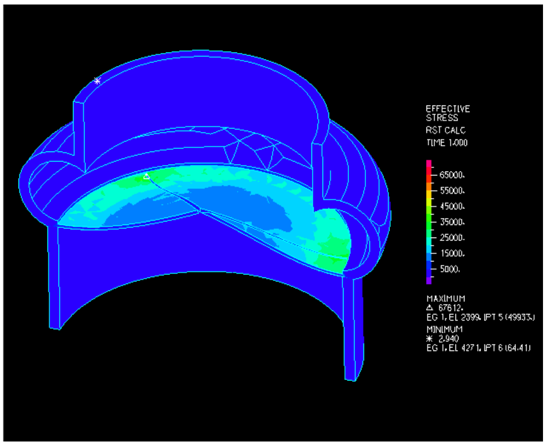
(g)



(h)



(i)



(j)

Figure 4.13: Effective stress analysis for normal size of AV sinus at time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j) 1.00 Sec

During the initial opening leaflets phase, the effective stress in the leaflets is very small. The stress relation to the opening is a linear correlation due to the increased pressure applied in the system. The most striking aspect of this simulation was the dynamic motion of leaflets deformed was animated in time varying range 1seconds.The model of mitral valve simulation results illustrates the clear view of critical area on the leaflets. The effective stress analysis is more focus to the leaflets structure of the aortic valve. From the sequences of figures, the leaflets stress value depends on how much the pressure acts on it. When the pressure leaflets receive high pressure, so the impact stress of the pressure also high. During the pressure act, the leaflet of aortic valve goes upward due to the pressure direction. The initial value of effective stress recorded is equal to 15656 N/m^2.

Next, the changes of effective stress can see clearly by representing colors. In this result analysis, the major part of an area that had been seeing is occurring at the edge of top and bottom leaflets which nearer to the wall of the aortic valve. The increasing of effective stress as much as 52.74 percent due to the increasing pressure as much as 52.65 percent between time 0 to 0.5 second. From this result, this 3D simulation of the aortic valve can give clear views about the relation between the pressure and effective stress of the structure.

Based on the theoretical validation, when pressure increase, the stress also increased. The result showed the value of effective stress is linearly increased. The maximum effective stress achieved in time 1 second with the working pressure 23.99 kPa is 67612 N/m^2.

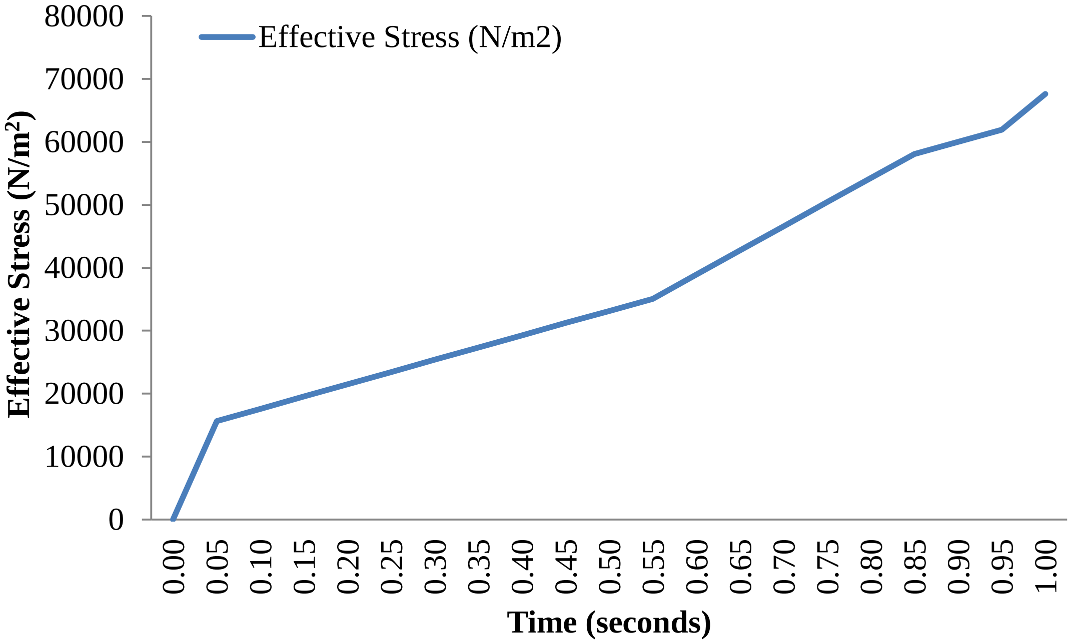
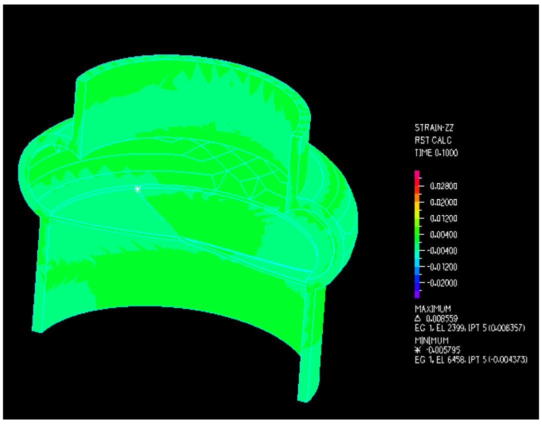


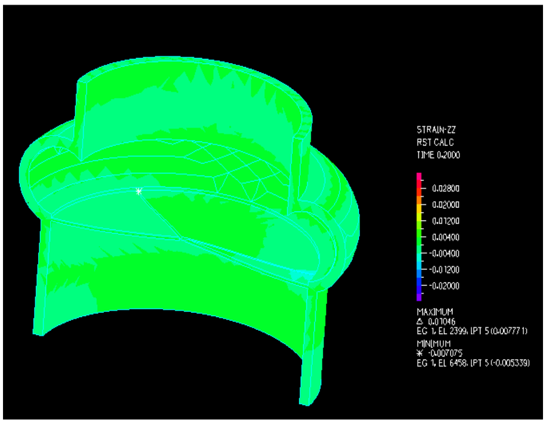
Figure 4.14: Graph Effective Stress versus time for Aortic Valve

The graph is a linear correlation within time 1 second. The lowest effective stress obtained when the pressure giving into the aortic valve is equal to 15656 N/m^2 .From graph analysis, the effective stress value is increased when pressure acting on the aortic valve leaflets increased during the systole phase. The effective stress increased as much as 15562 N/m^2 from an initial condition with time range 0 to 0.5 second .The maximum effective stress recorded in 1 second is equal 67612 N/m^2 with percentages increased as much as 76.84 percent when the working pressure is 23.99 kPa. The detailed of velocity value within time 1 second can be referred in Table 4.2.

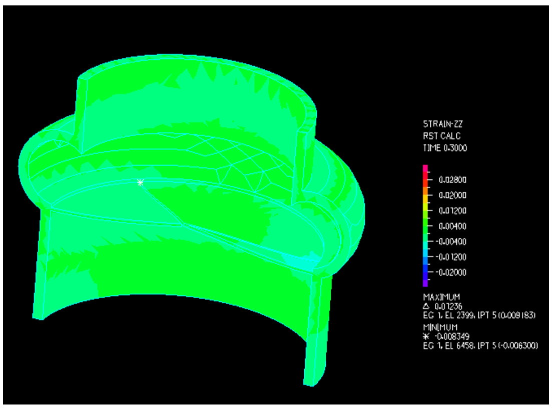
4.3.8 Strain Analysis (Normal Size of Sinus)



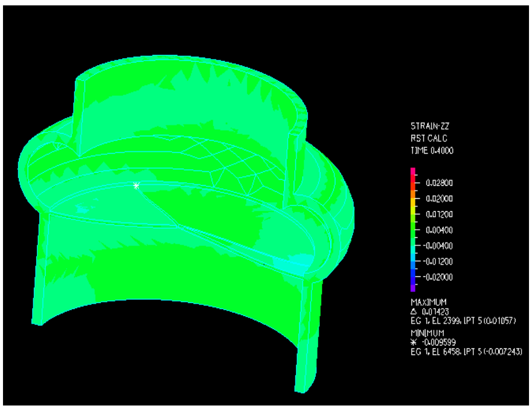
(a)



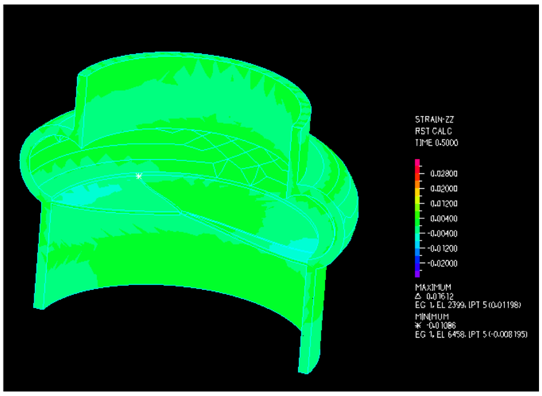
(b)

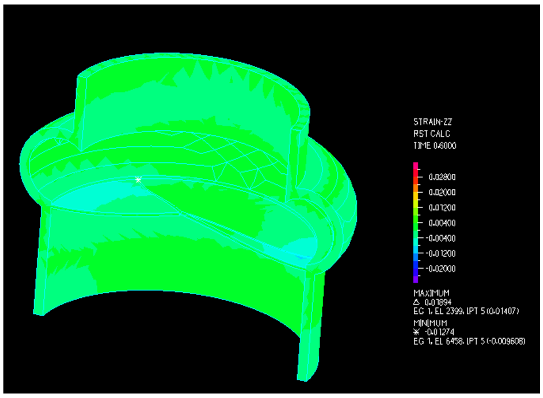


(c)

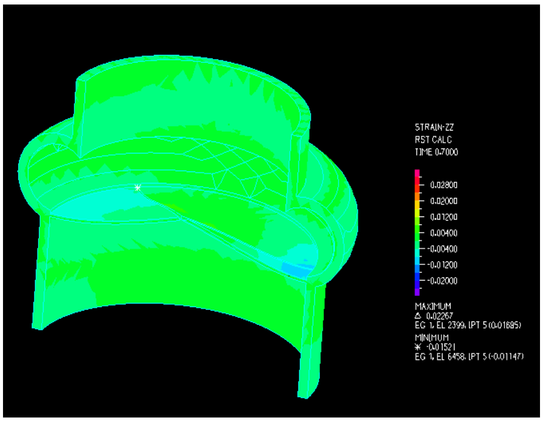


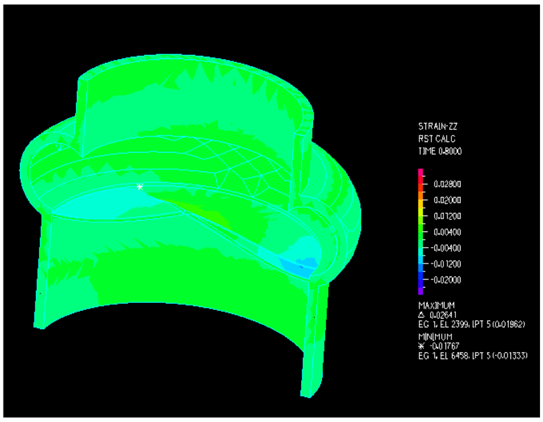
(d)



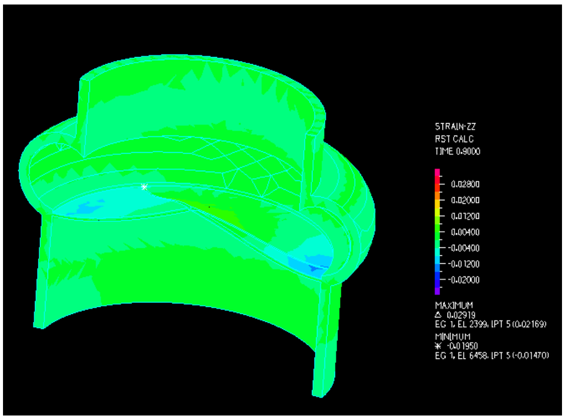


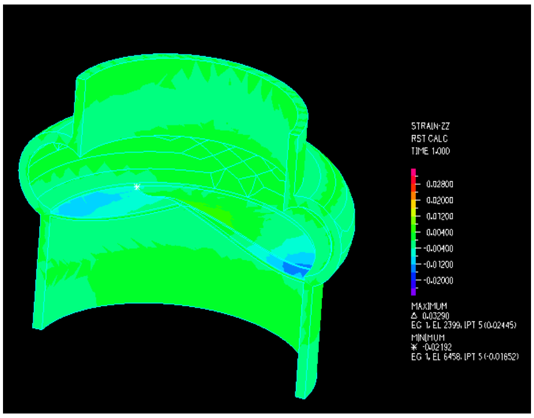
(e) (f)





(g) (h)





(i) (j)

Figure 4.15: Strain analysis for normal size of AV sinus at time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j) 1.00 Sec

One of the main advantages in conducting numerical simulations is that they yield information on some characteristics of the valve such as developed strain at the leaflets structure. The deformation of leaflets brings the effect to the velocity of blood flow, stress on the leaflets structure, and strain analysis. From the illustrations figures, the aortic valve leaflets kinematics is a linear correlation with the load applied in the heart system. During time 0.1 seconds, the strain value recorded is equal to 0.008559. The strain increased 10.98 percent from the beginning phase. The strain value percentage of strain analysis increased as much as 523.74 percent with pressure equal to 113.32x10^3 dyns/cm^2 (11.32 kPa) within time 0.05 second to 0.5 second .The results presented the critical area mostly happen at the edge of leaflets near the aortic valve wall structure. However, the maximum pressure 240x10^3 dyns/cm^2 applied through the aortic valve is still in safe condition for human hearts, e.g. the pressure given not much effect to the abnormalities of the heart.

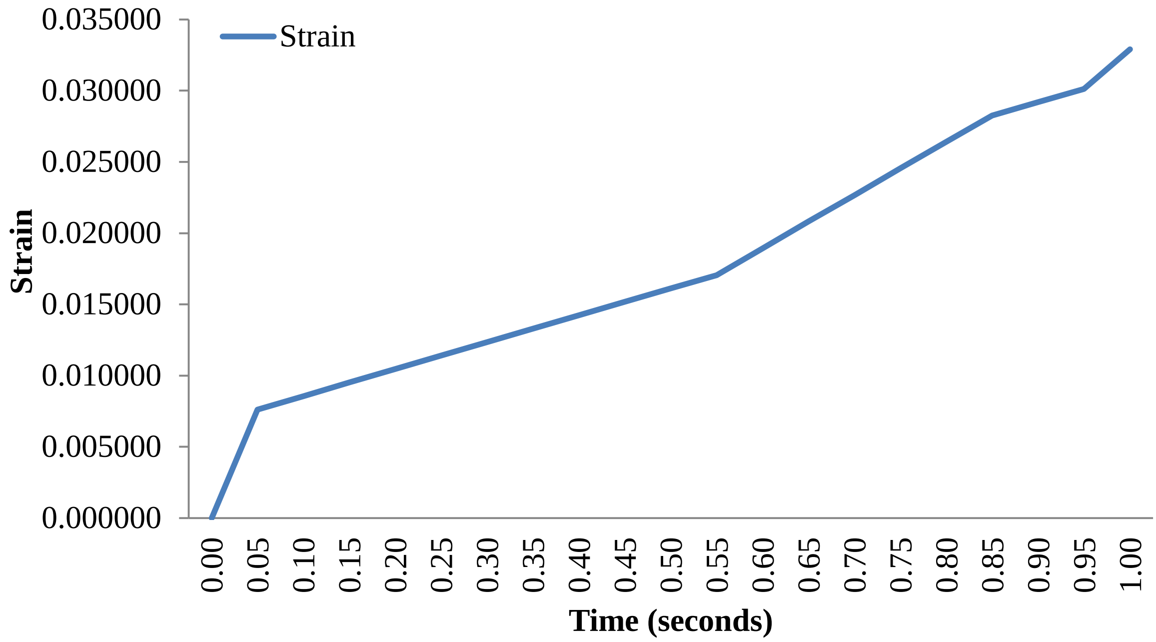


Figure 4.16: Graph Strain versus time for Aortic Valve

The graph showed the linear positive increasing within time 1 second. During initial pressure acting to the aortic valve leaflets structure, the strain results obtained from the simulation is 0.007619.From graph analysis, the strain value is increased when pressure acting on the aortic valve leaflets increased during the systole phase. The percentage strain increased as much as 0.008501 from an initial condition between time range 0.05 to 0.5 second .The maximum strain recorded in 1 seconds is equal 0.032900 with percentages increased as much as 76.84 percent when the working pressure is 23.99 kPa. The detailed of velocity value within time 1 second can be referred in Table 4.3 in attachment appendix D1.

