# Blood Flow Simulation in Descending Thoracic Aorta and the Effect of Geometry and Pathology on Flow Behavior

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# ABSTRACT

Nowadays, the importance of detecting reasons for pathology of blood vessel and their removal is increasing. One of the best solutions which can be brought up is CFD blood flow simulation for human in vivo condition. This study concerns blood flow simulation in descending thoracic aorta as comparative investigation to find the effect of the geometry and conditions (i.e., aneurysm and atherosclerosis) on flow behavior, without and with considering non-linear isotropic elastic vessel wall. In this analysis, laminar transient flow is used for simulation according to the time variant velocity and pressure value in boundary conditions. Carreau_Yasuda non-Newtonian rheology is used as the blood property to find its effect as the shear stress on the vessels wall. Finally, results showed the effect of the geometry and pulsatile flow on formation of vortex flow in different regions of the vessel as well as their effects on flow stagnation which can be led to plaque accumulation and diseases. The distribution of wall shear stresses (WSS) and their maximum and minimum regions also showed the importance of this parameter on rapture and diseases incidence in those regions.

**Keywords:** Blood; Flow; Pulsatile; Simulation; Aneurysm; Atherosclerosis; Thoracic; Iliac; Renal

# ÖZ

Günümüzde, kan damarı ve bunların kaldırılmasına yönelik patolojik nedenlerin tespitinin önemi artmaktadır. İnsan vüaudunumun iç durumu için, bugüne kadar getirilebilen en iyi çözümlerden biridir. Bu çalışma, doğrusal olmayan izotropik damar duvarı dikkate alınmadan, akış ve davranışı üzerindeki şantlarin ve hastalıkların (anevrizma ve ateroskleroz) etkisini öğrenmek için karşılaştırma ve soruşturma olarak atardamardaki artan kan akış simülasyonu ile ilgilidir. Bu analizde, sınır koşullarında değisken hız ve basınc değerlerine göre simülasyon için, tabakalı ve geçiçi akış getirildi. Kan özellikleri, sınır duvarında yırtılma stresi olarak etkisini öğrenmek için, Carreau_Yasuda newtoniyan olmayan rheolojisi seçilmiştir. Sonuç olarak, plak birimi ve hastalıklara neden olan ve akış durgunluğunun üzerindeki etkisi farklı bölgedeki girdap akışının, insan üzerindeki geometri ve pulsatil akımının etkisidir. Duvar kayma gerilemeleri (WSS)ve onun maksimum ve minimum bölgelere dağılımı da bu parametrelerin önemini ve bu bölgelerde olan hastalığın olasılığını göstermektedir.

**AnahtarKalimeler**: Kan, akış, pulsatil, simülasyon, anevrizma, ateroskleroz, göğüs, kalça kemiği, böbrek

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| | |
|---|---|
| FSI | Fluid Solid Interaction |
| WSS | Wall Shear Stress |
| RMS | Residual Mean Square |
| C-Y | Carreau_Yasuda |
| CFD | Computational Fluid Dynamic |
| 2-D | Two-dimensional |
| 3-D | Three-dimensional |

# LIST OF SYMBOLS

$\dot{m}$ Mass Flow Rate [kg/s]

n Power law index [dimensionless]

P Pressure [mm Hg], [Pa]

u Cartesian velocity component, in X-direction [m/s]

v Cartesian velocity component, in Y-direction [m/s]

w Cartesian velocity component, in Z-direction [m/s]

Greek Symbols:

$\mu$ Shear viscosity [Pa s]

$\mu_0$ Low shear viscosity [Pa s]

$\mu_\infty$ High shear viscosity [Pa s]

$\lambda$ Time constant [s]

$\dot{\Upsilon}$ Shear strain rate [$s^{-1}$]

$\alpha$ Yasuda exponent [dimensionless]

$\rho$ Density [$kg/m^3$]

$\tau$ Shear stress [$N/m^2$]

σ Stress [$N/m^2$]

ε Strain [dimensionless]

# Chapter 1

# INTRODUCTION

Heart is one of the most important organs in our body which functions like a pump. Heart pumps blood being delivered to the different parts of body, cells, tissues, etc., which provide them with nutrition and oxygen, and then deoxygenated blood collects waste from cells and tissues returning to the heart in order to be prepared for next circulation. So, the cardiovascular mechanism is like a network which is made up of heart, vessels, arteries and many other different parts for circulating blood.

Since circulatory system and its parts are important in human body, there are some phenomena which affect normal blood circulation, such as aneurysm and atherosclerosis that mostly occur in aorta and its branches where a large amount of blood pass through. These phenomena occur at some part of the aorta, for example abdominal part, in such way that cause vessel wall diameter increases (aneurysm) or decreases (atherosclerosis) and loses its structural property since wall structure is elastic. Thus, the results might be vessel wall rapture especially at infected region or heart stroke.

Hence, blood flow simulation in circulatory system will be applicable to distinguish blood behavior at different parts and its influence on the wall structure during both healthy and illness conditions. The comparison between these results will show the disease effects on flow behavior which can be applicable for surgery operations to cure the vessel condition and also simulate artificial vessel.

## 1.1 Objective of Thesis

As it is mentioned before, most of the investigation on blood flow in cardiovascular circulatory system are performed in aorta and its branches, especially in descending aorta where abdominal aorta is a part of it. Therefore, this study is focused on blood flow simulation in descending aorta and its four main branches, left renal, right renal, left iliac and right iliac; although these are not all descending aorta's branches but also they are four most important branches as explored by other authors before.

To investigate the effect of vessel wall on flow, FSI (fluid solid interaction) condition is considered for this study because in reality the flow domain for blood is vessel which is considered as elastic one. So, there are two different phases as solid and fluid that can influence each other, and in vivo it is exactly as such because fluid contains some properties like pressure which can lead to the wall elongation. Then, it will go back to its primary condition because of its elasticity.

As for solid part, i.e. blood vessel structure, a material property treating as non-linear elasticity is defined according to data that are extracted from different tensile test on aorta specimen. A transient structural analysis system which is based on Finite Element Method (FEM) is used for simulating the solid part.

In order to perform fluid analysis, laminar and transient flow characteristics are used because of pulsatile nature of the blood velocity profile which is coupled with a pressure profile. So, it makes the velocity to be changed by time which is known as a transient condition. A non-Newtonian rheology is conducted to simulate blood property since its viscosity is not constant. Carreau_Yasuda (C-Y) is the model used to simulate blood as non-Newtonian fluid, although there are some other models such as Power law, Casson, etc. But also this is the most appropriate model which does not have other models disadvantages.

Computer simulation is utilized in order to simulate this FSI complicated problem. ANSYS software which has the best ability to overcome all condition was utilized. Two appropriate analysis systems are selected from ANSYS for each solid and fluid parts, Static structural and CFX.

## 1.2 Thesis Organization

This thesis is organized in 5 chapters including introduction, literature review, system modeling, results and discussion, and conclusion, which present computer-3D blood flow simulations in descending aorta during aneurysm and atherosclerosis phenomena.

First chapter is about the blood flow simulation and basic information of human blood circulatory system and its related diseases to make readers familiar with biomechanics of body. Literature review describes the background about blood flow numerical and computer simulations which different authors have been worked on before. Chapter 3 explains system modeling, mathematical background and computer modeling to define the problem more detailed and what is done to simulate the problem accurately. Chapter 4 includes results and discussion to show how well simulation is done and if they are acceptable as expected or not. Chapters 5 presents conclusion according to the results and limitations and present the future work.

# Chapter 2

# LITERATURE REVIEW

Pulsatile blood flow simulation in abdominal aorta and iliac arteries has been investigated by M. Alishahi et al.[1]. They studied two different healthy and tapered conditions. In their simulation they have used FSI (fluid solid interaction) condition in which two wall properties, rigid wall and linear elastic, were selected. The blood has been taken as laminar non-Newtonian (power law). Their results showed the effects of pathological conditions on pressure drop and they found that maximum wall shear stress occurs at branches region for healthy condition. For an ill person, the value of WSS is greater than healthy one because of tapered artery which influences on infected region growth.

Abraham et al. [2] discussed mechanic of blood in plaque-narrowed posterior tibial artery and the influence of removing plaque from tapered part whose geometries were found by ultrasound image. In their simulation they considered blood flow transient as the effect of its pulsatile nature. Both Newtonian and non-Newtonian (Ostwald–de Wael) rheology was used in this study. Their results illustrated that by removing plaque, blood flow increased 2.5 times during a cardiac cycle. They showed that shear stress on plaque removed artery wall was higher than narrowed one which means WSS is a favorable factor to avoid plaque coagulations on wall.

Blood flow is simulated as STT turbulent model in human aorta and its different branches by neglecting wall deformability by Benim et al. [3]. They investigated flow treatment as steady state and transient with Newtonian blood model. Their analysis mostly focused on the outlet boundary condition by defining a loss coefficient based-model introduced to it. They distinguished that pulsatile and steady state results differ significantly. They results showed the influence of accumulation of atherosclerotic plaque for extracorporeal circulation. The magnitude of the wall shear stress was greater for extracorporeal circulation in compression with physiologic circulation.

Prashanta Kumar Mandal[4] numerically studied non-Newtonian (power law) non-linear time variant laminar blood flow for stenosis arteries. Vessel wall structure deformability considered as non-linear visco-elasticity. A finite element approach is used to analyze the flow in two dimensions.

Wellnhofer et al. [5] investigated the effect of the coronary bifurcation on wall shear stress by CFD three dimensional simulating of blood flow in right coronary with aneurysmatic condition. They assessed non-Newtonian power law model of blood with laminar, steady and pulsatile consideration of flow in this study. Results illustrated artery branches affect wall shear stress because of reduction of the flow rate in main artery as branches are added to it. Also there is not a remarkable difference for wall shear in transient and steady model.

Bernsdorf and Wang[6] explored blood flow during aneurysm in cerebral artery. Non-Newtonian modeling was used to simulate the blood flow, and the best fit model to see the impact of the non-Newtonian flow in wall shear stress was Carreau-

Yasuda model. The result indicated the low viscosity and shear stress close to wall as the impact of non-Newtonian shear thinning in compression with Newtonian one.

Lantz et al. [7] preformed computer simulation of turbulent blood flow in descending aorta whose geometry was extracted from MRI data. Transient scheme was used according to time variant mass flow rate profile and turbulent model applied for simulation, was large-eddy. Blood was considered as having Newtonian property. Results described effects of pulsatile turbulent flow on wall shear stress. It was found that the magnitude of the WSS is high near the bifurcation and low shear stress on other part of the artery and both oscillating.

Nguyen et al. [8] investigated the impact of the geometry on blood flow behavior in carotid vessel and its branches. They explored blood to have constant viscosity (Newtonian) and transient laminar flow applied to investigation.

The effect of the geometry on blood flow characteristics was explored by Jozwik and Obidoswki[9] without considering vessel structure. Their case study was vertebral arteries to find out why there is flow property in right and left vertebral. Flow considered non-Newtonian power law model. Model was set as pulsatile according to variation of flow velocity characteristics. The results illustrated the velocity and flow rate as the function of the size and orientation of the arteries geometry.

Thubrikar et al. [10] discussed the mechanical property of aorta illness structure. They did uniaxial tensile test on regions of aneurysm part of abdominal aorta. They found aneurysm part behavior as orthotropic material and also non-linearity of the stress strain relation. They found that in longitudinal orientation the flimsiest part was the anterior. They explained an isotropic model according to average of data in different directions.

Blood flow behavior was studied in abdominal aorta by Viswanath et al. [11]. They assessed blood flow as pulsatile and Newtonian blood model. Wall structure was considered rigid in this study. They found the maximum magnitude of shear stress and pressure at the end of the distal. Also it was found that the amount of forces on wall changes as the result of pulsatile flow.

The comparison of the both Newtonian and non-Newtonian model of the blood with considering both steady and transient (pulsatile flow) state is performed by Liu et al. [12]. The case study was human aorta and the geometry was extracted from MRI data. The results indicated that non-Newtonian rheology and pulsation does not have remarkable impact on mass transport, except parts which have flow disturbance. But pulsatile blood regime can influence on oxygen flow in some region.