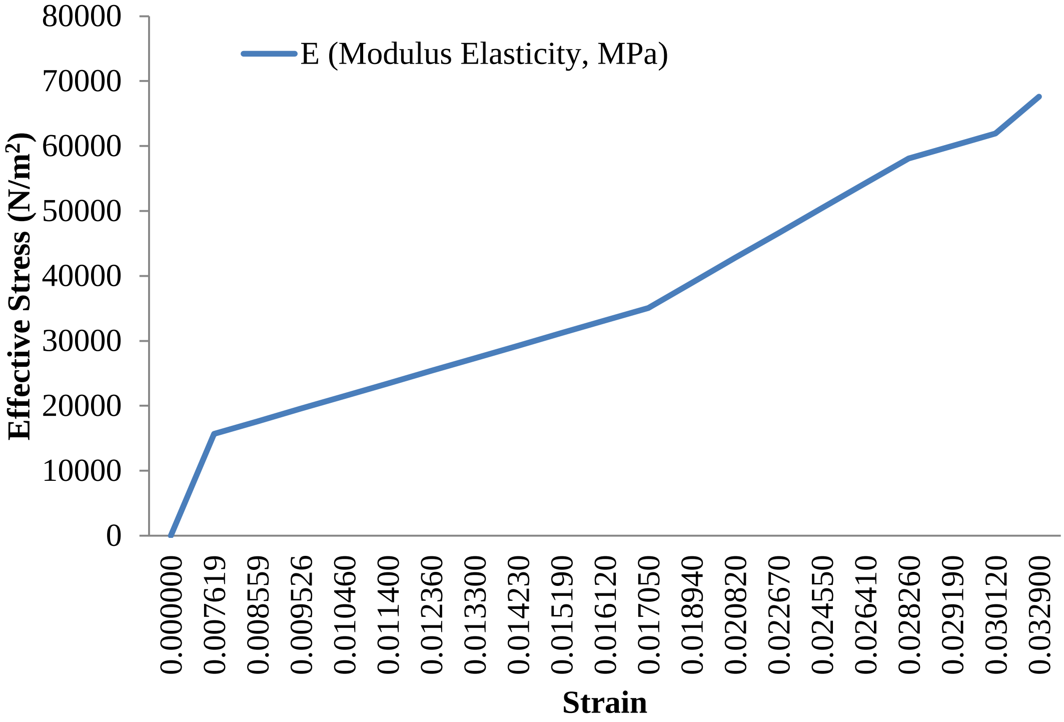
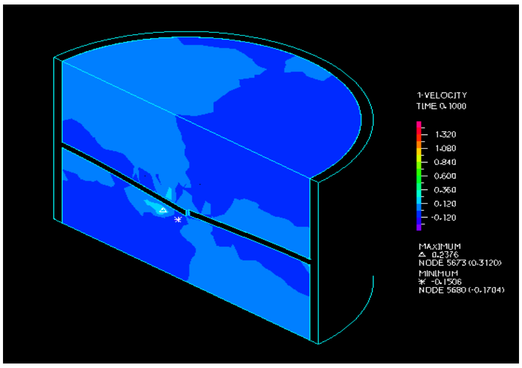


Figure 4.17: Graph Effective stress versus strain

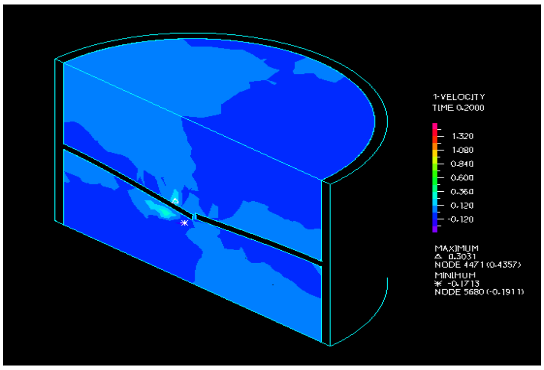
Graph stress-strain of the leaflets results show the linear relationship effective stress and strain. From the simple calculation from equation, $E = \frac{\sigma}{\varepsilon}$, the value of modulus elasticity for the maximum effective stress and maximum strain is 2.05 MPa. There is a 2.43 % difference compare to the theoretical value from Espino, D.M. . From this result, the findings value of young modulus elasticity can be referenced for medical practitioners to select suit material for replacement heart valve.

4.4 RESULTS FOR MITRAL VALVE

4.4.1 Velocity Analysis



(a)



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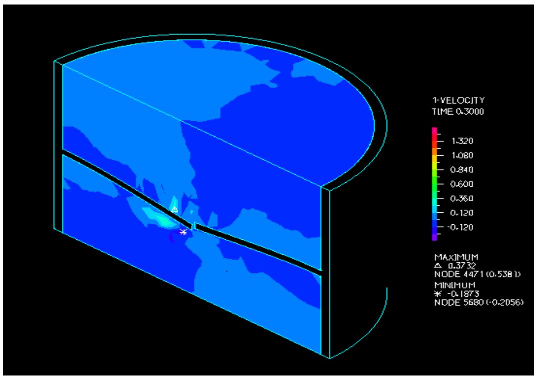
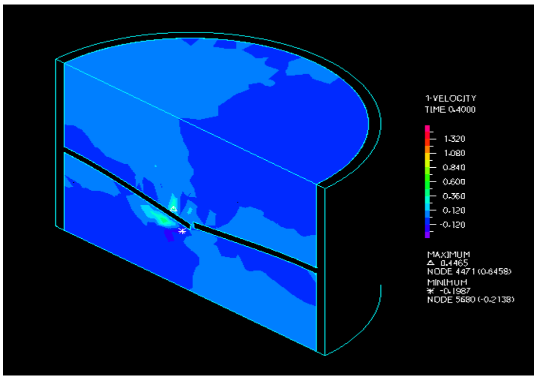
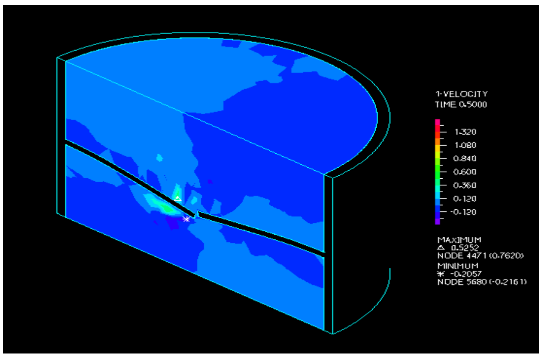


(c)



(d)



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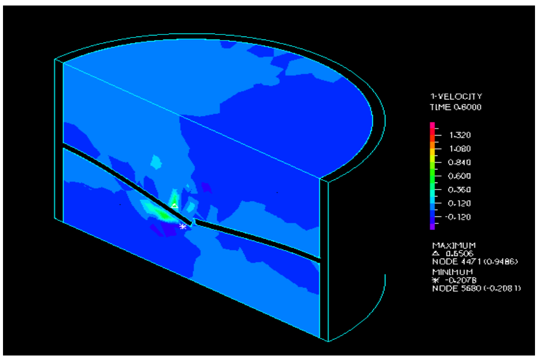

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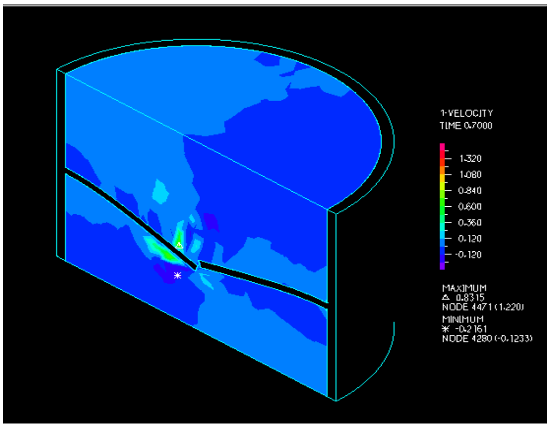
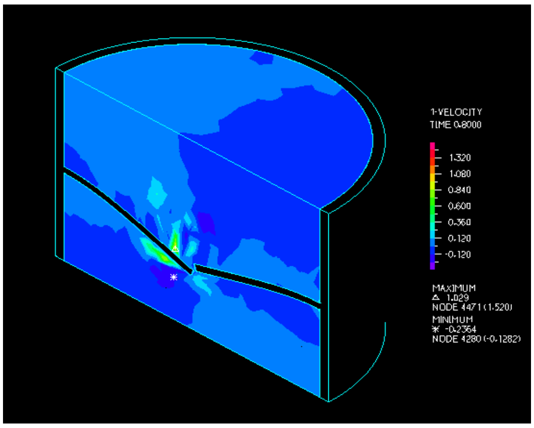
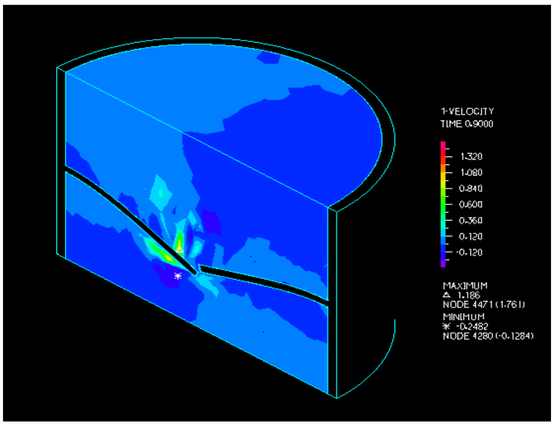


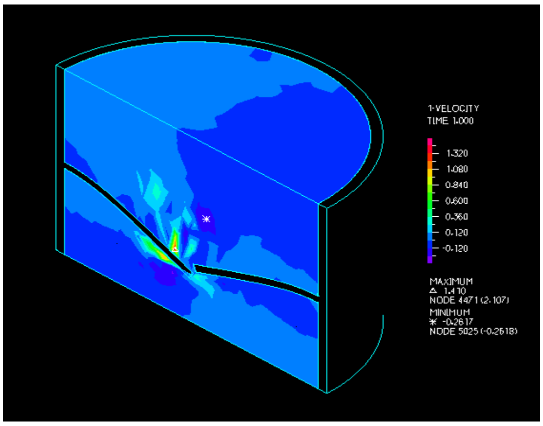
(g)



(h)



(i)



(j)

Figure 4.18: Velocity analysis of MV model of time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j) 1.00 Sec

The velocity of the blood flow pattern inside the mitral valve is influenced by the pressure applied in the heart system. From the simulation, the leaflets dynamics go downward due to the pressure direction. At the initial condition of velocity flow pattern in the mitral valve, no changes or movement happens to the leaflets. It happens cause no pressure supplied to the system. But when the pressure entered the valve region at time 1 second with pressure value 40mmHg, the velocity recorded 0.2376 m/s. . .Regarding on the sequence of figures, the deformation of leaflets is linear proportion to the pressure given into the mitral valve. The different colors represented different value of velocity. The increasing of pressure as much as 47.1 percent of the initial pressure given, the velocity obtained 0.5252 m/s at time 0.5seconds. In this condition, the opening of the leaflets is bigger than before. The maximum velocity obtained is 1.410 m/s at time 1 second, with increasing of pressure value starting from initial condition as much as 18.66 kPa.

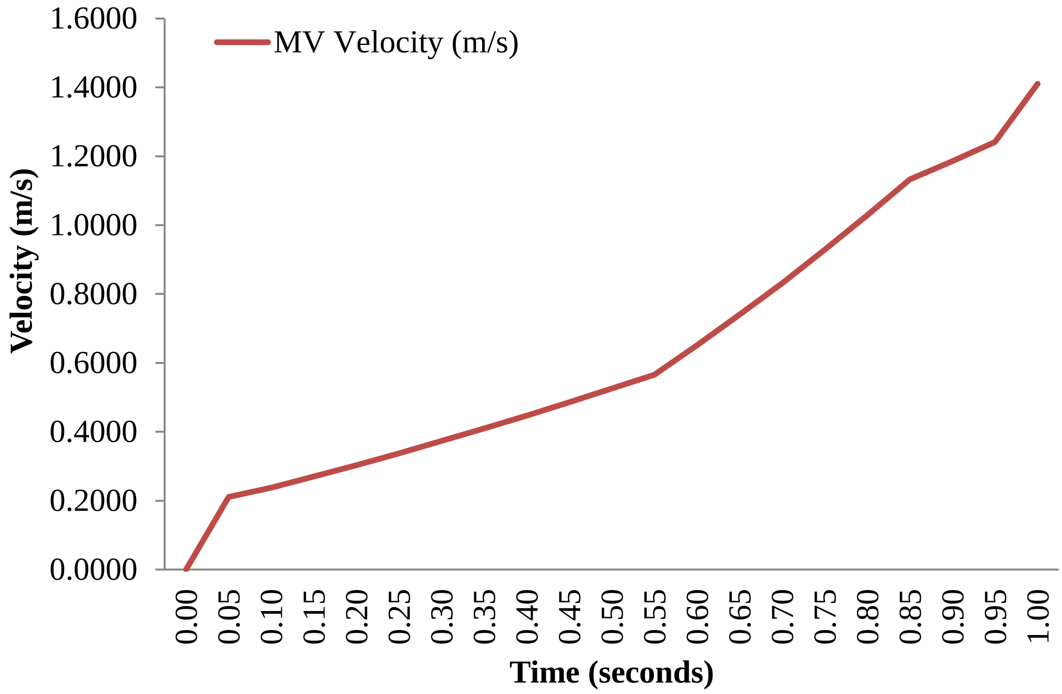
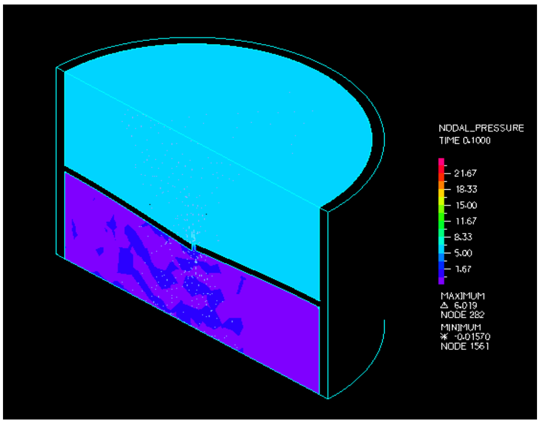
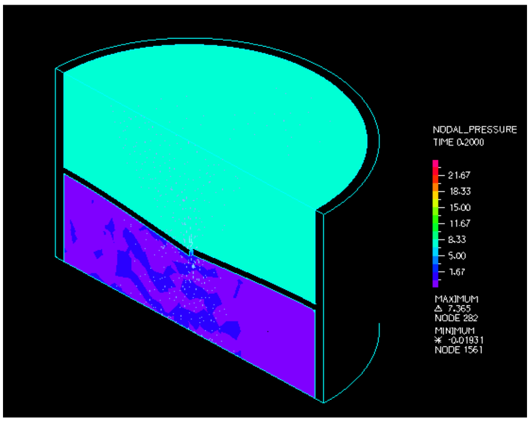


Figure 4.19: Graph Maximum velocity versus time for mitral valve

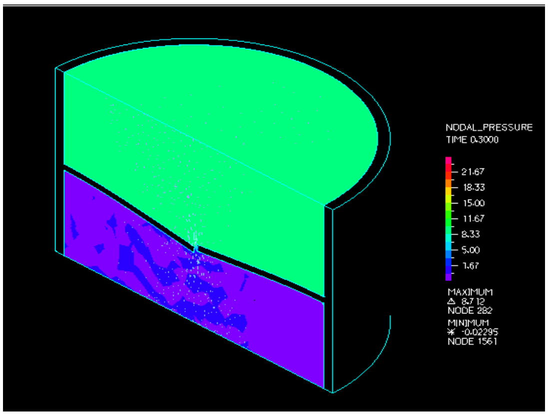
The graph is a linear correlation within time 1 second. The lowest velocity obtained, when the pressure give into the mitral valve is equal to 0.6881 m/s. From graph analysis, the velocity value is increased when pressure acting inside the mitral valve also increased during the systole phase. The velocity increased as much as 59.90 percent within time range 0 to 0.50 second. The maximum velocity is equal 1.410 m/s when the working pressure is 23.99 kPa. The detailed of velocity value within time 1 second can be referred in Table 4.6 such in attachment appendix D1.

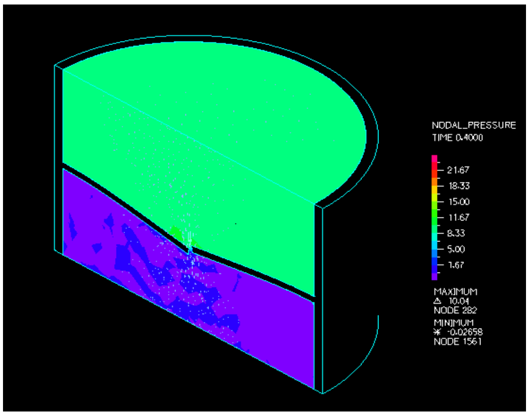
4.4.2 Nodal Pressure Analysis



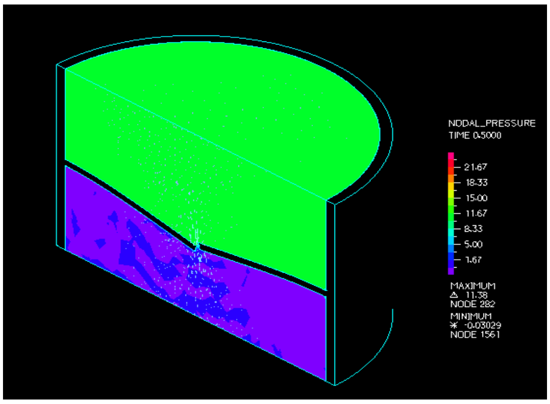


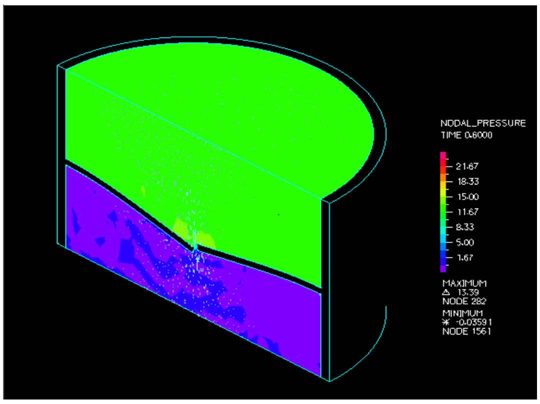
(a) (b)



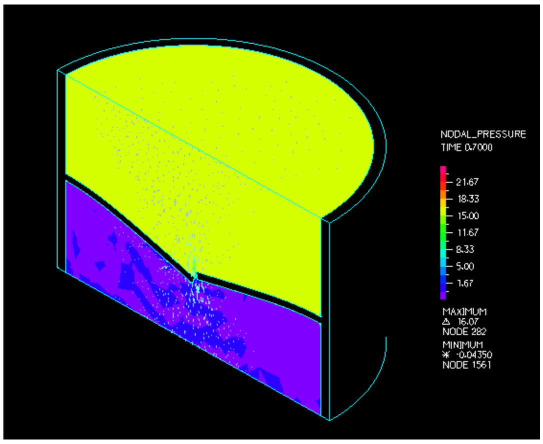


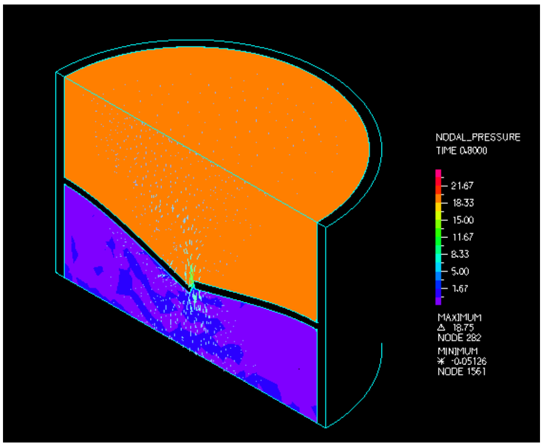
(c) (d)



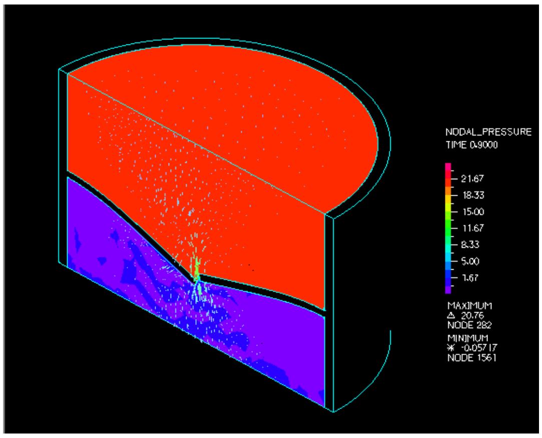


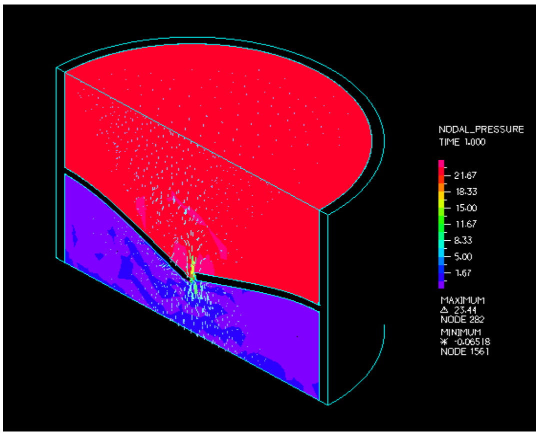
(e) (f)





(g) (h)





(i) (j)

Figure 4.20: Nodal pressure analysis of MV model of time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j) 1.00 Sec

Illustrations sequences figures showed the vectors and the magnitude of nodal pressure which are represented by a different color. In the initial state, no vector move flow from the inlet through the mitral valve leaflets until the applied pressure acting on the system. During time at 0.1 second, the nodal pressure magnitude is equal to 5.9969 kN/m^2 which increased 11.03 % from the initial state. The vectors performed the direction of flow in the system during the systole process. The initial state of the model showed the vector flow via the opening leaflets at a time 0.05 seconds valve when the loads applied in the system. Loads pressure was applied as much as 40 mmHg to 175 mmHg to the mitral valve model. The results showed clear illustrations of fluid dynamics and interaction of the structure within time 1 second. The nodal pressure recorded 11.3580 kN/m^2 after 0.5 second with the value of pressure acting is 85mmHg. The nodal pressure depends on how much the pressure applied to the heart system and the ability of leaflets to allow the blood through it. The maximum states at load pressure 175 mm Hg applied to the fluid and structure model of this analysis create the number of vector flow through the leaflets become more compare than before condition. The value of an increased percentage of nodal pressure value from initial state to the peak condition is 77.24 percent.

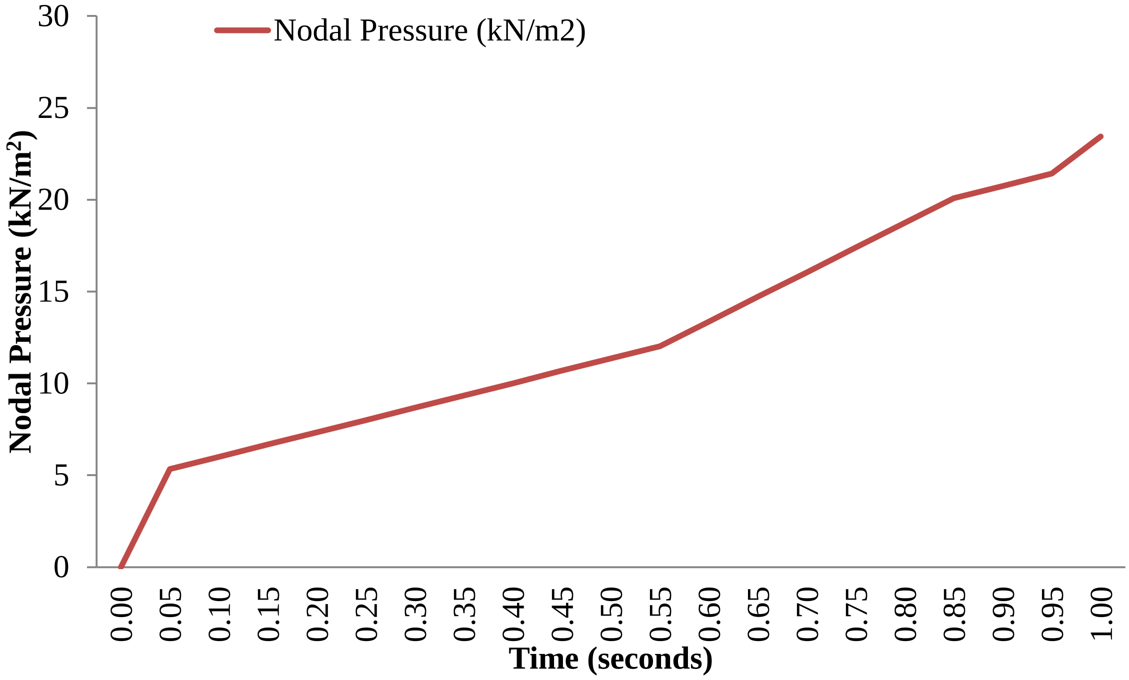


Figure 4.21: Graph Nodal Pressure versus time for Mitral Valve

The graph is a linear correlation within time 1 second. The initial state of the model showed the vector flow via the opening leaflets (at time 0.05 seconds) valve when the loads applied in the system. Loads pressure was applied as much as 40 mmHg to 175 mmHg to the mitral valve model. The initial nodal pressure obtained, when the pressure give into the mitral valve is equal to 5.3352 kN/m^2. From graph analysis, the nodal pressure value is increased when pressure acting inside the mitral valve also added within time 1 second during the systole phase. At time 0.5 second, the nodal pressure percent increase as much as 53.02 % started from the opening phase of leaflets. While the maximum of nodal pressure for mitral valve recorded in 1 second is 23.4462 kN/m^2 with working pressure 175 mmHg.The detailed of nodal pressure value within time 1 second can be referred in Table 4.10.

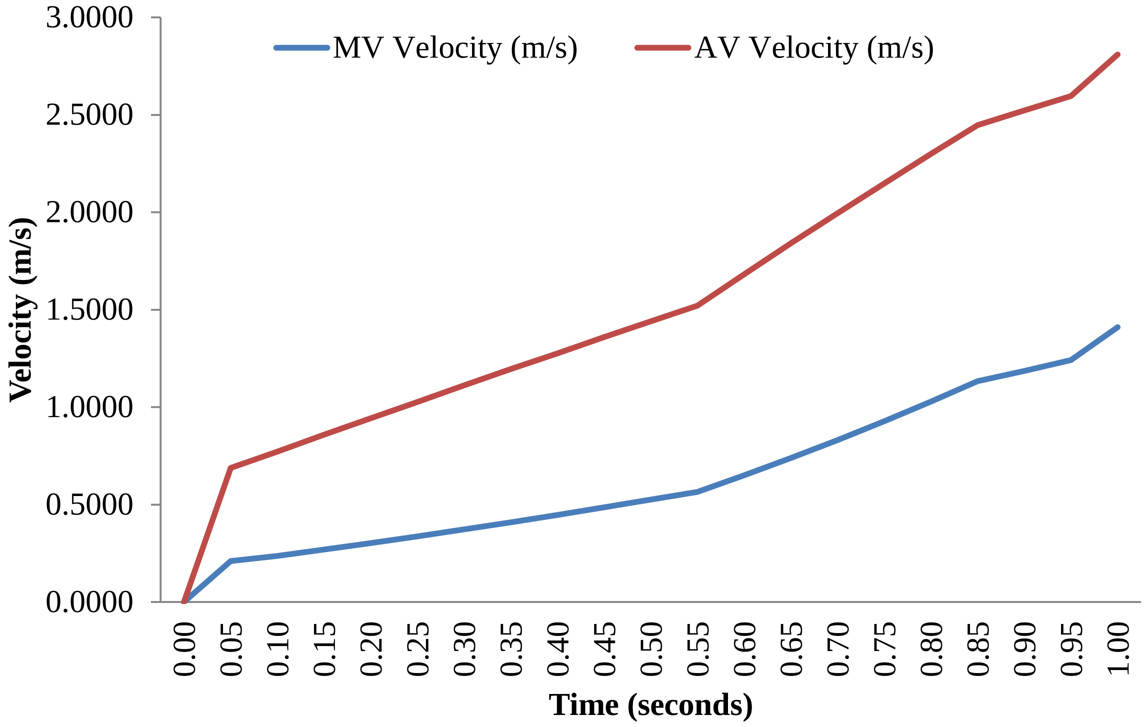
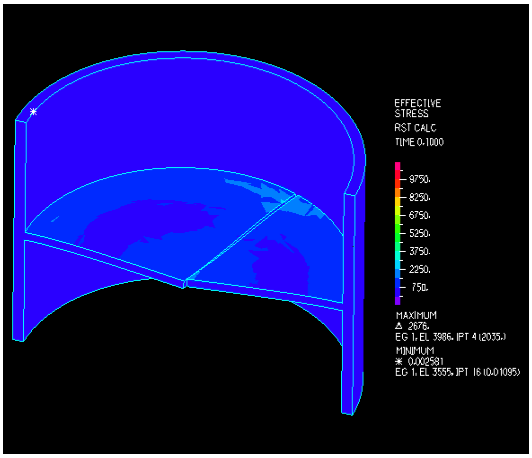


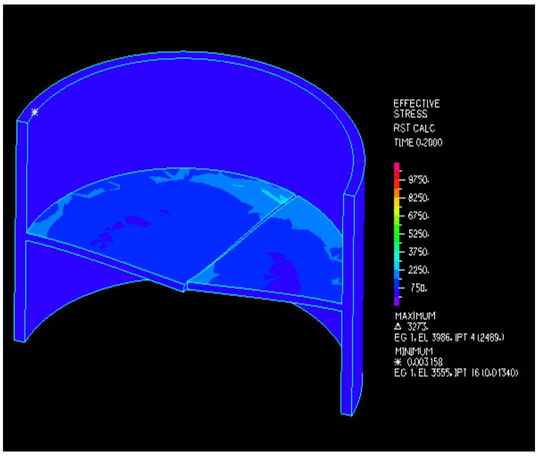
Figure 4.22: Graph Velocity comparison for Aortic Valve and Mitral Valve

Figure 4.22 performed the difference velocity value .The different percentage in the initial state of the model for MV and AV is 69.39 %. The main reason is to emphasize this theory is valid with the previous study by statement the left chamber have a greater workload because the left side of heart need to pump the blood to the entire of body, meanwhile the right chambers just work to pump the blood flow only to the lungs. Both of the models presented results with positive linear pattern on graph. Maximum value recorded AV velocity is 2.809 m/s ,while the MV maximum velocity 1.399 m/s slower compared to the AV velocity.

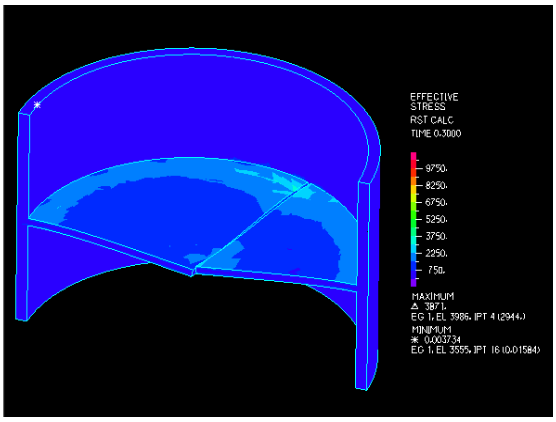
4.4.3 Effective Stress Analysis



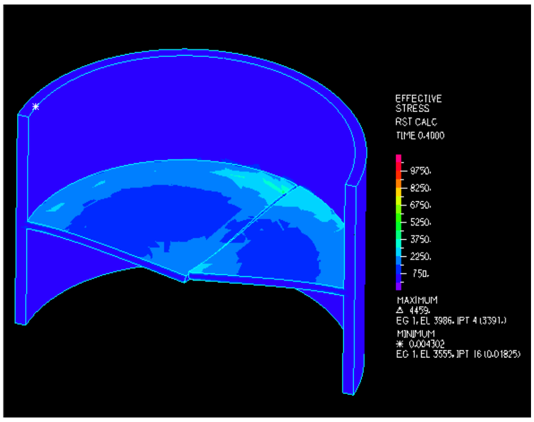
(a)



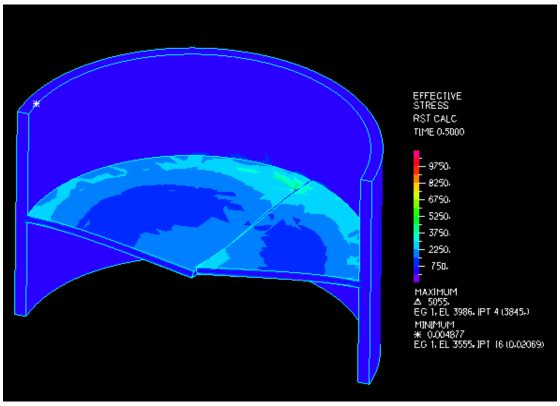
(b)



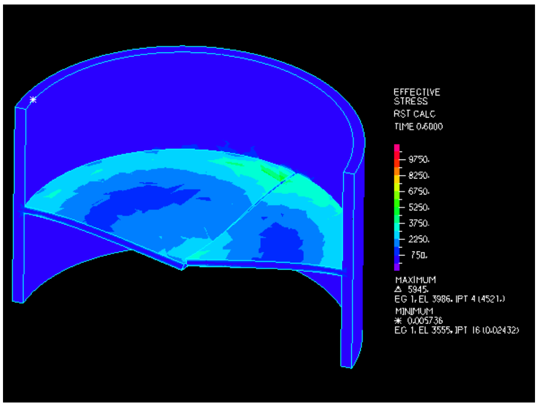
(c)



(d)



(e)



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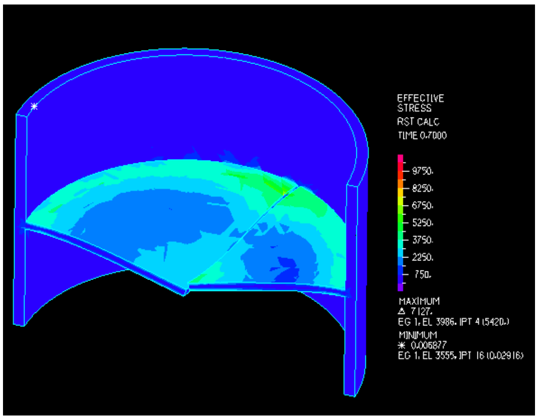
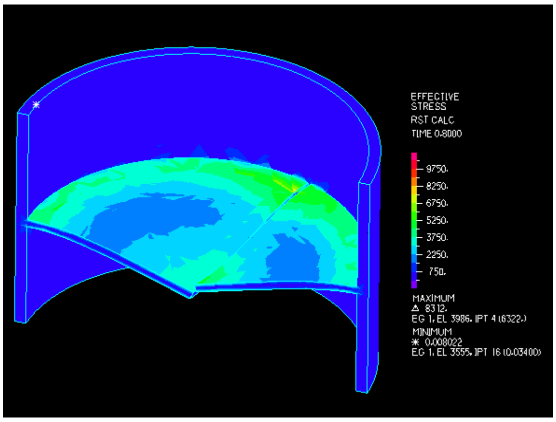
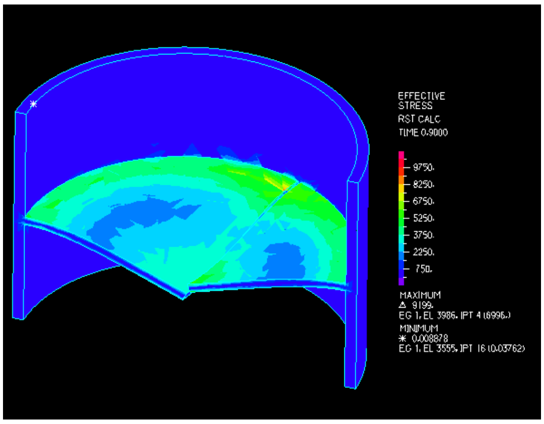


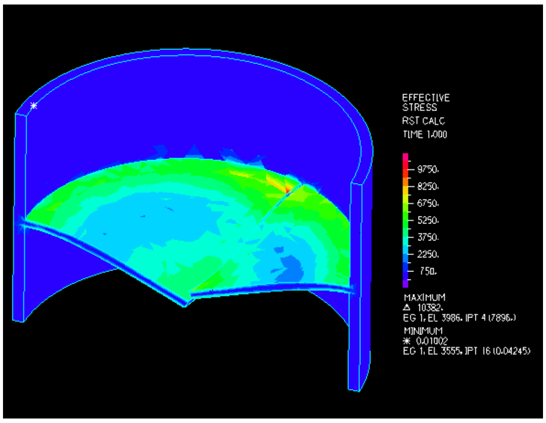
(g)



(h)



(i)



(j)

Figure 4.23: Effective stress analysis of MV model of time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j) 1.00 Sec

The mitral valve effective stress is corresponding to the pressure applied on the leaflets structure. During the initial opening leaflets stage, the effective stress occurred at the mitral valve leaflets is very small due to the small of pressure applied to it. The most striking aspect of this simulation was the dynamic motion of leaflets deformed was animated in time varying range in 1seconds.The model of mitral valve simulation results illustrates the clear view of critical area of effective stress occurred on the leaflets. When the mitral valve leaflets structure received a high pressure from the heart system, so the impact of stress on the structures also becomes higher. During the pressure act, the leaflet of mitral valve goes downward due to the pressure direction. The initial value of effective stress recorded is equal to 2381 N/m^2. Next, the changes of effective stress can see clearly by representing colors. In this result analysis, the major part of an area that had been seeing occurs at the edge of leaflets near the aortic valve wall structure. When the pressure applied to the system as much as 11.33 kPa at time 0.5 second, the effective stress value recorded is equal to 5055 N/m^2 .There is 2674 N/m^2 more than the initial condition. From this result, this 3D simulation of mitral valve can give clear views about the relation between the pressure and effective stress of the mitral valve structure. Based on the theoretical validation, when pressure increase, the stress also increased. The result showed the value of effective stress is linearly increased. The maximum effective stress achieved 10382 N/m^2 in time 1 second with the working pressure 23.99 kPa (175mmHg).