Paul et al. [13] analyzed blood flow in simple 3D channel whose characteristic was stenosis. They simulated the flow by Large-Eddy turbulent model which was characterized pulsatile. Blood considered as Newtonian incompressible. They recognized that the sub-grid model affect on flow energy dissipation. It was found that after tapered part flow has transformed to turbulent one and magnitude of the shear stresses and vortex are high which will be harmful at the posterior stenosis because of flow stagnation.

The impact of the blood turbulent flow on wall stresses is studied by Khanafer et al. [14]. Their study focused on aortic aneurysm considering the vessel wall impact. The flow was investigated as pulsatile turbulent model in study. They showed at the post-peak systole the maximum amount of the wall stress and deformation happens. The occurrence of the vortices at aneurysm part led to increase of pressure and shear stress and forced wall to vibrate.

Simulation of blood flow in abdominal aorta and its branches wasperformed by Lee and Chen [15]. This study was conducted for steady flow considering rigid vessel wall property and Newtonian blood model. A finite volume approach based on pressure was utilized to solve the problem. The results illustrated that the highest value of the wall shear stresses occurred close to branch curvature.  The scale of the reverse flow was greater in SMA branch in comparison with celiac trunk. They suggested that steady result can be considered as the average of the transient (pulsatile) flow.

Sokolis et al. [16] conducted uniaxial tensile test on three regions of aneurysm to find the vessel layer mechanical property variations. They results showed that the adventitia layer was the most resistant part to rapture. Also, it was illustrated that in circumferential direction adventitial and medial were stronger than longitudinal. According to their stress strain curve results, all layers showed orthotropic material property.

Wangand Bernsdorf performed [17] a numerical simulation on blood flow to compare non-Newtonian and Newtonian flows in 3D generic stenosis. They used a lattice Boltzmann simulation with Carreau-Yasuda non-Newtonian blood model. This comparison demonstrated that describing non-Newtonian blood rheology was necessary for solving the shear dependent viscosity for flow patterns.

To examine the post stenosis flow phenomena, numerical simulations were conducted on pulsatile blood flow in straight tube stenosis models by Long et al.[18]. They analyzed some flow characteristics like wall shear stress (WSS) distributions with high spatial and temporal resolutions for laminar flow. Results demonstrated that for asymmetrical cases, regions of wall shear stresses oscillation in the axial direction were larger than corresponding axisymmetrical model. For asymmetric models, the positive wall shear stress always peaked at the upper wall near the stenosis as a result of skewed jet flow.

Vignon-Clementel et al.[19] developed a coupled multi-domain method to simulate 3D finite element blood flow and specialization to resistance and impedance of boundary conditions. Then, they studied this approach for a subject-specific model of the human abdominal aorta in a straight cylindrical blood vessel (a bifurcation model with a stenosis on one side). This analysis demonstrated the need of proper boundary conditions in computing pulse pressures.

In order to find the mechanical behavior of thoracic aorta, Garcı´a-Herreraa et al. [20] did uniaxial tensile and pressurization tests on healthy descending aorta specimens of different ages. The tensile test data were utilized to find the parameters for Holzapfel model which showed proper characteristic for orthotropic structure behavior and elongation.

Mechanical characterization of healthy and aneurysmatic subject ascending aorta through biaxial tensile test for different regions of the vessel investigated by Choudhury et al. [21]. Results showed that inner surface of vessel is the most elastic part and the outer is stiffest. They investigated mechanical behavior of bicuspid and tricuspid vessel conditions. In addition the results showed bicuspid patients were at higher risk to vessel failure.

Healthy and patient circular ascending aorta specimens were used in tensile test experiment concerning to find their mechanical characteristics by Garcı´a-Herrera et al. [22]. They found that age affected the vessel structure and caused its weakness. No remarkable mechanical strength difference was seen between bicuspid and aneurysmatic aorta samples. Also, they investigated safety factor for average wall stresses on ill aorta specimen.

# Chapter 3

# BLOOD FLOW MODELING

As it was mentioned before, this study is about blood flow simulation in circulatory system and the case study is descending aorta. Aneurysm and atherosclerosis are the most important conditions in descending aorta which affect the blood flow. So, in order to investigate these cases there are some requirements that should be identified regarding cardiovascular circulatory system. These requirements can be as following:

1. Circulatory system
2. Circulatory subsystem
3. Mechanic of heart
4. Biological disease in cardiovascular system

## 3.1 Blood Cardiovascular Circulatory System

### 3.1.1 Circulatory System

Cardiovascular circulatory system consists of many parts including heart and blood vessels, which latter are subdivided into arteries, veins, capillaries and lymphatic. It means that circulatory system behaves like a network in which heart is the main part functioning like a pump (see Fig 3.1). The system task is to transport extracellular fluid and return it to heart again. The role of this circulation is to deliver oxygen and nutrients to cells and tissues and collect the carbon dioxide and other waste from them. Another function is to keep the body temperature in normal condition and transfer extra temperature to those parts of the body which can dissipate it into environment [23].

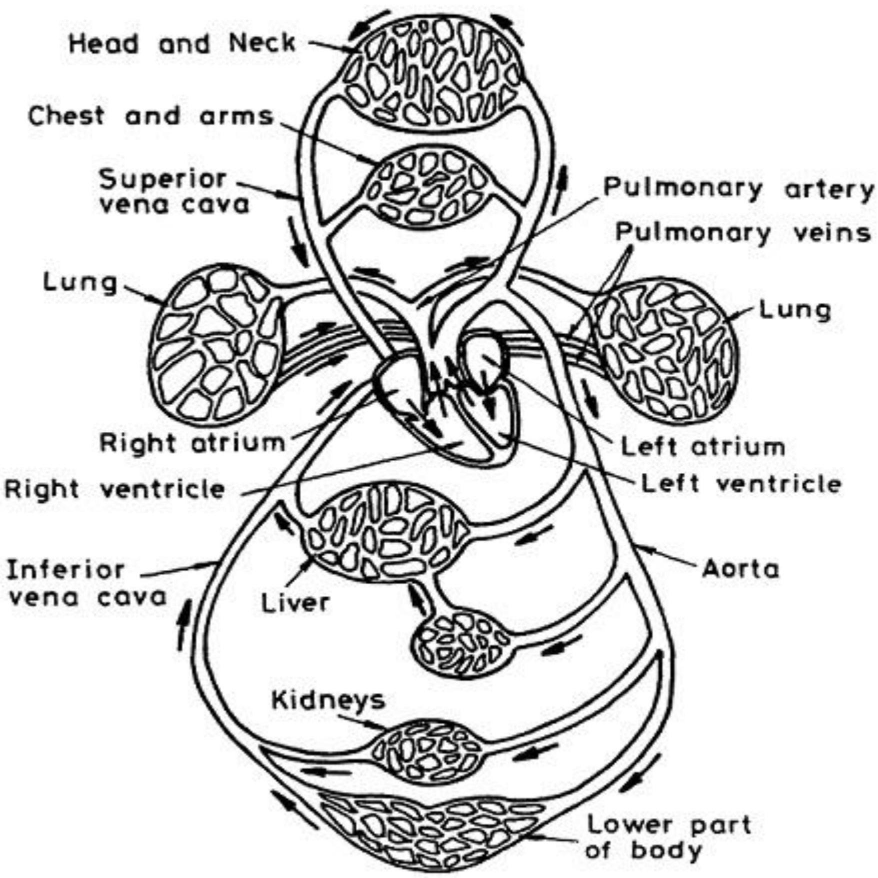


Figure 3.1. Schematic view of blood circulatory system[23]

### 3.1.2 Blood Circulatory Subsystems

Cardiovascular circulatory system is made up of three parts. The pulmonary and systemic circulation as well as vessels which are connected together to form the system [24] (Fig 3.2 and Fig 3.3).

As a low pressure system, pulmonary system supplies blood for lungs. But systemic circulation feeds aorta to pump blood to all parts of the body (except lungs): a high pressure system because blood should be pumped to body parts which located far away from heart, such as brain to which blood should even to be pumped against gravity force [23, 24].

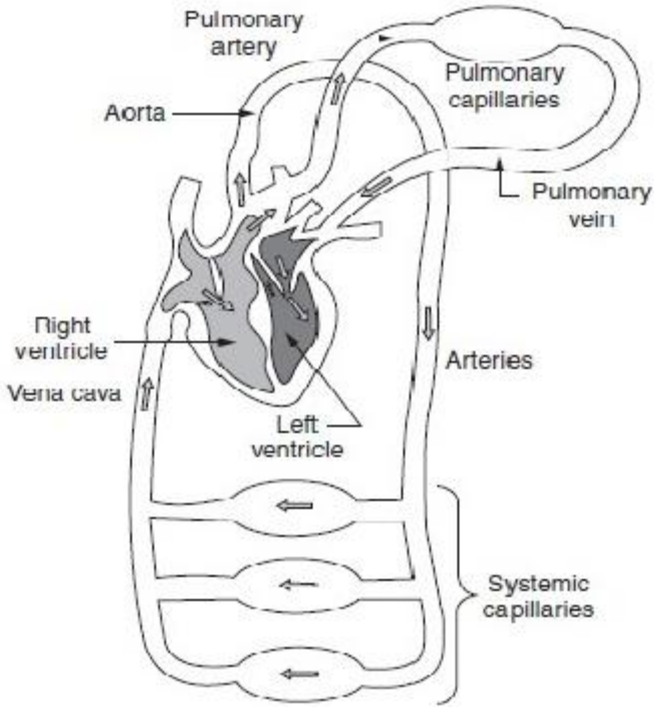


Figure 3.2. Full schematic view of blood circulatory subsystem

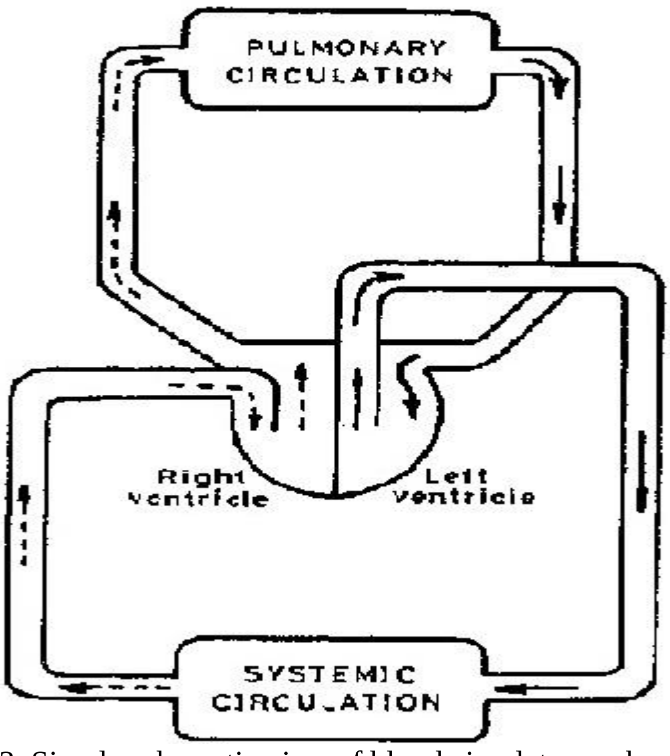


Figure 3.3. Simple schematic view of blood circulatory subsystem [23]

### 3.1.3 Mechanic of Heart

Heart is a pump which exerts force to move the blood to be circulates through cardiovascular system. Heart is made up of 4 chamber that can be classified to atria (two upper chambers), ventricles (two lower chambers), in other word they can be categories to right atrium, right ventricle, left atrium and left ventricle. First blood goes through the right atrium from vena cava, and then it is pumped through the right ventricle and from there to it is pumped to through the pulmonary artery which goes to lung to rich the oxygen from it and releases carbon dioxide. After that in the other side of the heart (left side) the oxygenated blood enters left atrium from pulmonary veins and then it is pumped to left ventricle and by then with a high pressure to aorta [24] (see Fig 3.2).

## 3.2 Circulatory System Pathologies

### 3.2.1 Atherosclerosis

This is a common disease occurring in cardiovascular circulatory system where the blood arteries become narrow especially at vessel branches or in places where vessel geometry is curved which might results in accumulation of fatty substances, cholesterol, fibrin and etc which is known as plaque[2, 12, 15]. It causes the vessel wall to become rigid gradually and can be led to vessel blockage and heart attack (stroke) by excessive plaque accumulation. This phenomenon can also lead to blood coagulation that might result in a stroke. Furthermore, it is likely sometimes that clot being cut off from vessel wall and moves through small arteries barring them which again can result in stroke [25](see Fig 3.4).

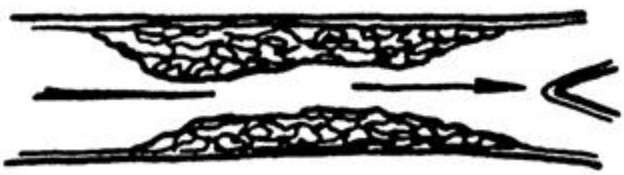

Figure 3.4. Schematic view of arthrosclerosis pathology [23]

### 3.2.2 Aneurysm

Aneurysm is one of the cardiovascular pathological condition which mostly appear in abdominal aorta (abdominal aortic aneurysm) and brain (cerebral aneurysm) and it does not have any symptoms[25]. It is vessel wall elongation more 50% of normal diameter and blood flow will decrease (see Fig 3.5 and Fig 3.6). Without curing and surgery the diameter will increase dramatically and lead to vessel rupture which in this case approximately 60% of suffering persons die before arriving hospital [10, 16].

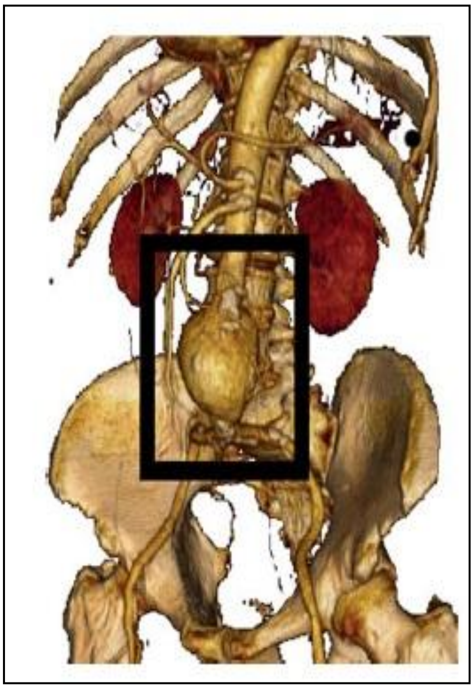

Figure 3.5. Scan image of aneurysm[26]

Figure 3.6. Scan image of aneurysm

## 3.3 Descending Aorta

Aorta is a part of systemic circulatory system (high pressure) which feeds all part of body. As aorta comes out from the heart it is divided into ascending and descending parts. Descending part is the case of study whose diameter decreases as it comes down through the abdomen (stomach) and legs. Many branches spring out of it; the most significant ones are left renal, right renal, left iliac, right iliac, inferior mesenteric, superior mesenteric and celiac trunk (Fig 3.7). The first four branches are considered for this study.

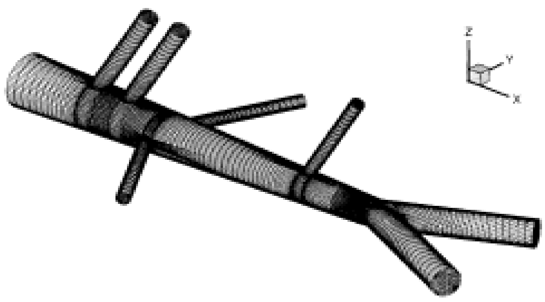

Figure 3.7. 3-D Schematic drawing of aorta [15]

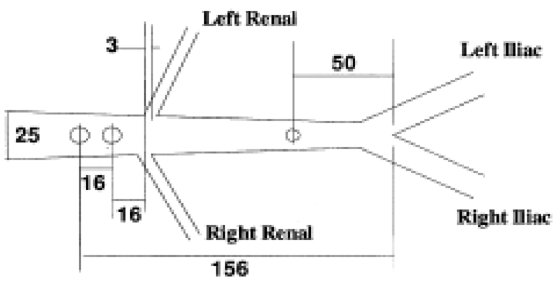


Figure 3.8. 2-D Schematic drawing of aorta [15].

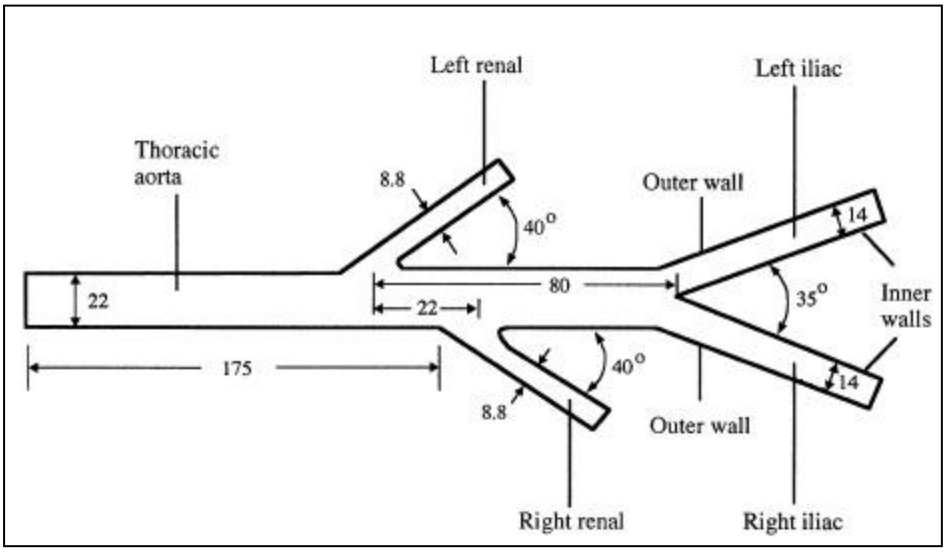

Figure 3.9. 2D Diagram of aorta with branches with dimension [27]

As it can be seen from Fig 3.9, thoracic aorta is not descending and also there are six branches in Fig 3.8. The geometry data are extracted from Fig 3.8 (descending part) and Fig 3.9 (four main branches) and mixed together which is shown in Fig 3.10.

The simulation is performed under both normal and abnormal conditions to find differences and effects of sickness on flow and structure parameters. So, another geometry is constructed to which aneurysm and atherosclerosis diseases are added. One of them is added to left and another to the right iliac branch (see Fig 3.11). The geometry dimension is completely same as Fig 3.9 except injured parts that their diameter and dimension are collected from [10], and also as explained before, vessel diameter increases and decreases 50% in aneurysm and atherosclerosis.

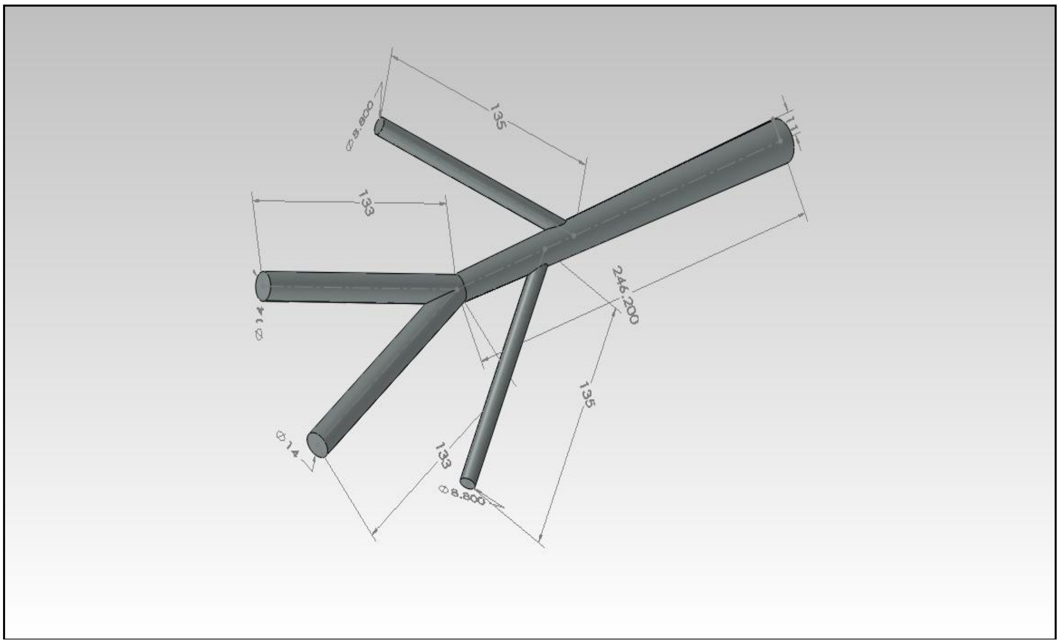


Figure 3.10. Sketch of healthy aorta

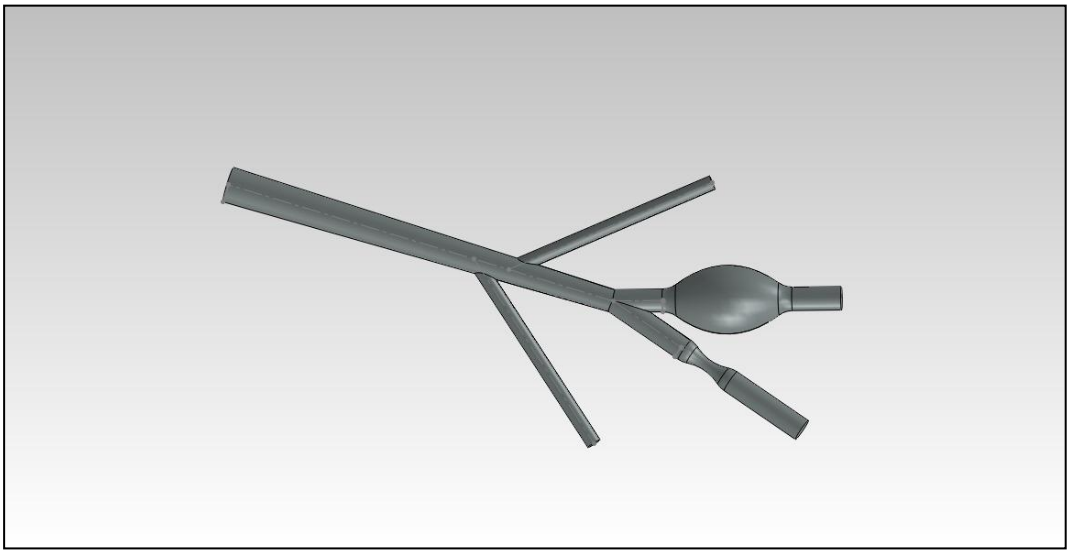

Figure 3.11. Sketch of ill (aneurysm in left and atherosclerosis in right) aorta

## 3.4 Fluid Solid Interaction (FSI)

Fluid Solid Interaction or FSI is a situation in which fluid and solid influence each other. The purpose of this study is to simulate both solid and fluid phases, so FSI can meet both of these requirements because fluid impacts on vessel wall by its pressure which is pulsatile, and the structure is deformed elastically and comes back to its normal condition. But as the solid part goes back to its primary condition it influences fluid as well.

In order to simulate the condition described, a wall of 2mm thickness [22] is considered for the whole body which its geometry mentioned before to be like a shell (Fig 3.12). Elastic and mechanical properties are allocated to this tube which will be explained later.