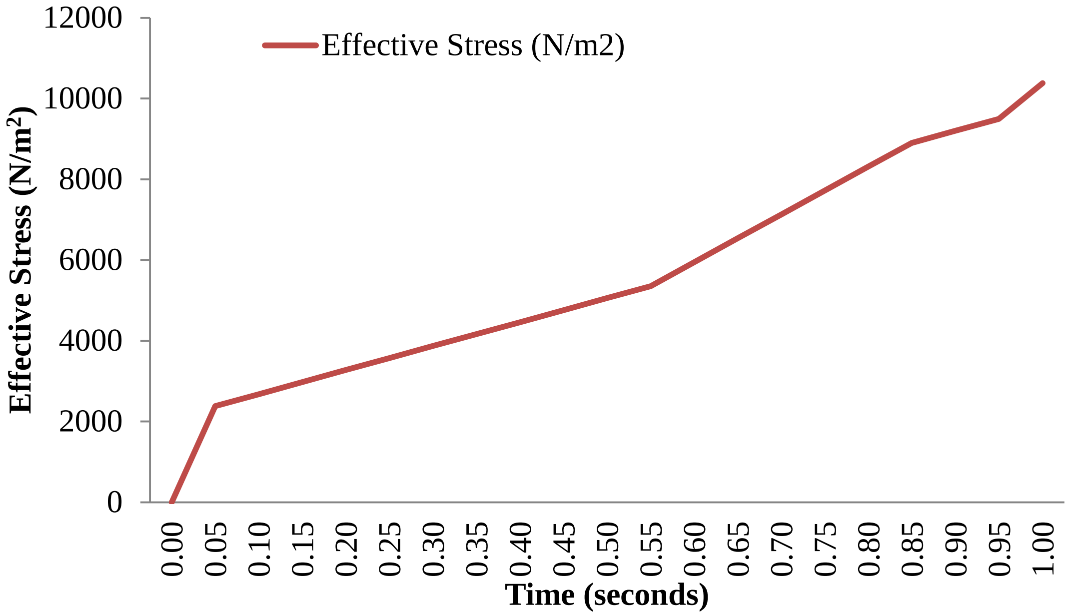
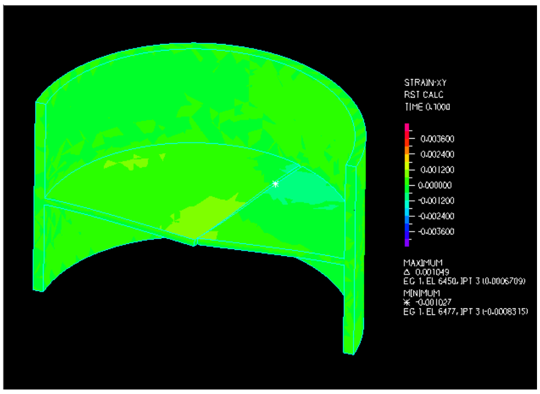


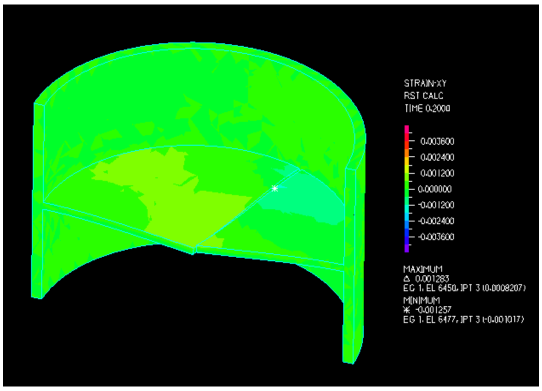
Figure 4.24: Graph Effective Stress versus time for mitral valve

The graph is a linear correlation within time 1 second. The lowest effective stress obtained when the pressure giving into the mitral valve is equal to 2381 N/m^2. From graph analysis, the effective stress value is increased when pressure acting on the mitral valve leaflets increased during the systole phase. The effective stress increased as much as 2675 N/m^2 from an initial condition with time range 0 to 0.5 second .The maximum effective stress recorded in 1 seconds is equal 10382 N/m^2 with percentages increased as much as 77.06 percent from the working pressure 5.33 kPa (40 mmHg) to 23.99 kPa (175 mmHg). The detailed of velocity value within time 1 second can be referred in Table 4.7 such in appendix D1.

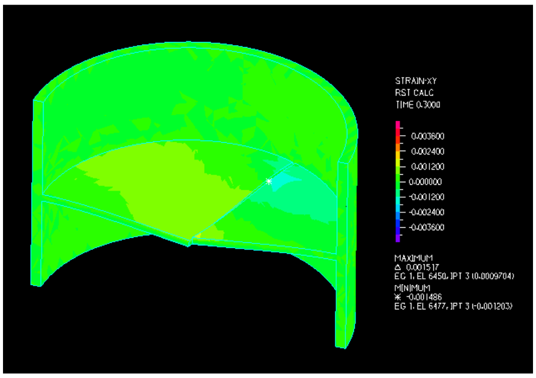
4.4.4 Strain Analysis



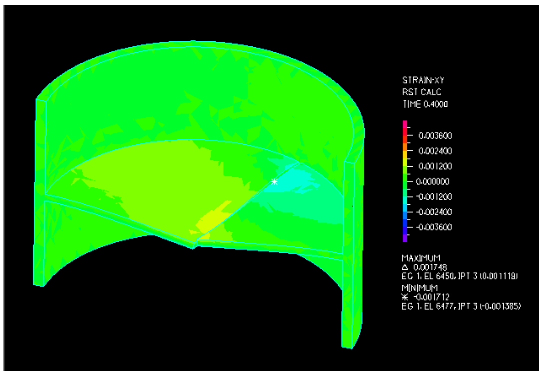
(a)



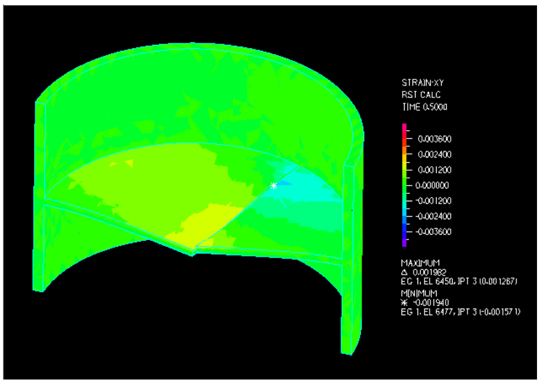
(b)



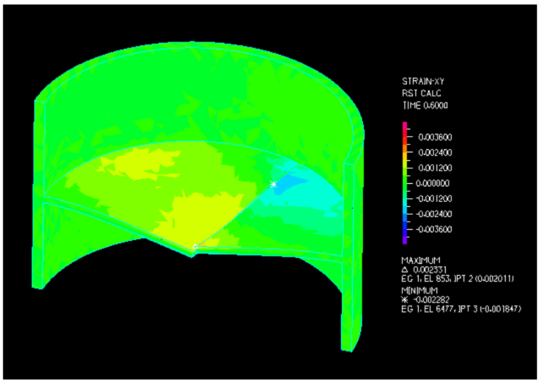
(c)



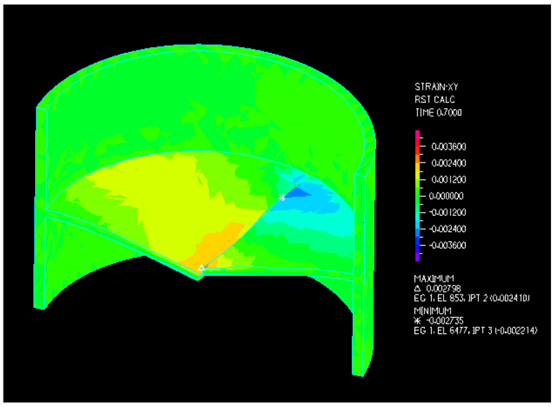
(d)

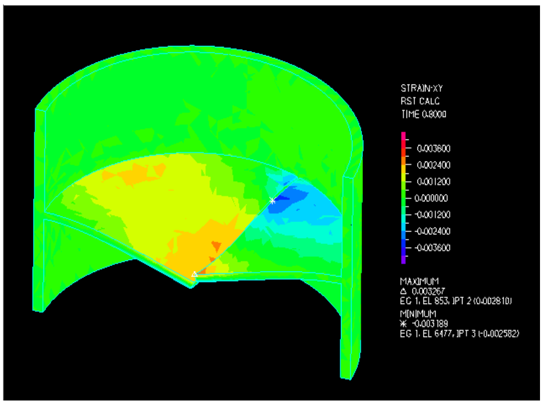


(e)

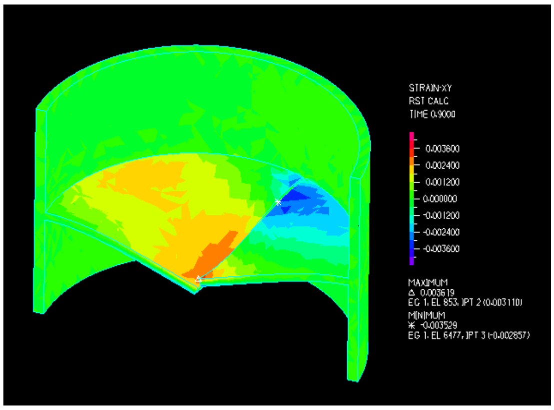


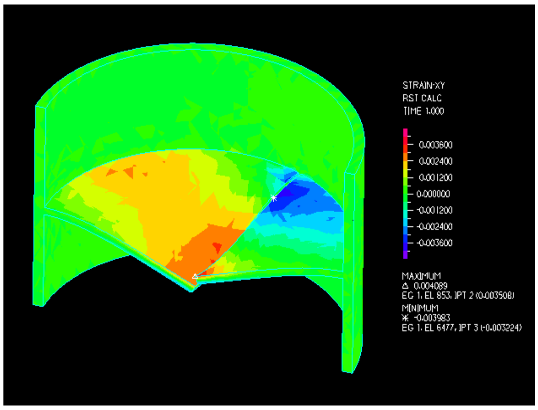
(f)





(g) (h)





(i) (j)

Figure 4.25: Strain analysis of MV model of time, (a) 0.10 Sec (b) 0.20 Sec (c) 0.30 Sec (d) 0.40 Sec (e) 0.50 Sec (f) 0.60 Sec (g) 0.70 Sec (h) 0.80 Sec (i) 0.90 Sec (j) 1.00 Sec

The deformation of leaflets brings the effect to the velocity of blood flow, stress on the leaflets structure, and strain analysis. From the illustrations figures, the aortic valve leaflets kinematics is a linear correlation with the load applied in the heart system. One of the main advantages in conducting numerical simulations is that they yield information on some characteristics of the valve such as developed strain at the leaflets structure. During time 0.1 seconds, the strain value recorded is equal to 0.0009337. The strain increased 10.99 percent from the beginning phase. The strain value percentage of strain analysis increased as much as 52.89 percent with pressure 113.32x10^3 dyns/cm^2 (11.32 kPa) within time 0.05 second to 0.5 second .The results presented the critical area most happen in the opening tip of mitral valve leaflets. From the illustration results, the maximum pressure 239x10^3 dynes/cm^2 (23.99 kPa) applied through the mitral valve showed the structure of mitral valve have achieved the maximum strain and when the pressure added to the system, the structure of the leaflets becomes filled.

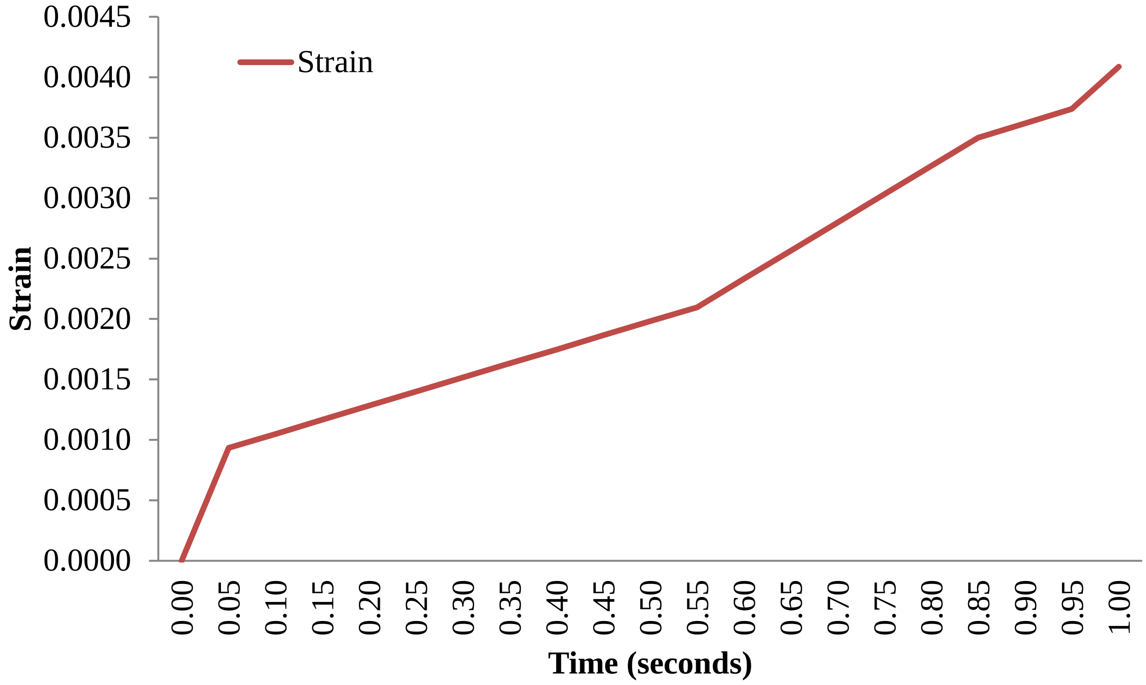


Figure 4.26: Graph Strain versus time for mitral valve

During initial pressure acting to the mitral valve leaflets structure, the strain result's value obtained from the simulation is 0.0009337. The graph showed the linear positive curve increasing within time 1 second. From graph analysis, the strain value is increased when pressure acting on the mitral valve leaflets increased during the systole phase. At time 0.1 second, the strain value differences as much as 0.0001153 compare to the initial state. The percentage strain increased as much as 47.04% from an initial condition to 0.5 second .The maximum strain recorded in 1 seconds is equal 0.004089 at the working pressure applied on the system 23.99 kPa (175mmHg) The detailed of velocity value within time 1 second can be referred appendix D1 for Table 4.8.

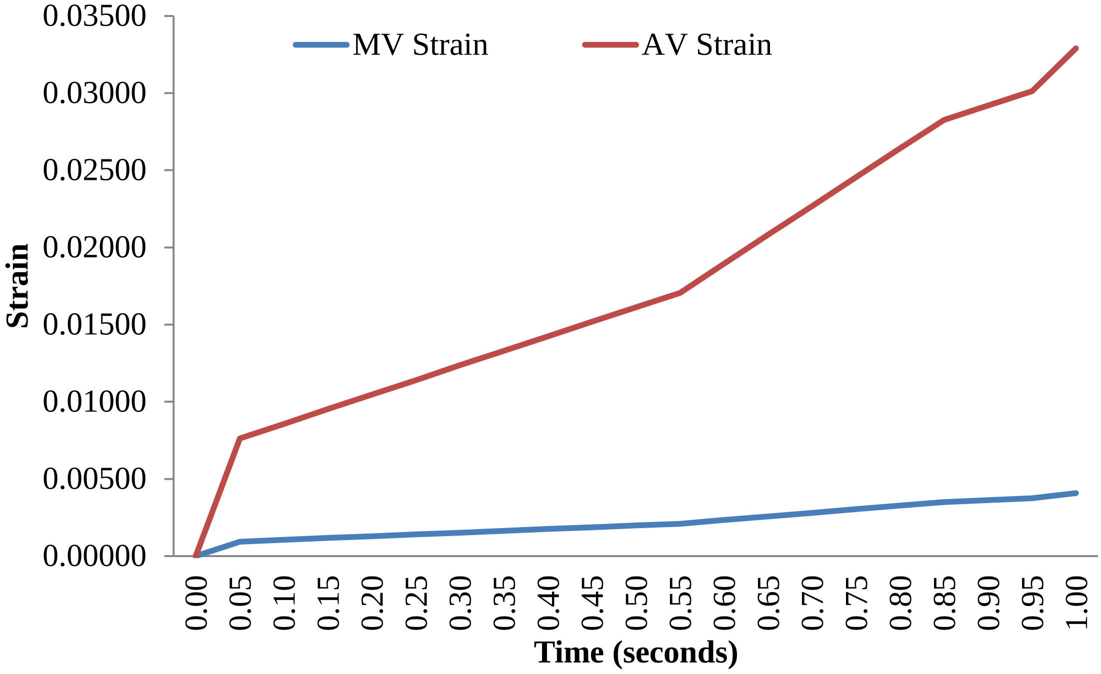


Figure 4.27: Graph Comparison strain for AV and MV

For illustration purpose, the graph pattern of strain value for both valves have been performed. The results showed the approximate linear straight line pattern for both valves. It is interesting to compare the value of effective stress between MV and AV structure. Loads were gradually applied to this model up to maximum 175mmHg (23.33 kPa) from minimum pressure 40mmHg at initial period time 0.05 seconds. During time range 0 seconds to 0.50 seconds at the working pressure not more than 90mmHg, the strain value for aortic valve give a higher reading compare to the mitral valve. From the results, the pattern on graph increased positively within time 1 seconds. The possible reason for this occurred situation is caused by the value of effective stress for the aortic valve is larger than the mitral valve model. With the constant, modulus elasticity ,the strain for AV produced more higher compare to the strain MV.The detailed strain value within time 1 seconds can be referred from appendix D1 for Table 4.9.