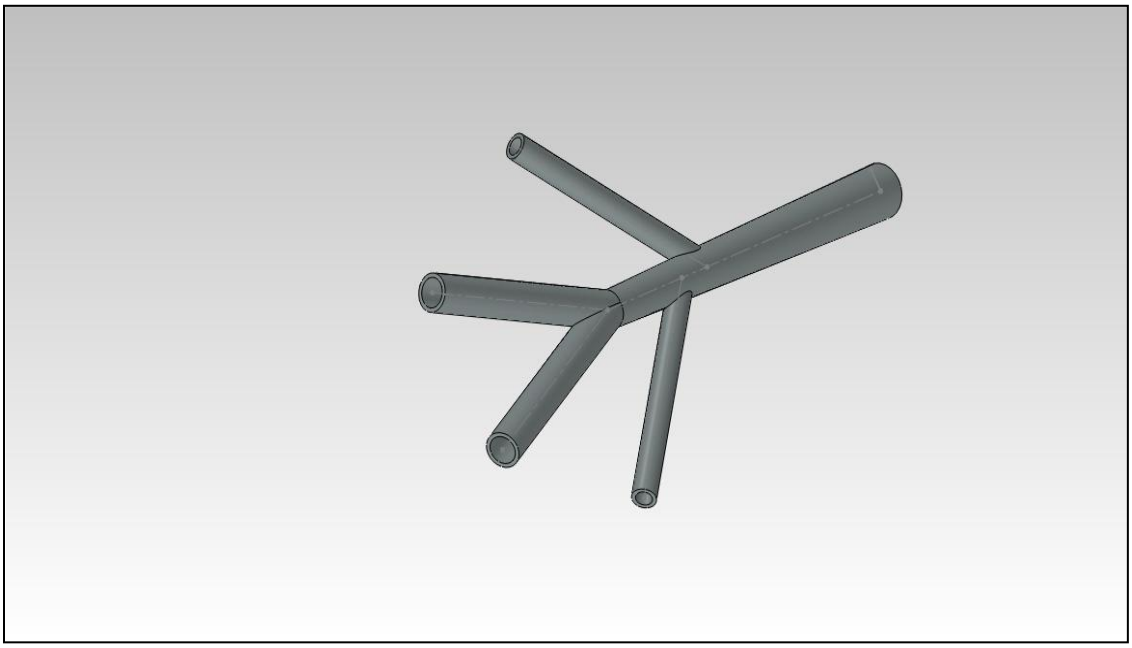

Figure 3.12. Sketch of aorta wall

## 3.5 Vessel Structure

Vessel wall structure is composed of a complicated material. It includes three main layers: tunica externa, tunica media and tunica intima [25]. Each of these layers plays an important role in wall structure. Vessel structure generally has an elastic property that can be elongated by the pressure applied to it which means it can go back to its primary diameter. Its elastic property can be assumed as an isotropic linear material. Linear material is one whose stress strain relation in elastic part is linear. Isotropic elasticity is a mechanical property that is identical in different directions. But in reality, authors identify vessel structure as having non-linear orthotropic elasticity. Orthotropic material is one whose its mechanical properties are not identical in different directions.

The vessel density applied is $1050\mathrm{kg/m^3}$. Mechanical properties such Poisson’s ration and Young modules are set 0.45 and 2.7Mpa [14]. In order to define the most exact behavior of structure, uniaxial tensile test data on aneurysmatic part of abdominal aorta specimen is extracted from Fig 3.13 and defined as a material property.

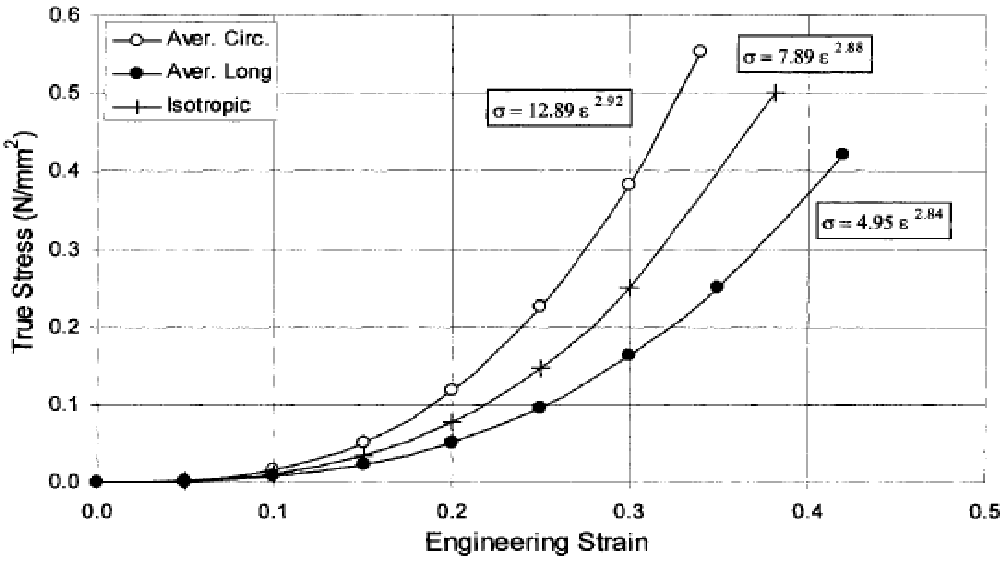


Figure 3.13. Uniaxial tensile test data on abdominal aorta [10]

Although in Fig 3.13 uniaxial tensile test data are illustrated as non-linear stress strain curve in two directions (orthotropic), circumferential and longitudinal, also there is a curve for isotropic property which is the mean of two other curves. Thus, the isotropic curve is used for material definition. So the mechanical properties which is used for this study is assumed to be non-linear isotropic elastic.

It was explained, Fig 3.13 is related to aneurysmatic part of the vessel which obviously has lower stiffness that other parts. Although this is the best material fit for aneurysmatic part, for other parts of the vessel it was assumed they have this mechanical property as well. This material characteristic is satisfied structure weakness according to aneurysm and atherosclerosis property definition.

## 3.6 Meshing

For each of geometries used for the simulation a tetrahedral computation mesh elements are conducted. As mesh quality will affect solution accuracy, some inflated boundaries are introduced to the wall of the meshed fluid geometries which leads to prism elements in proximity of the wall in order to solve flow parameter more accurately. Number of the prism element layers is set to be 12 with a 1.2 growth factor meaning the first prism element size near the wall will be multiplied by growth factor in order to produce subsequence prism element. So, prisms size increases sequence up to 12 layers.

## 3.7 Blood

It is known that Newtonian fluid is one that its viscosity at any shear rates is constant. Homogeneous fluid mostly behaves as Newtonian. So, the fluid in which the relation between shear stress rates and shear strain rates is non-linear is called non-Newtonian fluid. There are non-Newtonian fluid categories such as power law, Bingham fluid, Casson fluid, prandtl fluid, Carreau_Yasuda, etc. [24].

In addition, blood is not a homogeneous fluid and also it shows two types of abnormality regarding shear stress and shear strain rates [23]. At high shear rates it treats as Newtonian fluid but at low shear rate it exhibits non-Newtonian property.

The size of the vessel also affects this phenomenon in which the viscosity is low in small tube in comparison with large tube.

For this study blood is considered as non-Newtonian fluid. Carreau_Yasuda is the best fit model for the blood properties and it does not have disadvantages of other models. Power law model is weak to show non-Newtonian property of blood because it has gradient and infinite viscosity and Casson is invalid for high range of shear [6].

$$\frac{\mu-\mu_\infty}{\mu_0-\mu_\infty} = (1+(\lambda\gamma^{\cdot})^{\alpha})^{(n-1)/\alpha} \qquad \textbf{(3.1)}$$

Which λ = 0.110s is the time constant as the (characteristic viscoelastic time of the fluid), n = 0.392 is power law index, α = 0.644 is Yasuda exponent, $\mu_0$ = 0.022 Pa s and $\mu_\infty$= 0.0022 Pa s are low and high shear viscosity.

## 3.8 Boundary Conditions

To solve the problem three boundary conditions are applied to fluid domain, inlet, outlet and wall. In reality blood has pulsatile nature whose velocity, pressure and all other properties related to these parameters including mass flow rate, change by time as blood comes out of heart during each heartbeat.

### 3.8.1 Inlet Boundary Conditions

An inlet boundary condition is put on fluid domain. A time dependent mass flow rate pulse is introduced to this boundary in which the value becomes negative in some parts (Fig 3.14). It is expected that results can show the reverse flow direction.

### 3.8.2 Outlet Boundary Conditions

For the outlet boundary condition also pressure pulse is applied to the domain. Some other characteristics are applied for this boundary that will be explained later in computer modeling. In Fig 3.14, it is shown that these two pulses are coupled together meaning during the systole as the mass flow rate increases, pressure is amplified as well.

### 3.8.3 Wall Boundary Conditions

No slip wall boundary condition is applied to domain wall in order to have proper velocity gradient near the wall.

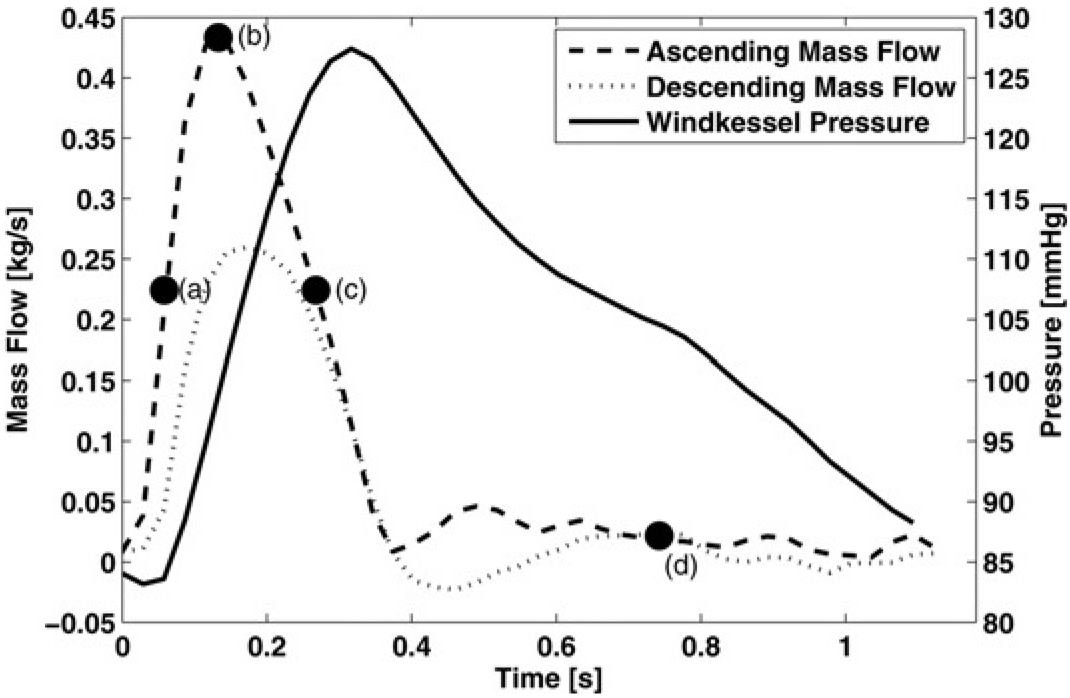

Figure 3.14. Time variant velocity and pressure profile [7]

## 3.9 Numerical Method

Momentum and continuity equations are conserved in each element of meshed domain. One cardiac cycle of heart beat which lasts one second is chosen as the total time step. To solve equation at each step, time step period is divided to 50 sub-steps, with each step lasts for 0.01 second. A transient scheme is considered for whole cardiac cycle to converge the equation. Second-order backward Euler is applied as the transient discretization scheme. To evaluate the advection term in momentum and continuity equations, high resolution transient scheme is used which in turn uses the second order backward Euler scheme wherever and whenever possible and reverts to the first order backward Euler scheme when required to maintain a bounded solution.

For convergence of the solution, 20 iterations are set to all sub-steps. A maximum root-mean square (RMS) residual of $10^{-6}$ is introduced to momentum (Navier-Stokes) and continuity and the convergence target is assigned 0.05 [2] in order to calculate the convergence domain imbalances below this value.

According to what is said so far, the simulation is conducted as compression case to investigate the blood behavior in thoracic aorta and its branches during healthy and illness (aneurysm and atherosclerosis) considering without and with non-linear elastic vessel wall. By applying laminar flow according to the mass flow rate pulse (as inlet boundary condition) and pressure pulse (as outlet boundary condition) which are coupled together, transient property is used for investigation. Carreau_Yasuda non-Newtonian rheology is chosen as the blood properties to find out its effect as the shear stress on the wall boundary and to increase the simulation accuracy.

## 3.10 Mathematic Background

As it is considered before, this study simulates blood flow in two cases, healthy and illness conditions, that for each of them a single 3-D geometry is designed. A three dimensional, laminar, unsteady incompressible (constant density) flow is utilized for analysis. The fundamental equations which are employed according to the conditions are: three dimensional Navier-Stokes (momentum) and conservation of mass (continuity).

The continuity equation is:

$$\frac{\partial u}{\partial x}+\frac{\partial v}{\partial y}+\frac{\partial w}{\partial z}=0 \qquad \textbf{(3.2)}$$

Three Navier-Stokes equations in three directions are:

$$\rho\left(\frac{\partial u}{\partial t}+\frac{\partial}{\partial x}(u^2)+\frac{\partial}{\partial y}(uv)+\frac{\partial}{\partial z}(uw)\right)= -\frac{\partial p}{\partial x}+\frac{\partial \tau_{xx}}{\partial x}+\frac{\partial \tau_{yx}}{\partial y}+\frac{\partial \tau_{zx}}{\partial z} \qquad \textbf{(3.3)}$$

$$\rho\left(\frac{\partial v}{\partial t}+\frac{\partial}{\partial x}(uv)+\frac{\partial}{\partial y}(v^2)+\frac{\partial}{\partial z}(uw)\right)= -\frac{\partial p}{\partial y}+\frac{\partial \tau_{xy}}{\partial x}+\frac{\partial \tau_{yy}}{\partial y}+\frac{\partial \tau_{zy}}{\partial z} \qquad \textbf{(3.4)}$$

$$\rho\left(\frac{\partial w}{\partial t}+\frac{\partial}{\partial x}(uw)+\frac{\partial}{\partial y}(uw)+\frac{\partial}{\partial z}(w^2)\right)= -\frac{\partial p}{\partial z}+\frac{\partial \tau_{xz}}{\partial x}+\frac{\partial \tau_{yz}}{\partial y}+\frac{\partial \tau_{zz}}{\partial z} \qquad \textbf{(3.5)}$$

*P* term and its gradients are crossed off from continuity equation because of the constant density of the fluid (incompressible),other than the continuity equation is:

$$\frac{\partial \rho}{\partial t}+u\frac{\partial \rho}{\partial x}+v\frac{\partial \rho}{\partial y}+w\frac{\partial \rho}{\partial z}+\rho\left[\frac{\partial u}{\partial x}+\frac{\partial v}{\partial y}+\frac{\partial w}{\partial z}\right]=0 \qquad \textbf{(3.6)}$$

These equations are in Cartesian coordinates whose three directions are X, Y and Z. Corresponding velocity components are $u$, $v$ and $w$and $\tau$ is the shear stress.

According to the consideration of non-Newtonian rheology of the fluid, shear stresses gradients which are described in Navier-Stokes equation can be evaluated by using Carreau_Yasuda model formula (non-Newtonian model which is utilized).

For the effect of the fluid and solid on each other, the vessel wall structure is considered as non-linear isotropic material whose stress and strain relation satisfies:

$$\sigma = 7.89.\varepsilon^{2.88} \quad \textbf{(3.7)}$$

This formula is achieved experimentally by uniaxial tensile test as the average value of stress-strain curve in two circumferential and longitudinal directions. σ describes the engineering stress and ε explain the amount of the strain in relation to σ.

## 3.11 Computer Modeling

In order to simulate this case study, ANSYS software is employed as the most appropriate software to analyze fluid dynamic aspects of blood and vessel structure.

According to FSI condition, there are two phases for this study, solid and fluid. To simulate each part properly two analysis components of software are selected: transient structural for solid part and CFX for fluid part. In order to satisfy interaction between solid and fluid phases, a connection is applied between two systems (Fig 3.15).

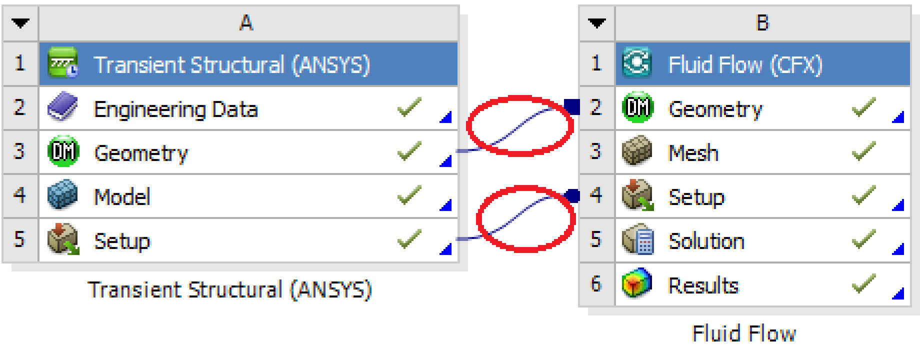


Figure 3.15. Scheme of two coupled system in ANSYS

As it is illustrated in figure above, the geometries and setup of two systems are connected together. It means that geometries contractions are well defined for software. Setups connection explains coupled solving of two systems which means at first the fluid system is going to be solved and then the results from fluid will be put as pressure or force on structure, then structure is going to be solved. This procedure is going to be repeated at each time step up to end of the solution as two way fluid solid interactions.

### 3.11.1 Solid System

Transient structural analysis component which is chosen for solid part is a finite element solver base system. It includes many parts that should be setup for simulation, engineering data (material property), geometry, meshing and setup. The procedure for simulating an analysis in transient structural can be classified to:

- Defining material property
- Defining geometry
- Meshing
- Applying support
- Applying load
- Solving

In engineering data which is related to material property, vessel structure material is created. Material properties which are added to the new material were density, Poisson's ratio and Young module whose values are reported before. Uniaxial tensile test data are added also as material behavior by extracting point's position from isotropic curve Fig.14 and importing them into software to calculate the best fit curve for material experimental tensile stress and strain value for abdominal aorta specimen. Extracted data are listed below in Table 3.1.

Table 3.1. Extracted point data from isotropic curve of tensile test

| **Stress($N/m^2$)** | **Strain** |
|---|---|
| 3820 | 0.0974 |
| 6980 | 0.125 |
| 18900 | 0.154 |
| 27800 | 0.183 |
| 45500 | 0.214 |
| 71900 | 0.241 |
| 110000 | 0.266 |
| 145000 | 0.288 |
| 183000 | 0.315 |
| 221000 | 0.334 |
| 256000 | 0.35 |
| 308000 | 0.365 |
| 349000 | 0.38 |
| 395000 | 0.392 |
| 424000 | 0.4 |

The geometry for both solid and fluid parts is designed by SOLID WORKS software as an assembly part according to data which are explained in system modeling. This assembly is included two parts which are related together fluid and solid regions. Then it was imported as STP File to ANSYS design modeler.

The geometries are meshed in ANSYS meshing platform (ICEM) which their elements details are described before.

For solid part simulation a fix support is applied to inlet region of the structure geometry, in order to fix it not to move when the fluid passes through. A fluid solid interface load is introduced to whole inner regions of the shell (vessel wall structure) as the region where the fluid influences solid to define the fluid solid interaction condition for the software.

To perform analysis, total time duration for solving the problem is set one second with 0.02 second sub-step. The total number of the sub-steps was 50.

### 3.11.2 Fluid System

Fluid phase is solved by using CFX platform. CFX is finite volume base solver which has the ability for analyzing fluid dynamic, heat transfer, combustion, etc., and predicting various conditions such turbulent and vortices

For setting up the analysis in CFX platform the procedure can be categorized to:

- Defining geometry
- Meshing
- Fluid definition
- Domain definition and its properties
- Boundary conditions
- Initialization
- Solver preparation
- Solution

Because the geometry is imported in solid part (transient structural) as assembly which is included two parts itself, it does not need to import or generate geometry for fluid domain because according to the Fig.16 geometries of two systems are connected to each other. The procedure for the meshing is exactly like the solid part.

In CFX analysis component a new fluid as blood is created because there is not any fluid in CFX library which is match with blood properties. According to the transport dynamic property of the fluid (non-Newtonian Carreau_Yasuda), all values calculated for blood (section 3.7) are imported to software.

A fluid domain is created in relation with geometry designed for fluid. Laminar and transient properties are allocated to the domain and blood.

Corresponding to the explanation of the boundary condition in system modeling, mass flow rate and pressure curves data are extracted from Fig.15 for inlet and outlet boundary conditions as points' position. Then, these point positions data are imported in MATLAB software in order to find the best polynomial equations for curves which fit these data. At last 10 and 6 degree polynomial equations are found for mass flow rate and pressure pulse. These equations are:

$$\dot{\mathrm{m}} = [-119.7\,\mathrm{t}^{10} + 587.21\,\mathrm{t}^{9} - 909.806\,\mathrm{t}^{8} - 105.997\,\mathrm{t}^{7} + 2036.37\,\mathrm{t}^{6} - 2738.58\,\mathrm{t}^{5} + 1728.16\,\mathrm{t}^{4} - 551.41\,\mathrm{t}^{3} + 75.604\,\mathrm{t}^{2} - 1.971\,\mathrm{t}]\,[\mathrm{kg/s}] \qquad (\mathbf{3.8})$$

$$p = [4806.7\,\mathrm{t}^{6} - 16976\,\mathrm{t}^{5} + 22580\mathrm{t}^{4} - 13711\mathrm{t}^{3} + 3456.8\mathrm{t}^{2} - 148.54\,\mathrm{t} + 83.853][mmHg] \qquad \textbf{(3.9)}$$

An extra parameter which named fn $[s^{-1}]$, is introduced to each equations to make t parameter dimensionless. Then, an inlet boundary condition is created at the beginning of the thoracic aorta. The mass flow rate formula is introduced as an expression into inlet boundary condition which its direction is normal to the inlet.

Outlet boundary condition is created at end of the cylindrical sections of left renal, right renal, left and right iliac. Pressure equation is applied to the boundary condition and the property of the boundary is set to be opening. Opening condition is used for outlet boundary whenever and wherever there is a possibility for reverse flow direction because software does not let normal outlet boundary to have reverse flow. According to Fig 3.15 since there is reverse flow direction, so opening is an appropriate condition to be applied for boundary.

A wall boundary condition is applied to whole outer surface of the fluid geometry whose property was no slip wall where the velocity profile gradient is zero. For each boundary condition except wall, mesh motion property is set unspecified for the simulation. It means that boundary domain deformation and alteration are related to other boundaries. For wall boundary condition, mesh motion property is set to be related to motion of wall structure, or in other words, related to the total mesh displacement of vessel structure.

For analyzing a transient condition software needs an initialization value for some parameter. Three components of velocity profile (u, v and w) are set to zero and also for average static pressure.

# Chapter 4

# RESULTS AND DISCUSSIONS

This section presents the results for the 3D pulsatile blood flow simulation in thoracic descanting aorta which is solved by ANSYS® commercial software, version 12. In addition, results are extracted for both healthy and ill subjects as well as geometry effect, and also for the cases which include elastic wall assumptions as FSI condition regarding one second duration of the blood cardiac cycle. The results are presented by using ANSYS CFD-POST interface in categories such as velocity profile, pressure, wall shear stresses (WSS), Von-Misses stress, etc.