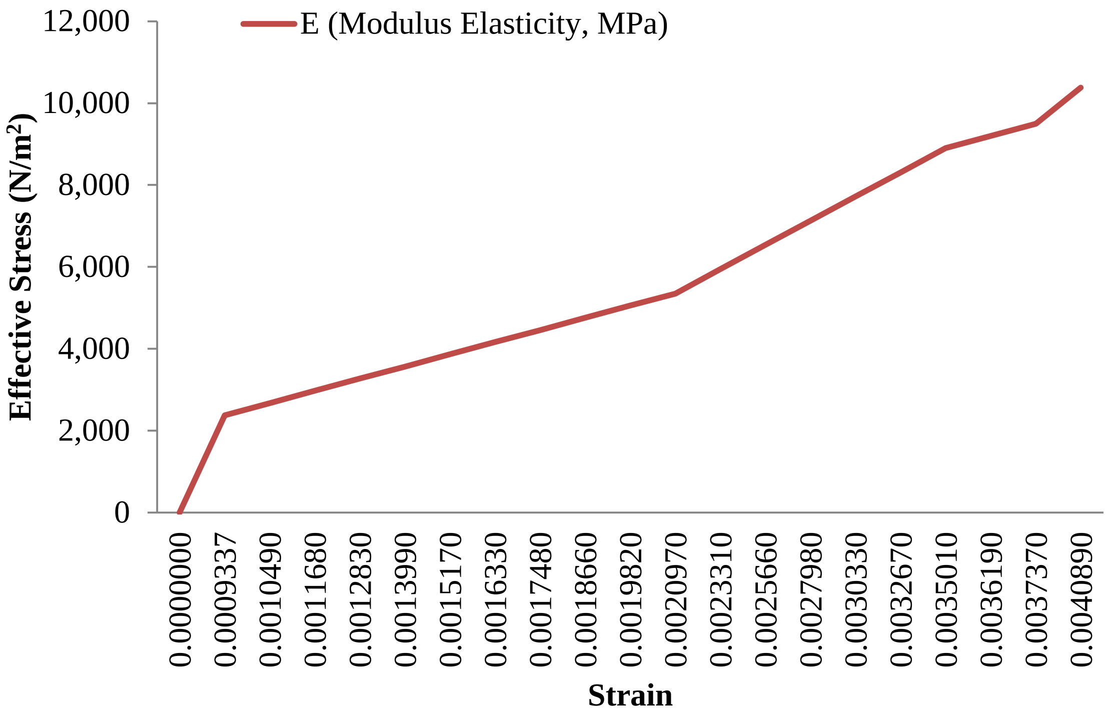


Figure 4.28: Graph stress versus strain

Graph shows the linear relationship effective stress and strain. From the simple calculation from equation, E= $\frac{\sigma}{\varepsilon}$, the value of modulus elasticity for the maximum effective stress and maximum strain is 2.53 MPa. There is a 20.94 % difference compare to the theoretical value from Espino. From this result, the findings value of young modulus elasticity can be referenced for medical practitioners to select suit material for replacement heart valve.

CHAPTER 5

CONCLUSION AND RECOMMENDATIONS

5.1 CONCLUSION

Three dimensional views for aortic and mitral valve models have been constructed by using idealized geometry based on anatomical investigation. Each model represents a significantly simplified version of the MV and AV physical. The scale of these simulations implied that the ventricle and both of models not be combined. Assumptions have been made by modeled the blood as Newtonian fluid and work in steady conditions. Here the function of the velocity blood flow via the aortic valve and mitral valve has been performed. Typically, velocity in the MV portion during the systole phase in the time range 1 seconds are lower compared with the velocity in AV. The critical area maximum stress and strain can be demonstrated clearly from simulation process. The deformation of the opening leaflets is an essentially linear correlation with pressure applied in the system. The results of the FSI simulations have shown that the maximum effective stress for aortic valve compares 84.64 % higher to the maximum effective stress for mitral valve. Moreover, theoretical mentioned the blood flow through the aortic valve is greater workload because the left side of the heart needs to pump the blood to the entire of the body. These finding information is helpful to the medical practitioners in order to decide better treatments and conditions before surgical replacement valves. Furthermore, the simulation results can be made available for doctors to detect the abnormalities in cardiac cycle.

5.2 RECOMMENDATIONS FOR THE FUTURE RESEARCH

5.2.1 Design Shape

In this project, the simple design has been implemented for heart valves, MV and AV, which are the design based on the previous study. For the planning, the scopes area in design heart valves dimension need to be specific, e.g. for Malaysian people. The collaboration with the IJN (Institut Jantung Negara) can be planned together to determine heart valves approximate dimension. The cumulative data for heart valve geometry can be a reference to a researcher in designing and do a simulation process to get a result based on the area of community The various types of valve leaflets also need to be implemented, to determine how much the efficiency of heart valves with different shape can work in normal to high condition of heart blood pressure. Meshing process is one of the important processes that need to be followed and be done properly. The size of meshing gives influenced by the results and flow illustrations. Smaller sizes of meshing give more accurate results.

5.2.2 Manipulated more parameter

In this project, the parameters for AV and MV results that have been determined from the fix Modulus Elasticity value, density and viscosity of the blood. The time range of pressure operating in this project is 1 seconds. For the next future project, the time range can be more specific, or smaller. Furthermore, the next project must to consider systole and diastole pressure condition. Other than that, others results such as the maximum displacement elongation or percentage of heart valve damage due to the maximum pressure act on the leaflets, research with unsteady condition.

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APPENDIX A1

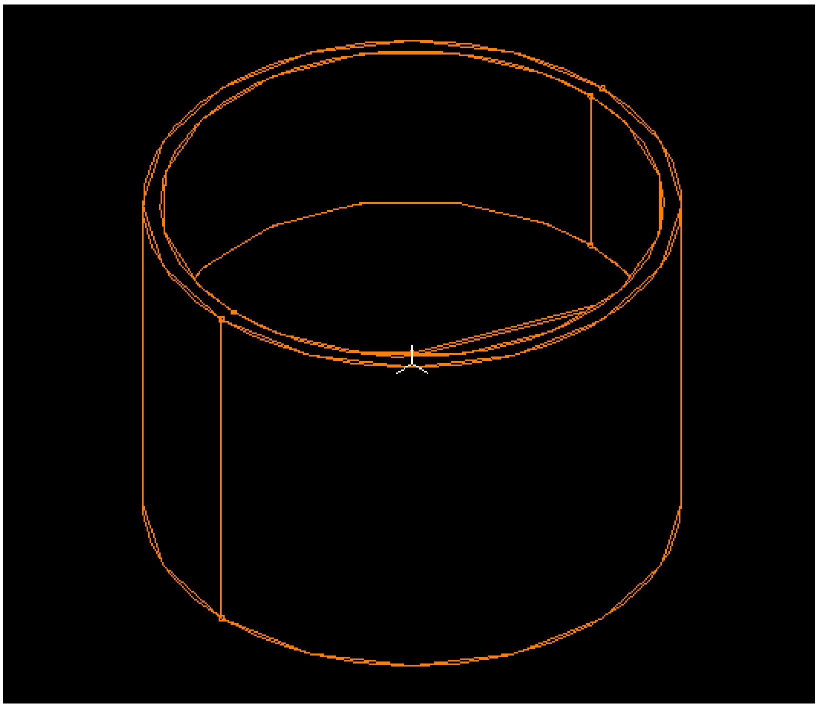


Figure A1-1: Mitral Valve Solid Design Imported to ADINA Application in File iges format

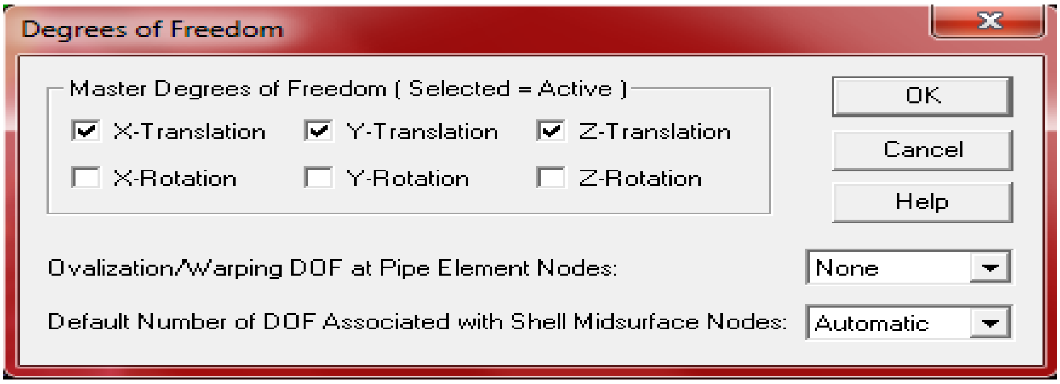


Figure A1-2: Defining the Degrees of Freedom

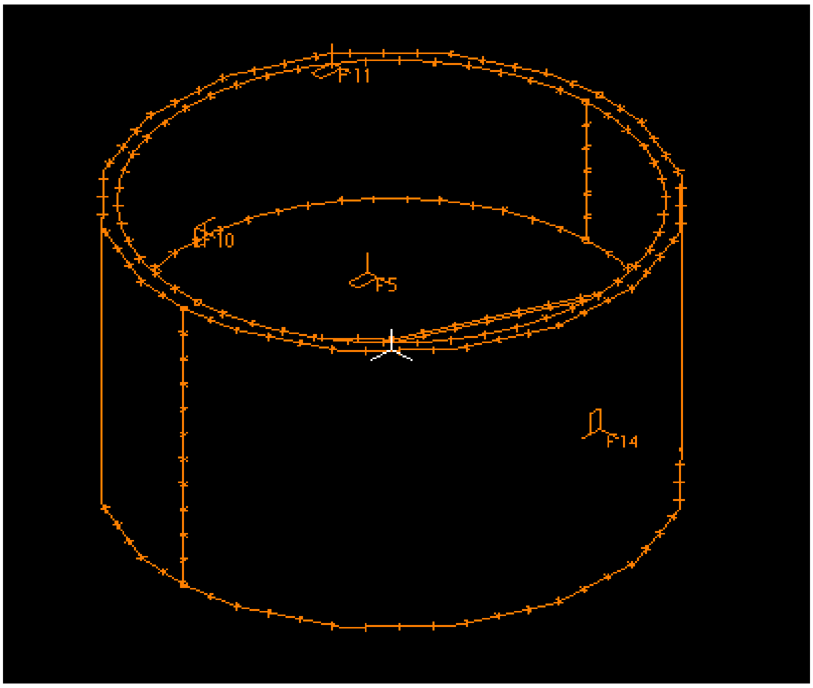


Figure A1-3: Mesh density was divided with a same length

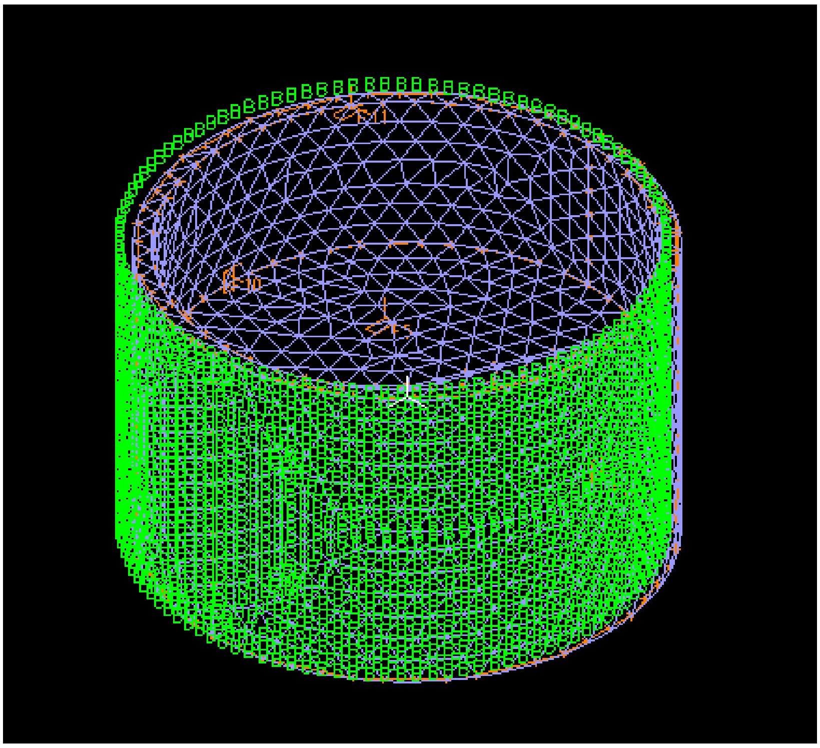


Figure A1-4: Outer wall of mitral valve has been selected as boundary conditions

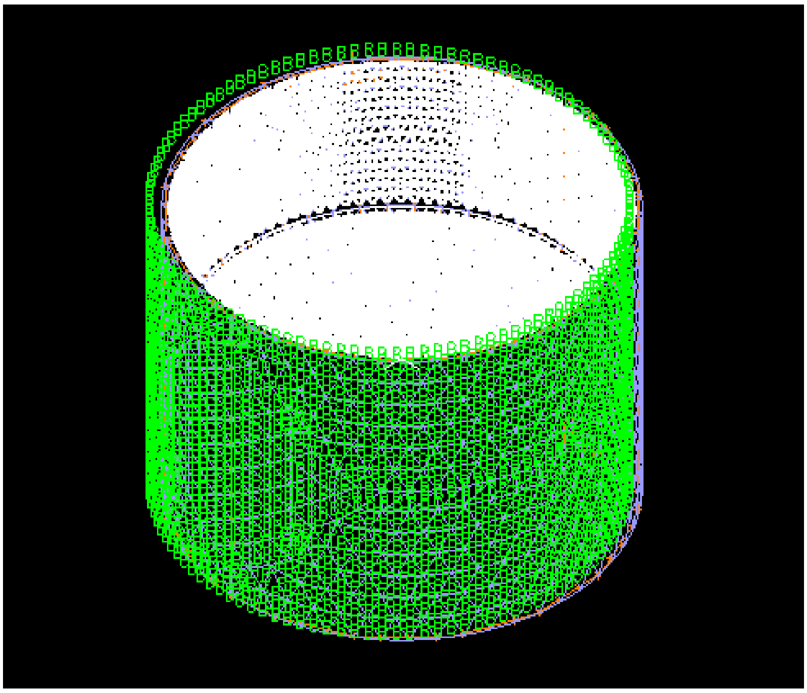


Figure A1-5: FSI for mitral valve represent in a white color

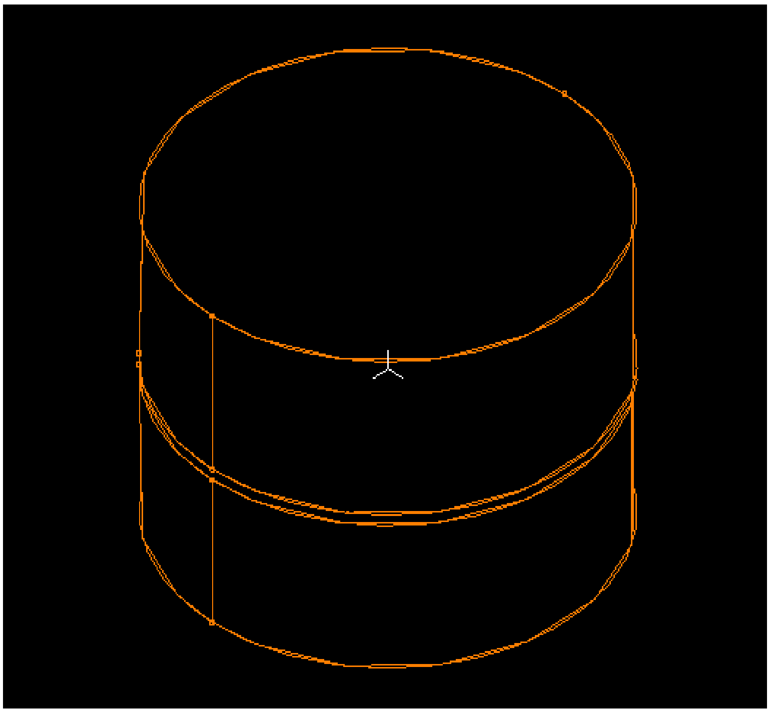


Figure A1-6: Mitral Valve Fluid Design Imported to ADINA Application in File iges format

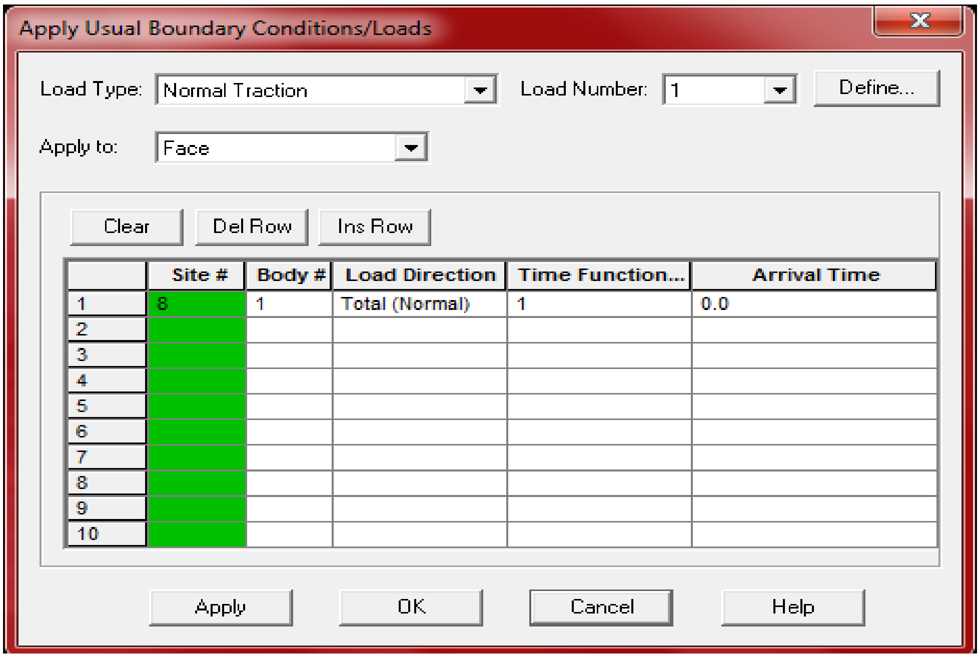


Figure A1-7: Normal traction will be used in the inlet channel

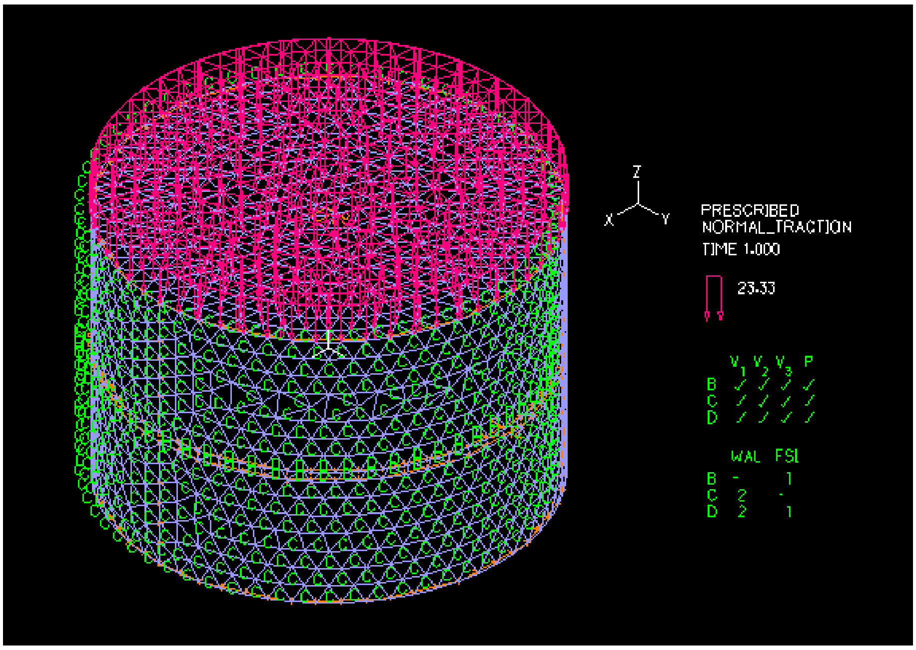


Figure A1-8: Complete process for mitral valve ADINA-F

APPENDIX A2

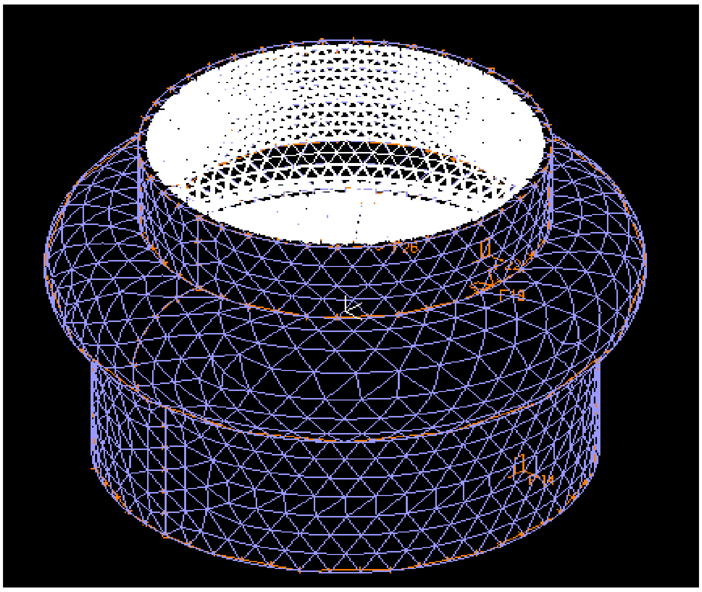


Figure A2-1: Aortic valve with complete meshing and selected FSI boundary

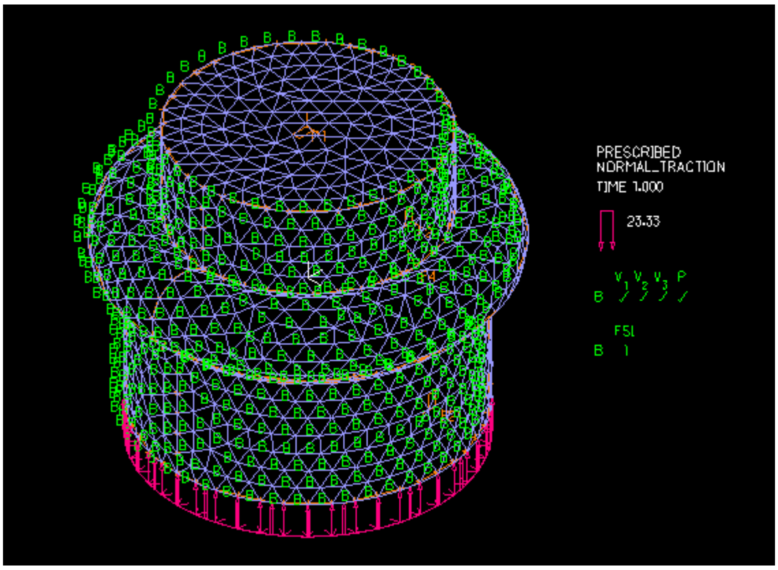


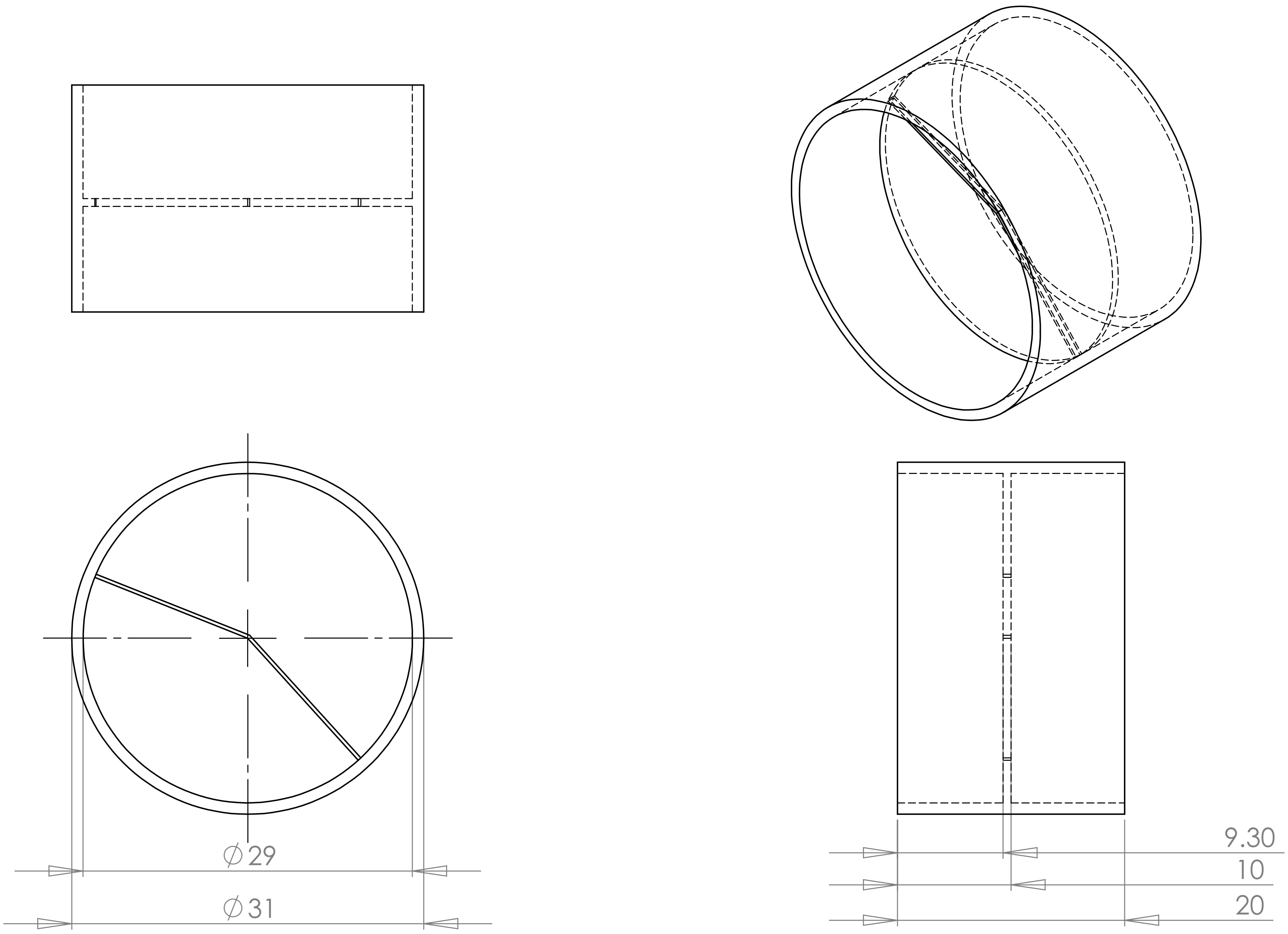
Figure A2-2: Complete process for aortic valve ADINA-F

APPENDIX B1

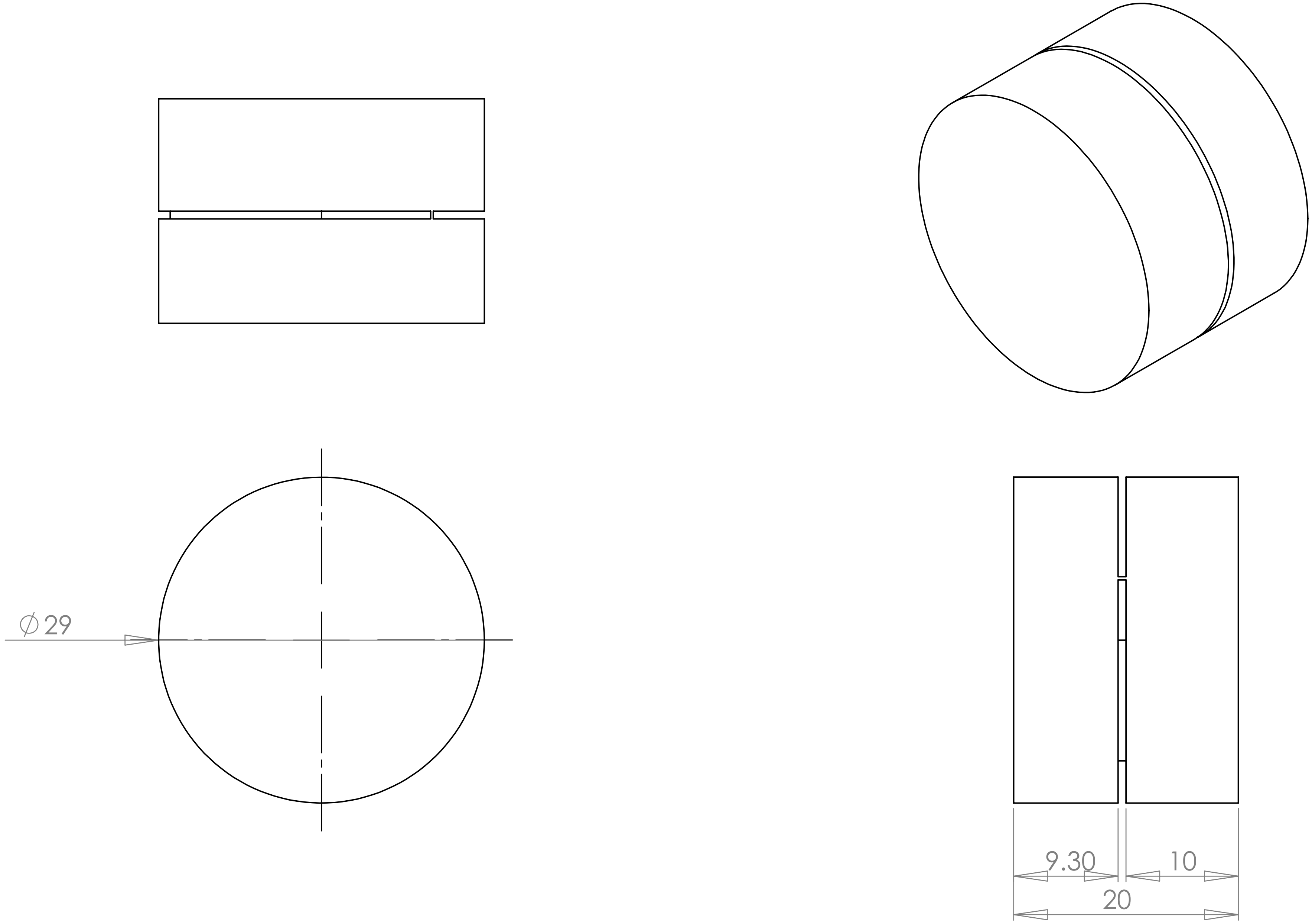
	WORK DESCRIPTION	W1	W2	W3	W4	W5	W6	W7	W8	W9	W10	W11	W12	W13	W14
TASK 1	BRIEFING ABOUT WORK PROJECT														
	DETERMINE OBJECTIVE AND SCOPES														
	PLANNING														
	ACTUAL														
TASK 2	COLLECT DATA INFORMATION ABOUT HEART DISEASE														
	COLLECT DATA ABOUT AORTIC AND MITRAL VALVE														
	PLANNING														
	ACTUAL														
TASK 3	SKECTH ROUGH DESIGN (AV & MV)														
	PROPOSED DESIGN TO SUPERVISOR														
	DESIGN DRAWING INTO 3D (SOLIDWORK)														
	PLANNING														
	ACTUAL														
	SET ALL THE PARAMETER FOR AV AND MV														
TASK 4	START MESHING THE DESIGN (ADINA)														
	RUN THE SIMULATION OF WORK PROJECT (ADINA)														
	PREDICTION OR INITIAL FINDING RESULT														
	REPORT WRITING AND PRESENTATION														
	PLANNING														
	ACTUAL														

APPENDIX B2

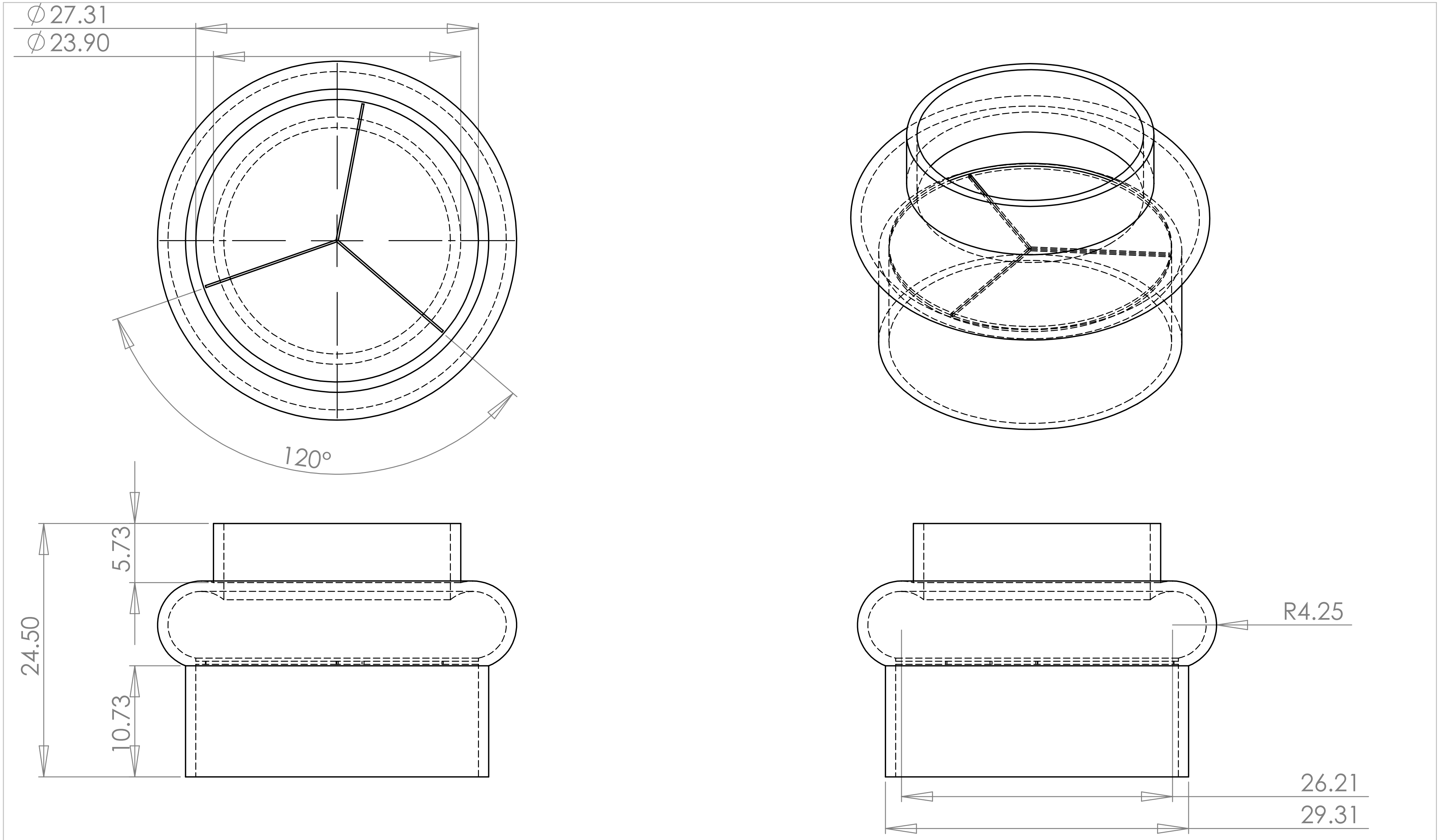
	WORK DESCRIPTION	W1	W2	W3	W4	W5	W6	W7	W8	W9	W10	W11	W12	W13	W14
TASK 5	FSI SIMULATION ON THE VALIDATION OF PARAMETERS														
	PLANNING														
	ACTUAL														
TASK 6	COLLECT AND ANALSYED DATA RESULT														
	PLANNING														
	ACTUAL														
TASK7	RESULTS DISCUSSION REPORT WRITING PRESENTATION														
	PLANNING														
	ACTUAL														