## 4.1 Descending Aorta without Branches

The simulation for this case is performed in order to show the effect of the branches as they are springing out of descending part, on the flow parameters for next cases. The geometry of descending part (thoracic trunk) was exactly like other models, except it does not have renal and iliac branches and it was the simplest result as a foundation for comparison. The velocity streamlines showed that the flow is approximately laminar except in some sub steps that became vortex when the pulse goes to its negative value which was expected to have reversal flow. So because the magnitude of vortices were not to much they can be neglected and assume flow as laminar. The velocity contour shows that as pulse passes through the end of the thoracic part, the velocity magnitude increases because of the diameter decrease, and the velocity core is formed at the end of the thoracic part. Velocity core goes back in reverse direction because of the negative value of the velocity in some time steps, that is reversal flow, this result is validated with [15] (Fig 4.1)

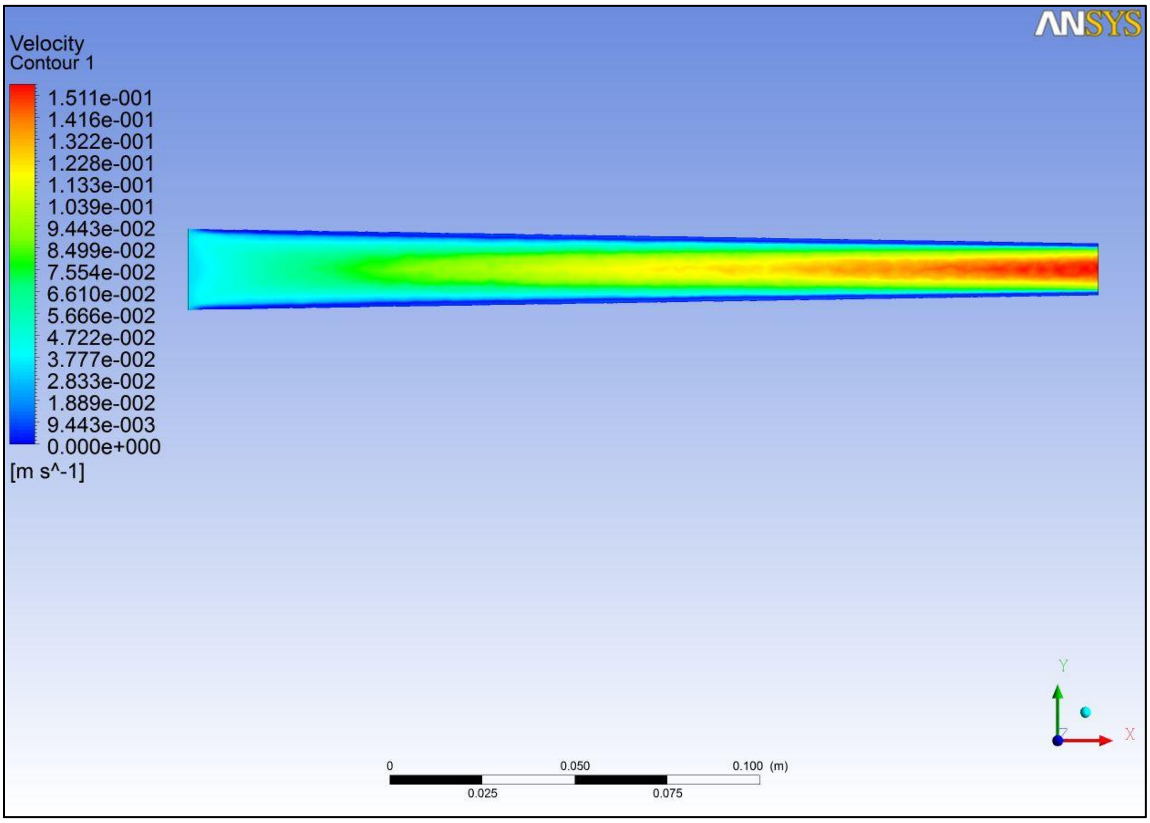


Figure 4.1. Sample of velocity contour for descending part

Also during the solution, solver draws the velocity pulse for both beginning and the end of descending part as the result of the time variant mass flow rate profile introduced at inlet boundary condition. It showed that the pulse also grows up as the result of blood passing through the thoracic trunk (Fig 4.2). Pressure drops throughout the vessel and the pulse shows this decrease as well.

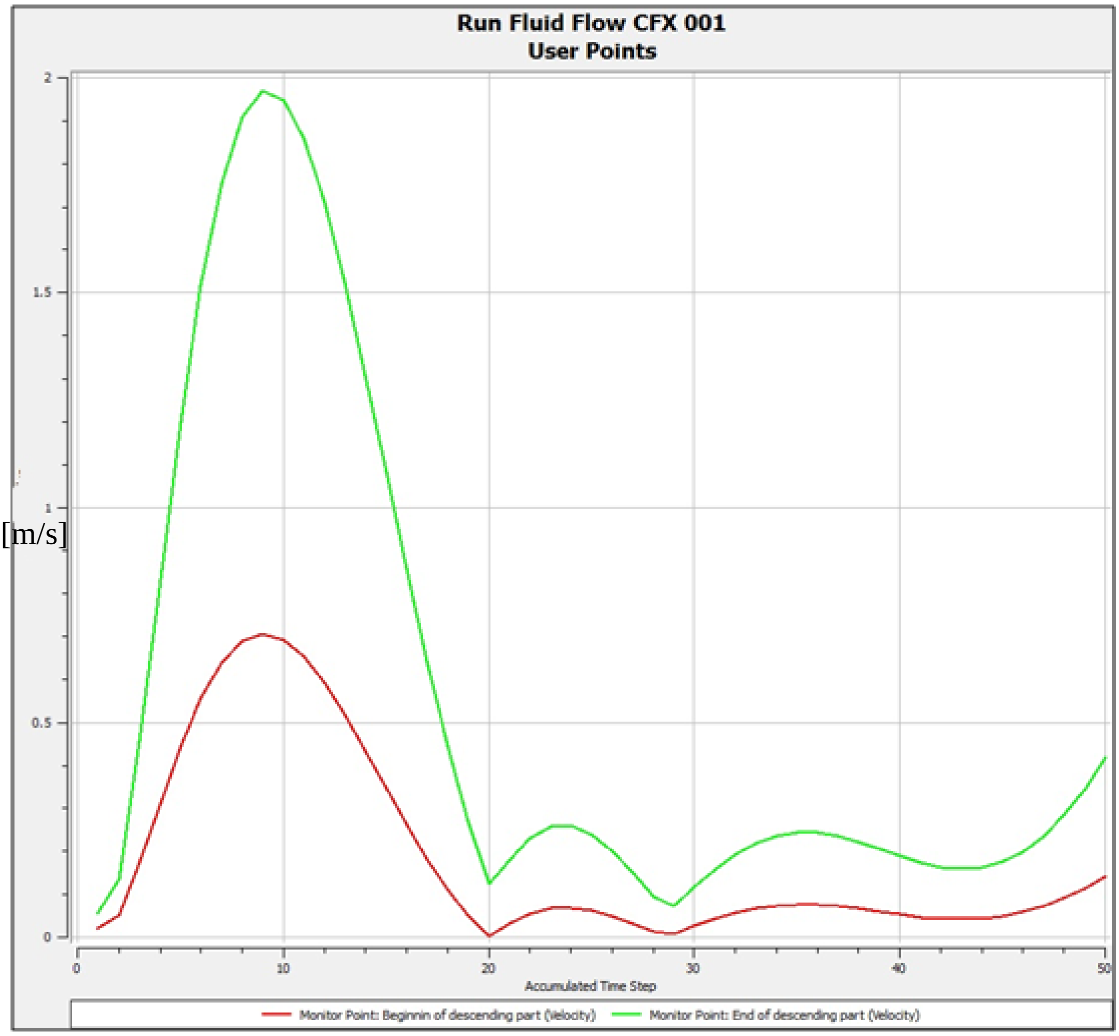

Figure 4.2. Time variant velocity pulse for region

## 4.2 Vessel with Branches

This case is analyzed in order to find the effect of the branches on flow behavior and its parameters. In this case right and left renal and iliac branches have been added to descending aorta. The result illustrated that branches cause flow deviation in some parts and these results are in coherence with [5] and [9]. It can be seen in thoracic aorta without branches that the flow streamlines were approximately laminar but as the bifurcations were added to vessel, vortices flow observed in some places. The magnitude and size of the vortices were various in different parts as the result of pulsation flow. The onset of the vortex flow was at 0.225 second (as the velocity pulse passed its peak) after curvature of the iliac branches. Vortices started to grow in renal branches at 0.275 s. For the region between renal branches in thoracic aorta at 0.375 s, when the time variant velocity trends its negative value and is expected to have reversal flow (Fig 4.3).

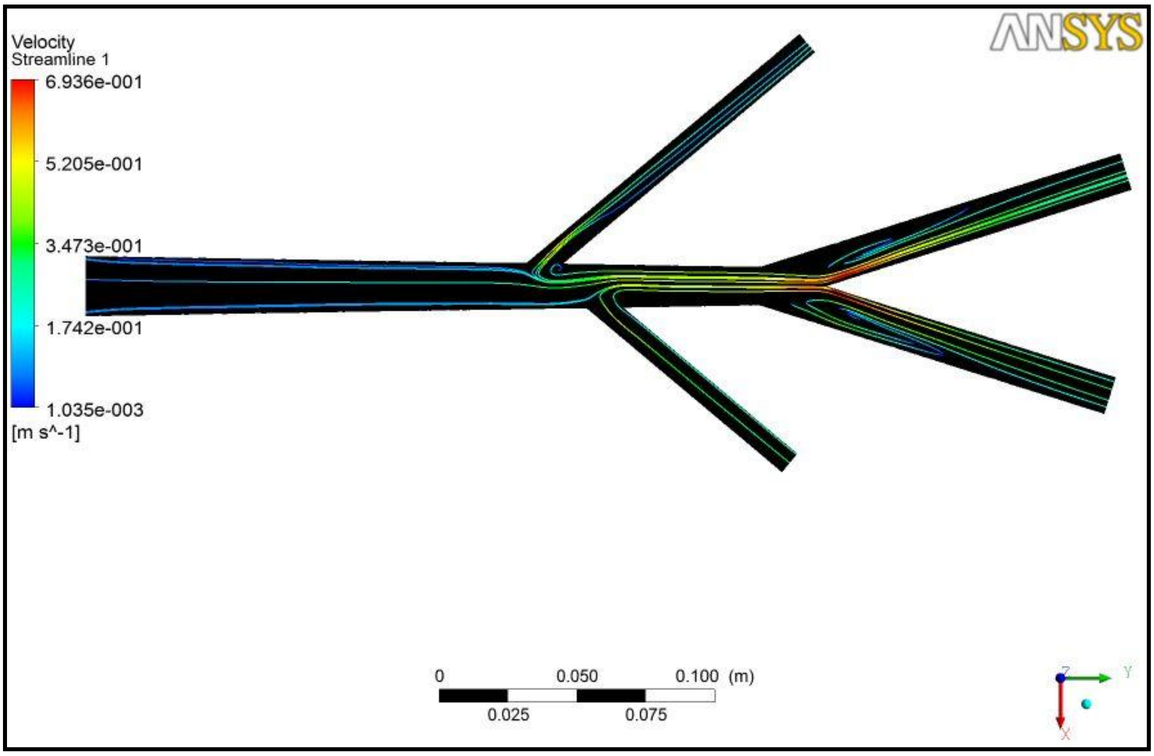

Figure 4.3. Sample of streamline contour

Highest size of the vortex flow was observed at 0.575 s then from 0.625 it showed downward trend to last time step when stream line became laminar totally. The velocity pulses were taken for beginning, middle and end of the main branches and it showed velocity increases as the diameter decreases except in some parts of the pulse (peak) and at the 0.325 flow reversal is detected in renal branches according to the inlet velocity pulse and this result showed the same trend as obtained results from [15] (Fig 4.4).

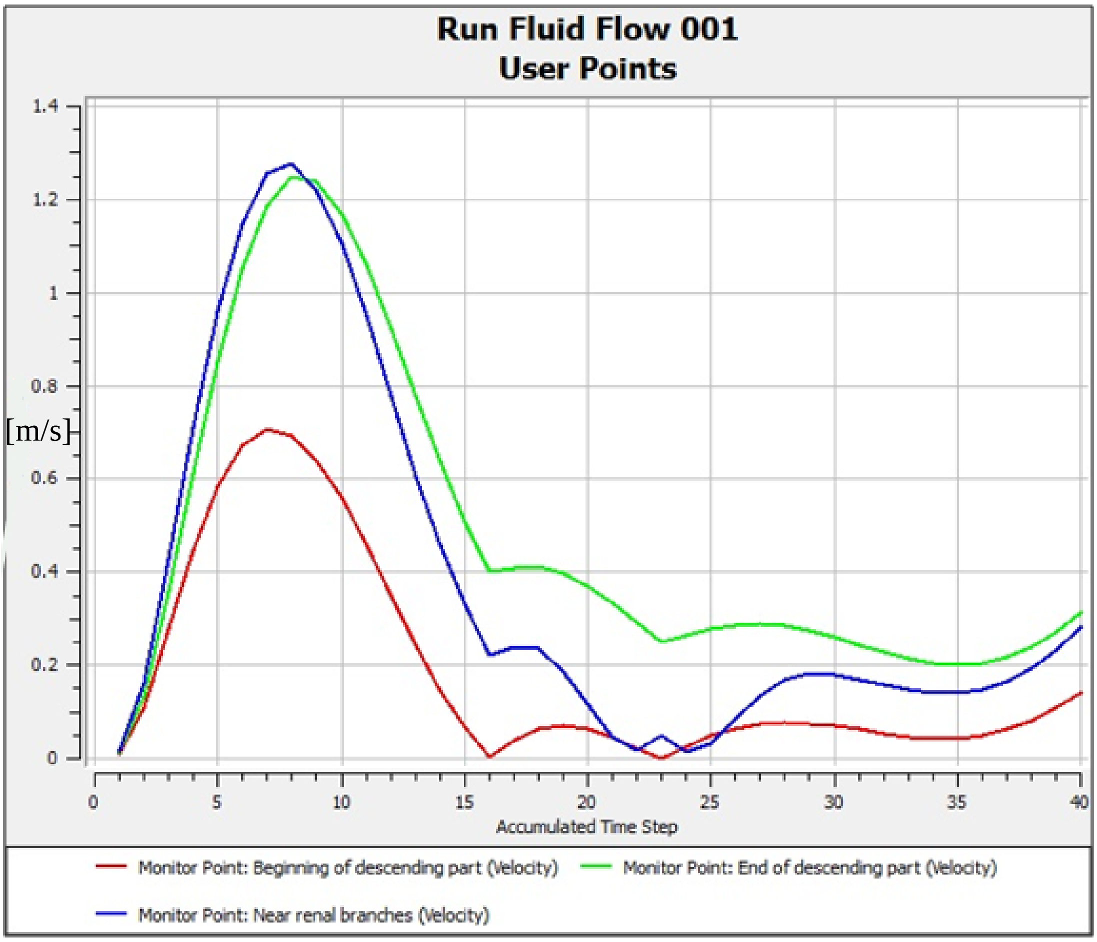


Figure 4.4. Time variant velocity pulse for region

About wall shear stress (WSS), results showed complexity as the influence of the branches, vortices and pulsation flow which was oscillated as well and it was validated with [7]. The maximum value of the WSS is observed near the renal and branches curvature as they are bifurcated from the thoracic trunk, in many time steps as in [1], [15] and [7] and also between the iliac bifurcations curvature as validated with [11] . But at interior face of the branches from beginning up to some length, a low magnitude of the WSS was observed which can be considered as a factor for plaques coagulation leading to vessel structure illnesses and flow deviation [7]. Also the highest value of wall shear stresses for different time steps was observed near the peak of pulse although it oscillated in different steps [14]. So WSS can be considered as vital parameter for plaques removal as validated in [2]. Also as it is explained before flow vortices at inner face of the branches cause the blood being inert in the vortices place and circulated which in turn leads to flow stagnation itself. Result for wall shear stresses in one of the most important step is shown in Fig 4.5.

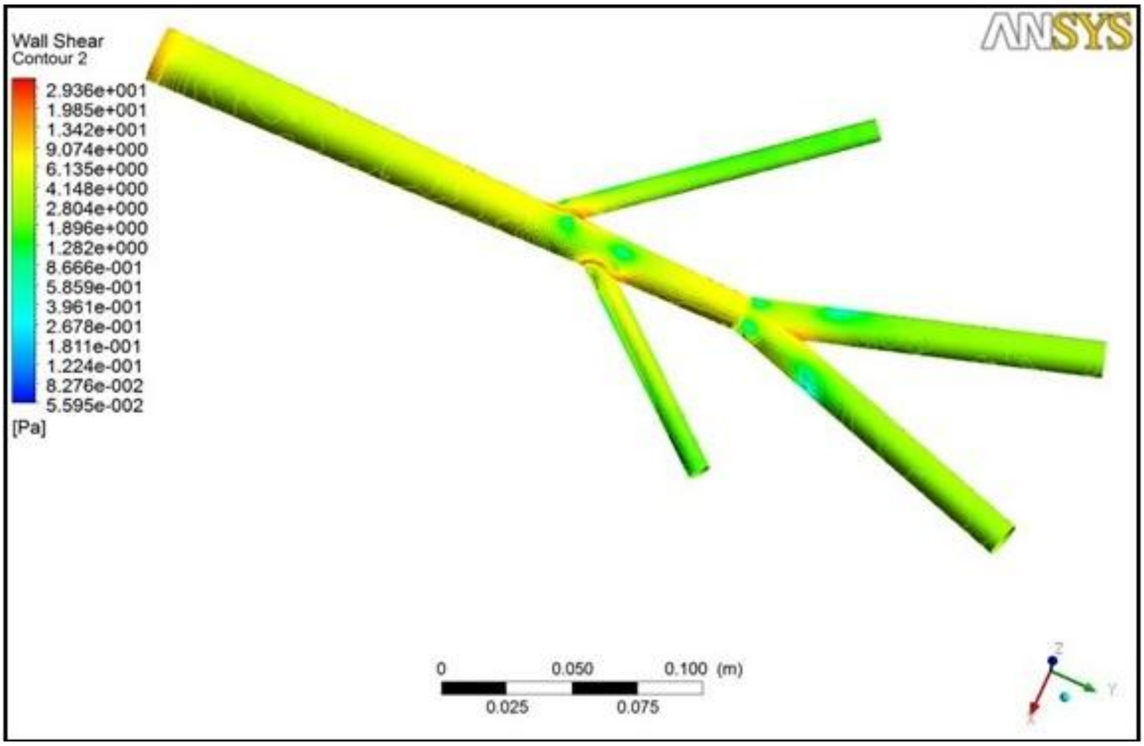


Figure 4.5. Sample of WSS contour

According to velocity and pressure counters, the magnitude of the velocity for the region where the wall shear stresses and vortices flow occurred was lower than the other parts, particularly lower than the opposite side of the faces where the regions for observing these behaviors (Fig 4.6).

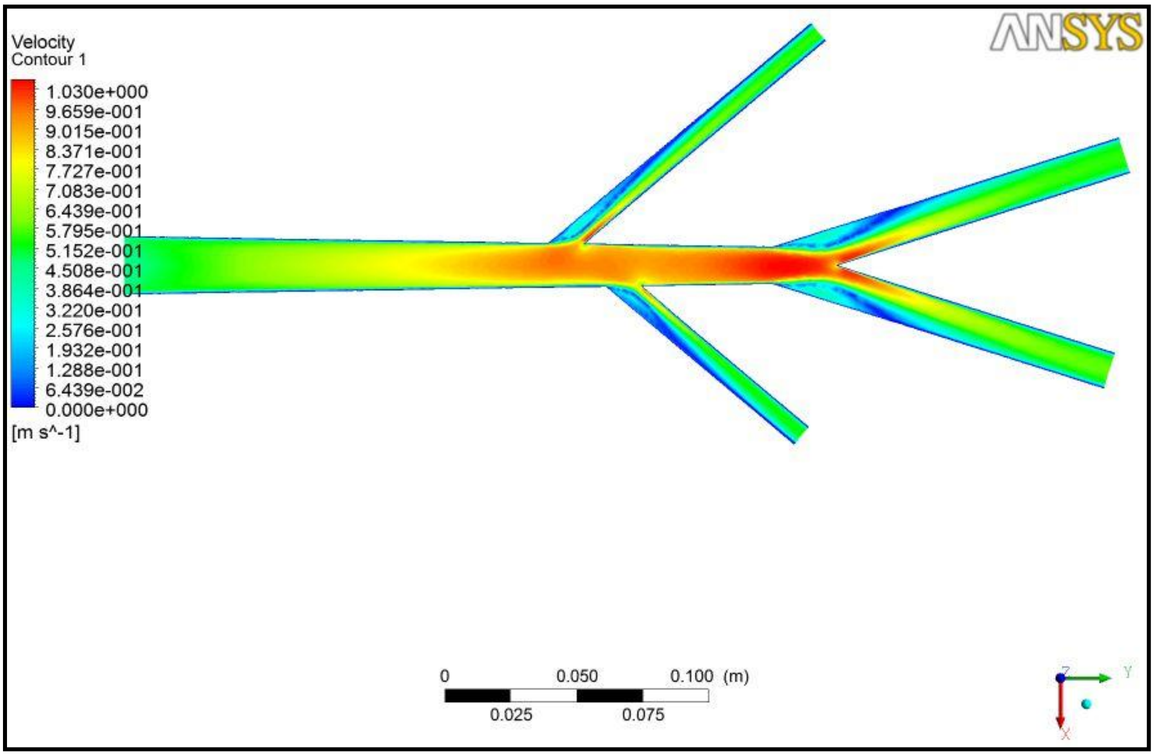

Figure 4.6. Sample of velocity contour

## 4.3 Vessel with Two Phenomena

Simulation is performed for this case with two phenomena, aneurysm and arthrosclerosis which are added to the left and right iliac branches. The results showed a little complexity to predict the flow parameter as branches and illnesses both together are added to the geometry for analysis. Velocity counters and streamlines illustrated when flow is passed through the domain in some parts especially at branches curvature (like previous case) and the illnesses parts became vortex as seen in [14] and [13]. The start of vortex flow in aneurysm region detected at 0.175 second and at 0.225 in anterior region of tapered part and gradually increased in posterior region of tapered part. The magnitudes of the vortices are increased at 0.275 second for both affected regions. For the other parts of vessel the size and region of the eddy flow was approximately similar to the previous case except in left renal branch that flow completely became vortex (Fig 4.7).

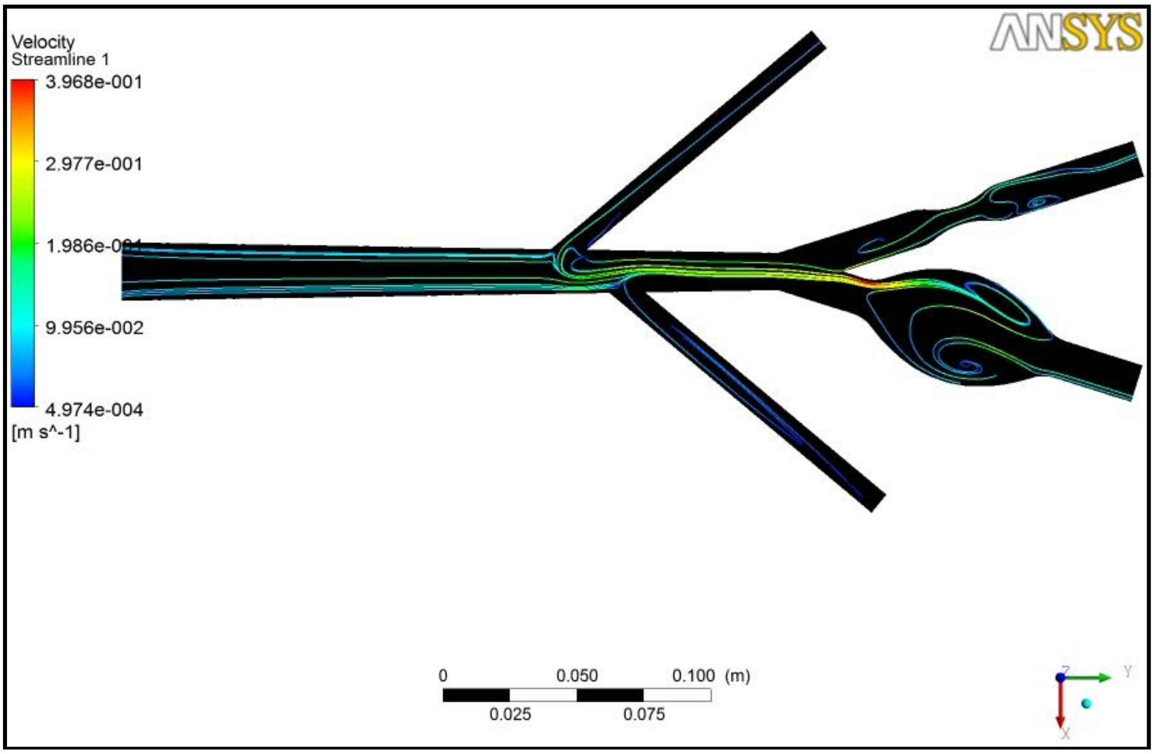

Figure 4.7. Sample of streamline contour

Velocity pulse which is taken for regions (beginning of descending part, near renal branches, end of descending part) showed this complexity as well. Although it was expected to have higher magnitude of velocity in some regions, particularly in peak part of the pulse, it became lower or equal to other parts (Fig 4.8).