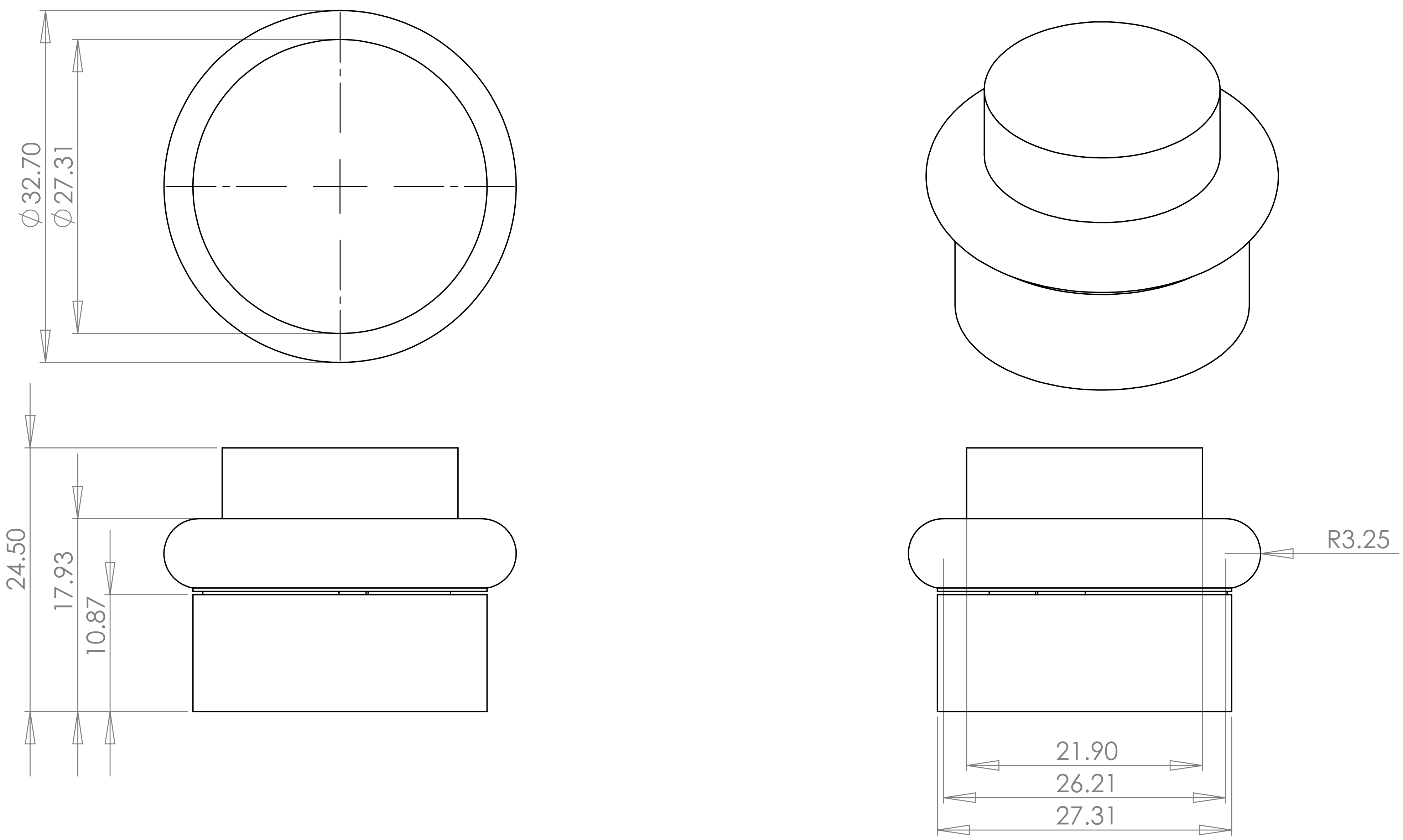
FINAL YEAR PROJECT	DRAWN	MOHAMMAD ISKANDAR OTHMAN		DIMENSIONS IN mm	TITLE:
FACULTY OF MECHANICAL ENGINEERING	CHECKED	MR AZRUL HISHAM MOHD ADIB			MITRAL VALVE DESIGN (SOLID)
	ID NO	MA 09012	DATE :	UNSPECIFIED TOLERANCES LINEAR ± 0.02 ANGULAR ± 0.5	
AORTIC VALVE DESIGN FOR WORK PROJECT TITLE ANALYSIS OF BLOOD FLOW INTO 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION		SCALE 2:1	A4		DRAWING NO. 1
					SHEET 1 of 8



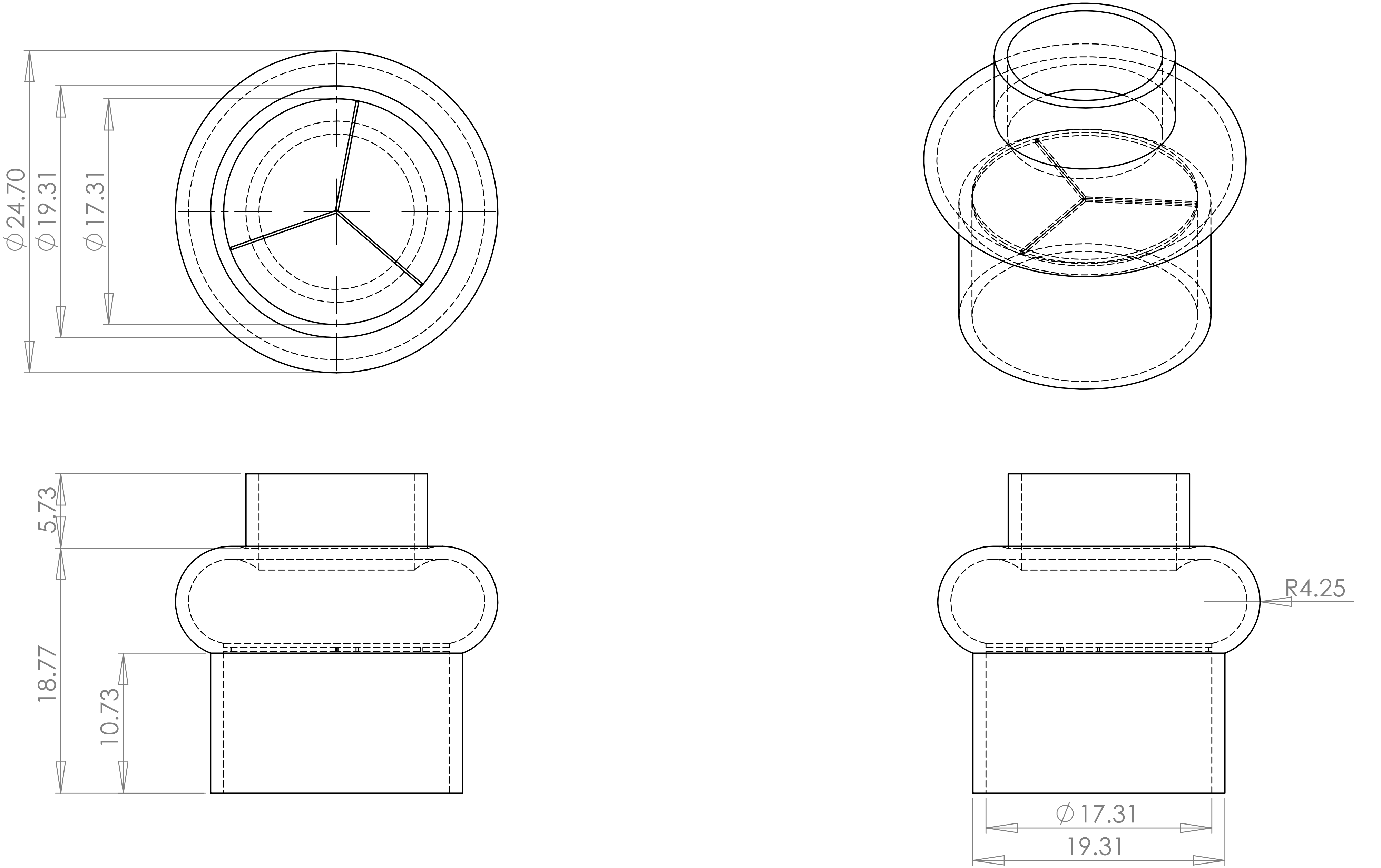
FINAL YEAR PROJECT FACULTY OF MECHANICAL ENGINEERING	DRAWN	MOHAMMAD ISKANDAR OTHMAN		DIMENSIONS IN mm	TITLE:
	CHECKED	MR AZRUL HISHAM MOHD ADIB			MITRAL VALVE DESIGN (FLUID)
	ID NO	MA 09012	DATE :	UNSPECIFIED TOLERANCES LINEAR ± 0.02 ANGULAR ± 0.5	
AORTIC VALVE DESIGN FOR WORK PROJECT TITLE ANALYSIS OF BLOOD FLOW INTO 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION		SCALE 2:1	A4		DRAWING NO. 2 — SHEET 2 of 8



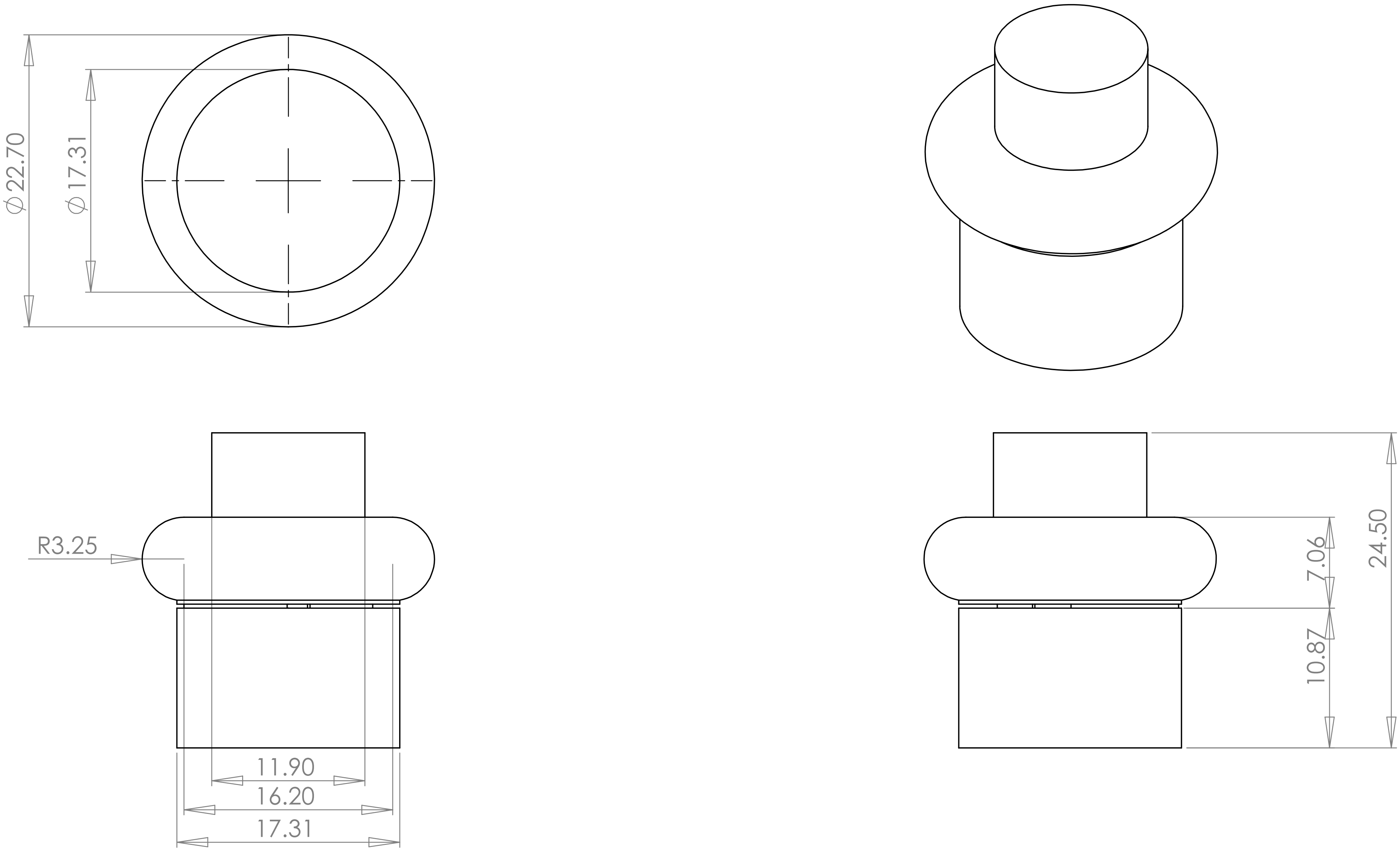
FINAL YEAR PROJECT FACULTY OF MECHANICAL ENGINEERING	DRAWN	MOHAMMAD ISKANDAR OTHMAN	DIMENSIONS IN mm	TITLE: AORTIC VALVE DESIGN NORMAL SIZE OF SINUS (SOLID)
	CHECKED	MR AZRUL HISHAM MOHD ADIB		
	ID NO	MA 09012	DATE :	
AORTIC VALVE DESIGN FOR WORK PROJECT TITLE ANALYSIS OF BLOOD FLOW INTO 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION	SCALE 2:1	A4	UNSPECIFIED TOLERANCES LINEAR ± 0.02 ANGULAR ± 0.5	DRAWING NO. 3 — SHEET 3 of 8



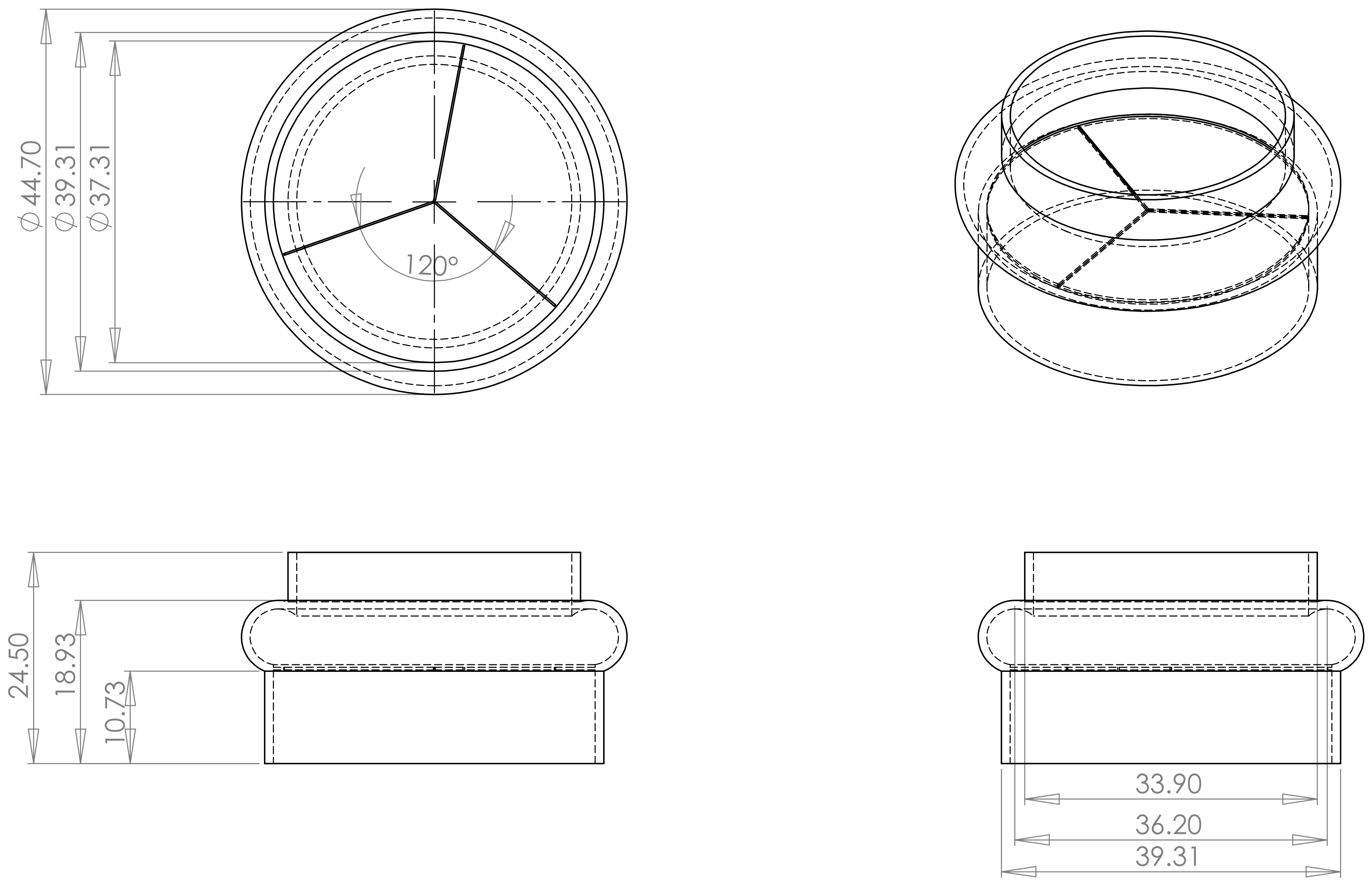
FINAL YEAR PROJECT FACULTY OF MECHANICAL ENGINEERING	DRAWN: MOHAMMAD ISKANDAR OTHMAN			DIMENSIONS IN mm	TITLE: AORTIC VALVE DESIGN NORMAL SIZE OF SINUS (FLUID)
	CHECKED: MR AZRUL HISHAM MOHD ADIB			UNSPECIFIED TOLERANCES LINEAR ± 0.02 ANGULAR ± 0.5	
	ID NO: MA 09012	DATE :			
AORTIC VALVE DESIGN FOR WORK PROJECT TITLE ANALYSIS OF BLOOD FLOW INTO 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION		SCALE 2:1	A4		DRAWING NO. 4 | SHEET 4 of 8



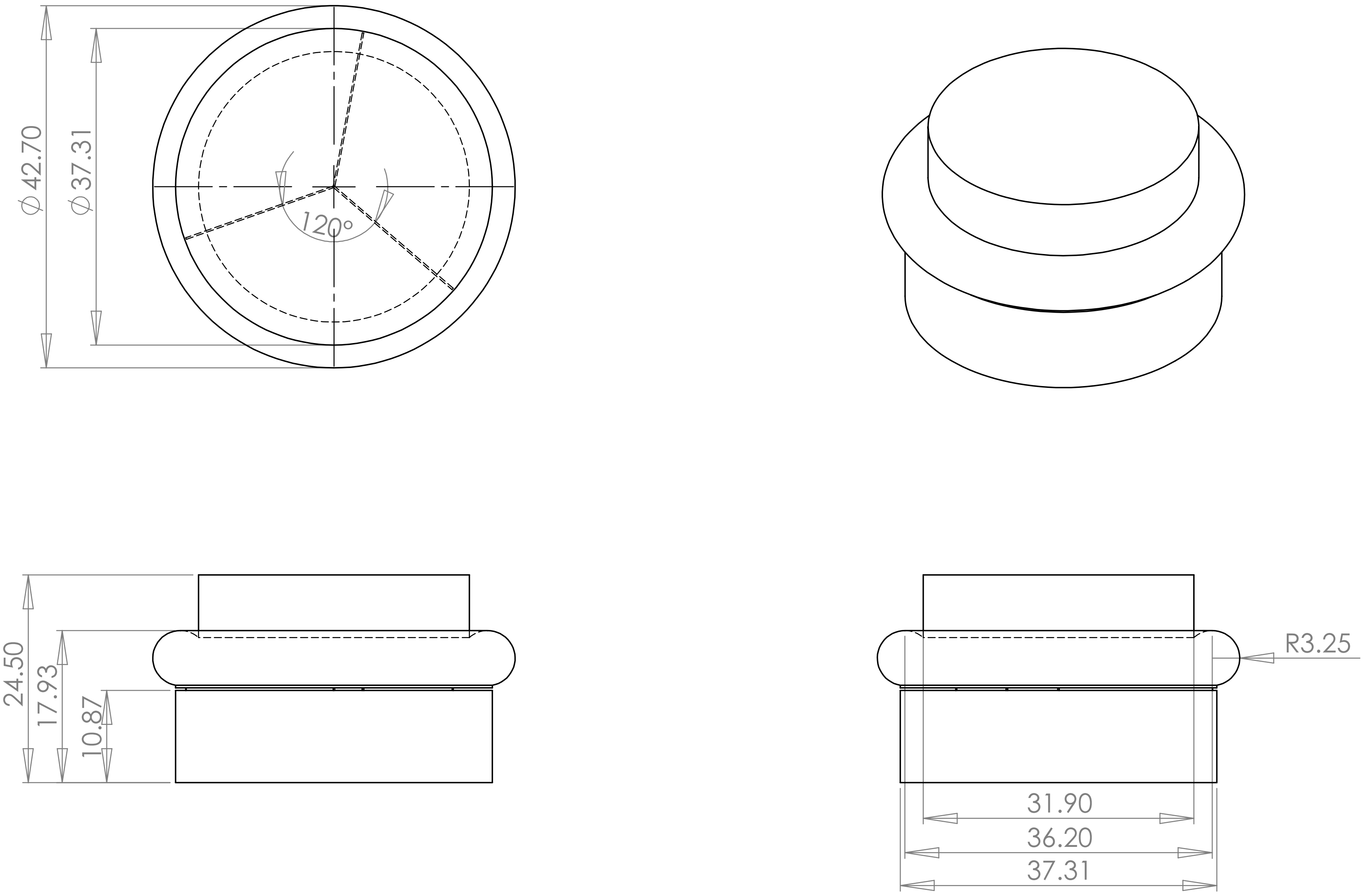
FINAL YEAR PROJECT	DRAWN	MOHAMMAD ISKANDAR OTHMAN	DIMENSIONS IN mm	TITLE: AORTIC VALVE DESIGN SMALL SIZE OF SINUS (STRUCTURE)
FACULTY OF MECHANICAL ENGINEERING	CHECKED	MR AZRUL HISHAM MOHD ADIB		
	ID NO	MA 09012	DATE :	
AORTIC VALVE DESIGN FOR WORK PROJECT TITLE ANALYSIS OF BLOOD FLOW INTO 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION	SCALE 2.5:1	A4	UNSPECIFIED TOLERANCES LINEAR ± 0.02 ANGULAR ± 0.5	DRAWING NO. 5
				SHEET 5 of 8



FINAL YEAR PROJECT	DRAWN	MOHAMMAD ISKANDAR OTHMAN	DIMENSIONS IN mm	TITLE: AORTIC VALVE DESIGN SMALL SIZE OF SINUS (FLUID)
FACULTY OF MECHANICAL ENGINEERING	CHECKED	MR AZRUL HISHAM MOHD ADIB	UNSPECIFIED TOLERANCES	
AORTIC VALVE DESIGN FOR WORK PROJECT TITLE ANALYSIS OF BLOOD FLOW INTO 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION	ID NO	MA 09012	LINEAR ± 0.02	DRAWING NO. 6
	DATE :		ANGULAR ± 0.5	SHEET 6 of 8
	SCALE 2.5:1	A4		



FINAL YEAR PROJECT FACULTY OF MECHANICAL ENGINEERING	DRAWN	MOHAMMAD ISKANDAR OTHMAN		DIMENSIONS IN mm	TITLE: AORTIC VALVE DESIGN LAGE SIZE OF SINUS (STRUCTURE)
	CHECKED	MR AZRUL HISHAM MOHD ADIB			
	ID NO	MA 09012	DATE :	UNSPECIFIED TOLERANCES LINEAR ± 0.02 ANGULAR ± 0.5	
AORTIC VALVE DESIGN FOR WORK PROJECT TITLE ANALYSIS OF BLOOD FLOW INTO 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION			SCALE 1.5:1	A4	DRAWING NO. 7
					SHEET 7 of 8



FINAL YEAR PROJECT FACULTY OF MECHANICAL ENGINEERING	DRAWN	MOHAMMAD ISKANDAR OTHMAN			DIMENSIONS IN mm	TITLE: AORTIC VALVE DESIGN LAGE SIZE OF SINUS (FLUID)
	CHECKED	MR AZRUL HISHAM MOHD ADIB				
	ID NO	MA 09012	DATE :		UNSPECIFIED TOLERANCES LINEAR ± 0.02 ANGULAR ± 0.5	
AORTIC VALVE DESIGN FOR WORK PROJECT TITLE ANALYSIS OF BLOOD FLOW INTO 3D HEART VALVE MODEL UNDER STEADY STATE CONDITION			SCALE 1.5:1	A4		DRAWING NO. 8
						SHEET 8 of 8

APPENDIX D1

Table 4.1: Value for velocity

Time (Sec)	Velocity (m/s)
0.00	0.0000
0.05	0.6881
0.10	0.7728
0.15	0.8596
0.20	0.9434
0.25	1.0270
0.30	1.1120
0.35	1.1950
0.40	1.2760
0.45	1.3600
0.50	1.4410
0.55	1.5210
0.60	1.6820
0.65	1.8410
0.70	1.9950
0.75	2.1480
0.80	2.2990
0.85	2.4470
0.90	2.5220
0.95	2.5960
1.00	2.8090

Table 4.2: Value for Effective Stress for Aortic Valve

Time (Sec)	Effective Stress (N/m^2)
0.00	0
0.05	15656
0.10	17586
0.15	19573
0.20	21499
0.25	23424
0.30	25404
0.35	27325
0.40	29244
0.45	31218
0.50	33133
0.55	35046
0.60	38924
0.65	42794
0.70	46599
0.75	50454
0.80	54272
0.85	58081
0.90	59998
0.95	61913
1.00	67612

Table 4.3: Value strain

Time (Sec)	Strain
0.00	0.000000
0.05	0.007619
0.10	0.008559
0.15	0.009526
0.20	0.010460
0.25	0.011400
0.30	0.012360
0.35	0.013300
0.40	0.014230
0.45	0.015190
0.50	0.016120
0.55	0.017050
0.60	0.018940
0.65	0.020820
0.70	0.022670
0.75	0.024550
0.80	0.026410
0.85	0.028260
0.90	0.029190
0.95	0.030120
1.00	0.032900

Table 4.4: Value for Modulus Elasticity (E) Aortic Valve

Time (Sec)	Effective Stress (N/m^2)	Strain	E (Modulus Elasticity, Pa)
0.00	0.000	0.000000	0.00000000
0.05	15656	0.007619	2054862.84
0.10	17586	0.008559	2054679.28
0.15	19573	0.009526	2054692.42
0.20	21499	0.010460	2055353.73
0.25	23424	0.011400	2054736.84
0.30	25404	0.012360	2055339.81
0.35	27325	0.013300	2054511.28
0.40	29244	0.014230	2055094.87
0.45	31218	0.015190	2055167.87
0.50	33133	0.016120	2055397.02
0.55	35046	0.017050	2055483.87
0.60	38924	0.018940	2055121.44
0.65	42794	0.020820	2055427.47
0.70	46599	0.022670	2055535.95
0.75	50454	0.024550	2055152.75
0.80	54272	0.026410	2054979.17
0.85	58081	0.028260	2055237.08
0.90	59998	0.029190	2055429.94
0.95	61913	0.030120	2055544.49
1.00	67612	0.032900	2055075.99

Table 4.5: Nodal Pressure value for Aortic Valve

Time (Sec)	Nodal Pressure (kN/m^2)
0.00	0.0000
0.05	5.3833
0.10	6.0493
0.15	6.7353
0.20	7.4010
0.25	8.0666
0.30	8.7522
0.35	9.4175
0.40	10.0826
0.45	10.7678
0.50	11.4327
0.55	12.0975
0.60	13.4468
0.65	14.7957
0.70	16.1239
0.75	17.4718
0.80	18.8093
0.85	20.1464
0.90	20.8199
0.95	21.4933
1.00	23.5026

Table 4.6: Value Maximum velocity for mitral valve

Time (Sec)	Velocity (m/s)
0.00	0.0000
0.05	0.2106
0.10	0.2376
0.15	0.2703
0.20	0.3031
0.25	0.3371
0.30	0.3732
0.35	0.4093
0.40	0.4465
0.45	0.4859
0.50	0.5252
0.55	0.5655
0.60	0.6506
0.65	0.7397
0.70	0.8315
0.75	0.9286
0.80	1.029
0.85	1.133
0.90	1.186
0.95	1.241
1.00	1.410

Table 4.7: Value effective Stress for Mitral Valve

Time (Sec)	Effective Stress (N/m^2)
0.00	0
0.05	2381
0.10	2676
0.15	2979
0.20	3273
0.25	3568
0.30	3871
0.35	4165
0.40	4459
0.45	4761
0.50	5055
0.55	5349
0.60	5945
0.65	6541
0.70	7127
0.75	7722
0.80	8312
0.85	8902
0.90	9199
0.95	9496
1.00	10382

Table 4.8: Value strain for mitral valve

Time (Sec)	Strain
0.00	0.0000000
0.05	0.0009337
0.10	0.001049
0.15	0.001168
0.20	0.001283
0.25	0.001399
0.30	0.001517
0.35	0.001633
0.40	0.001748
0.45	0.001866
0.50	0.001982
0.55	0.002097
0.60	0.002331
0.65	0.002566
0.70	0.002798
0.75	0.003033
0.80	0.003267
0.85	0.003501
0.90	0.003619
0.95	0.003737
1.00	0.004089

Table 4.9: Value for Modulus Elasticity (E) Aortic Valve

Time (Sec)	Effective Stress (N/m^2)	Strain	E (Modulus Elasticity, Pa)
0.00	0	0	0
0.05	2381	0.000934	2550069.62
0.10	2676	0.001049	2551000.95
0.15	2979	0.001168	2550513.7
0.20	3273	0.001283	2551052.22
0.25	3568	0.001399	2550393.14
0.30	3871	0.001517	2551746.87
0.35	4165	0.001633	2550520.51
0.40	4459	0.001748	2550915.33
0.45	4761	0.001866	2551446.95
0.50	5055	0.001982	2550454.09
0.55	5349	0.002097	2550786.84
0.60	5945	0.002331	2550407.55
0.65	6541	0.002566	2549103.66
0.70	7127	0.002798	2547176.55
0.75	7722	0.003033	2545994.07
0.80	8312	0.003267	2544230.18
0.85	8902	0.003501	2542702.09
0.90	9199	0.003619	2541862.39
0.95	9496	0.003737	2541075.73
1.00	10382	0.004089	2539007.09

Table 4.9: Value Comparison for AV and MV

Time (Sec)	MV Strain	AV Strain
0.00	0.00000	0.000000
0.05	0.00093	0.007619
0.10	0.00105	0.008559
0.15	0.00117	0.009526
0.20	0.00128	0.010460
0.25	0.00140	0.011400
0.30	0.00152	0.012360
0.35	0.00163	0.013300
0.40	0.00175	0.014230
0.45	0.00187	0.015190
0.50	0.00198	0.016120
0.55	0.00210	0.017050
0.60	0.00233	0.018940
0.65	0.00257	0.020820
0.70	0.00280	0.022670
0.75	0.00303	0.024550
0.80	0.00327	0.026410
0.85	0.00350	0.028260
0.90	0.00362	0.029190
0.95	0.00374	0.030120
1.00	0.00409	0.032900

Table 4.10: Value Nodal Pressure for Mitral Valve

Time (Sec)	Nodal Pressure (kN/m^2)
0.00	0.0000
0.05	5.3352
0.10	5.9969
0.15	6.6788
0.20	7.3409
0.25	8.0032
0.30	8.6858
0.35	9.3486
0.40	10.0115
0.45	10.6947
0.50	11.3580
0.55	12.0215
0.60	13.3692
0.65	14.7175
0.70	16.0464
0.75	17.3961
0.80	18.7363
0.85	20.0770
0.90	20.7526
0.95	21.4283
1.00	23.4462

Table 4.11: Value velocity comparison for Aortic Valve and Mitral Valve

Time (Sec)	MV Velocity (m/s)	AV Velocity (m/s)
0.00	0.0000	0.0000
0.05	0.2106	0.6881
0.10	0.2376	0.7728
0.15	0.2703	0.8596
0.20	0.3031	0.9434
0.25	0.3371	1.0270
0.30	0.3732	1.1120
0.35	0.4093	1.1950
0.40	0.4465	1.2760
0.45	0.4859	1.3600
0.50	0.5252	1.4410
0.55	0.5655	1.5210
0.60	0.6506	1.6820
0.65	0.7397	1.8410
0.70	0.8315	1.9950
0.75	0.9286	2.1480
0.80	1.0290	2.2990
0.85	1.1330	2.4470
0.90	1.1860	2.5220
0.95	1.2410	2.5960
1.00	1.4100	2.8090

Table 4.12: Comparison Effective Stress and Strain for model AV and MV

Time (Sec)	**MV Effective Stress (N/m^2)**	**MV Strain**	**AV Effective Stress (N/m^2)**	**AV Strain**	**Validation**	**Strain**
0.00	0	0.00000	0	0.000000	0	0.000000
0.05	2381	0.00093	15656	0.007619	1500	0.007619
0.10	2676	0.00105	17586	0.008559	3000	0.008559
0.15	2979	0.00117	19573	0.009526	4500	0.009526
0.20	3273	0.00128	21499	0.010460	6000	0.010460
0.25	3568	0.00140	23424	0.011400	7500	0.011400
0.30	3871	0.00152	25404	0.012360	9000	0.012360
0.35	4165	0.00163	27325	0.013300	10500	0.013300
0.40	4459	0.00175	29244	0.014230	12000	0.014230
0.45	4761	0.00187	31218	0.015190	13500	0.015190
0.50	5055	0.00198	33133	0.016120	15000	0.016120
0.55	5349	0.00210	35046	0.017050	16500	0.017050
0.60	5945	0.00233	38924	0.018940	18000	0.018940
0.65	6541	0.00257	42794	0.020820	19500	0.020820
0.70	7127	0.00280	46599	0.022670	21000	0.022670
0.75	7722	0.00303	50454	0.024550	22500	0.024550
0.80	8312	0.00327	54272	0.026410	24000	0.026410
0.85	8902	0.00350	58081	0.028260	25500	0.028260
0.90	9199	0.00362	59998	0.029190	27000	0.029190
0.95	9496	0.00374	61913	0.030120	28500	0.030120
1.00	10382	0.00409	67612	0.032900	30000	0.032900

Table 4.13: Nodal Pressure Comparison for model AV and MV

Time (sec)	MV Nodal Pressure (kN/m^2)	MV (mmHg)	AV Nodal Pressure (kN/m^2)	AV (mmHg)	Validation
0.00	0.00000	0.00	0.00000	0.00	0
0.05	5.33520	40.02	5.38330	40.38	5
0.10	5.99690	44.99	6.04930	45.38	15
0.15	6.67880	50.10	6.73530	50.53	28
0.20	7.34090	55.07	7.40100	55.52	38
0.25	8.00320	60.04	8.06660	60.51	42
0.30	8.68580	65.16	8.75220	65.66	50
0.35	9.34860	70.13	9.41750	70.65	60
0.40	10.01150	75.11	10.08260	75.64	68
0.45	10.69470	80.23	10.76780	80.78	79
0.50	11.35800	85.21	11.43270	85.77	88
0.55	12.02150	90.18	12.09750	90.75	90
0.60	13.36920	100.29	13.44680	100.88	86
0.65	14.71750	110.41	14.79570	111.00	87
0.70	16.04640	120.38	16.12390	120.96	88
0.75	17.39610	130.50	17.47180	131.07	89
0.80	18.73630	140.56	18.80930	141.11	98
0.85	20.07700	150.62	20.14640	151.14	125
0.90	20.75260	155.68	20.81990	156.19	118
0.95	21.42830	160.75	21.49330	161.24	105
1.00	23.44620	175.89	23.50260	176.31	100