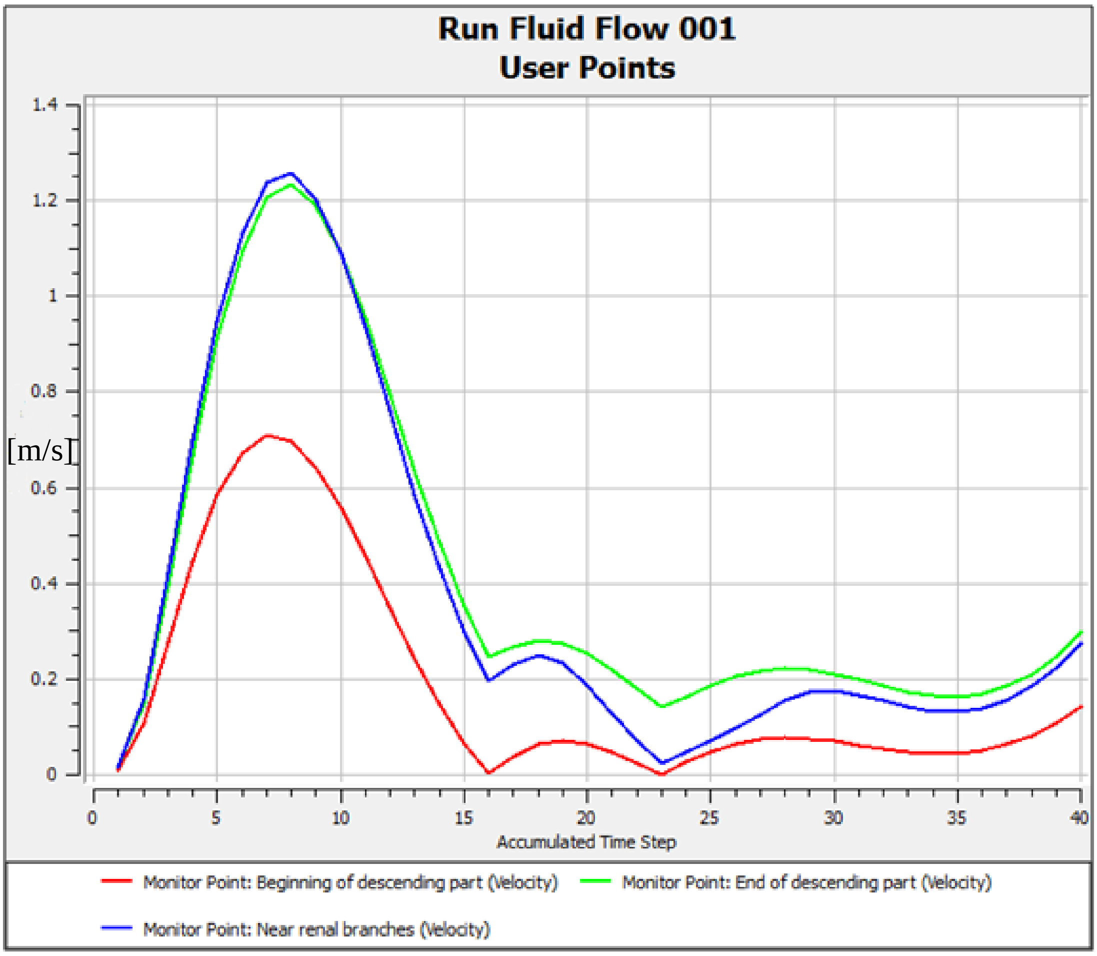


Figure 4.8. Time variant velocity pulse for region

As for the wall shear stresses, like thoracic healthy case with branches, the maximum magnitude of the wall shear stress are observed in the round curvature of the renal bifurcation and midpoint of the renal branches whose validation results described before in previous case. Also for lowest amount of the WSS, there was similarity for these regions. As for the area of geometry where pathological conditions are added, the distribution of the WSS was a sort of unpredictable. The highest value of WSS stresses for illnesses phenomena, especially for atherosclerosis in left iliac branch, was mostly observed at posterior and middle part of the affected area at many time steps as seen in [13] (Fig 4.9).

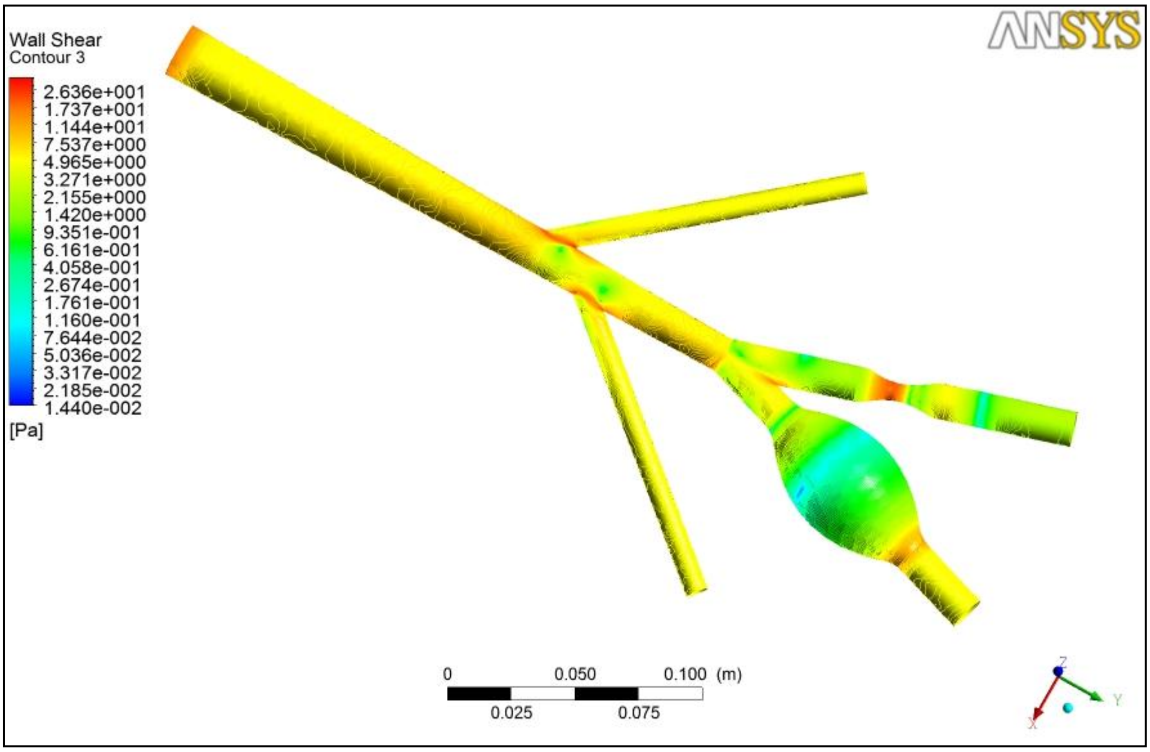


Figure 4.9. Sample of WSS contour

But after second 0.375 as the inlet mass flow rate pulse has downward trend from its peak and velocity pulse is progressed to far part of the geometry, the distribution of WSS in left renal became inhomogeneous. This heterogeneous behavior can be considered as the result of the vortex and pulsatile flow. For the aneurysm part highest value of the WSS are dispersed approximately next to the middle part of the aneurysm phenomenon (posterior) and in some steps at the end of the posterior curvature [28]. Also as the time passed, WSS scale was increased and dispread at the face between aneurysm and middle parts of the renal branches (Fig 4.10).

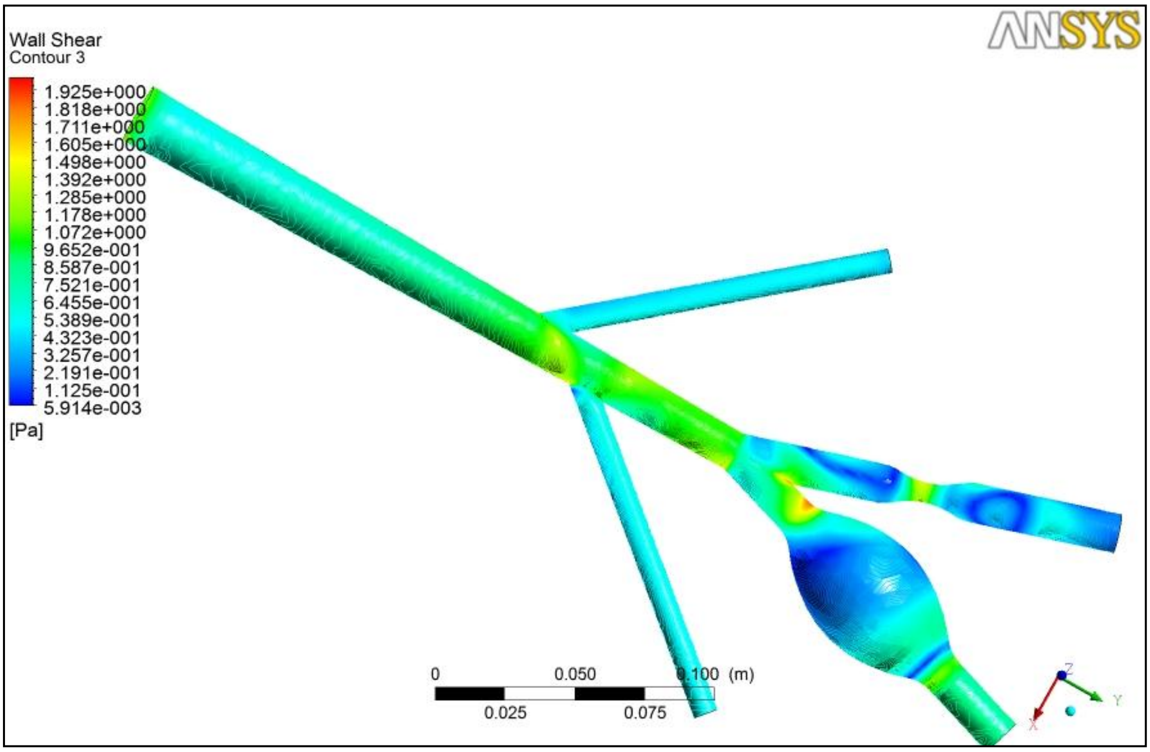


Figure 4.10. Sample of WSS contour

For anterior and posterior regions of the atherosclerosis and aneurysm the minimum amount of the WSS was observed; so it was for the outer faces of the renal branches (if it is assumed a symmetric geometry for the faces of the right and left renal which are not faced together). According to the observed regions and values for the high and low wall shear stresses, high value can lead to the rapture of the vessel particularly in affected parts, and low value of WSS will be a sort of prerequisite factor plaque and colt coagulation, particularly in the regions which have vortices flow around as seen in [13].

## 4.4 Vessel with Wall Structure

For this case vessel wall structure was added on the geometry of the fluid like a shell. It was defined as non-linear isotropic elastic material. Results showed that the wall elastic characteristic had effect on flow behavior and parameters. It affected the magnitude of the vortices to be higher in this case than the case without wall structure at the same time steps. Also by comparison of the velocity pulse for both this case and the case without wall structure, this observation showed that peak pulse was approximately 1.2 m/s. but in previous one it was more with some deviation at the start of the pulse in this case [14] (Fig 4.11).

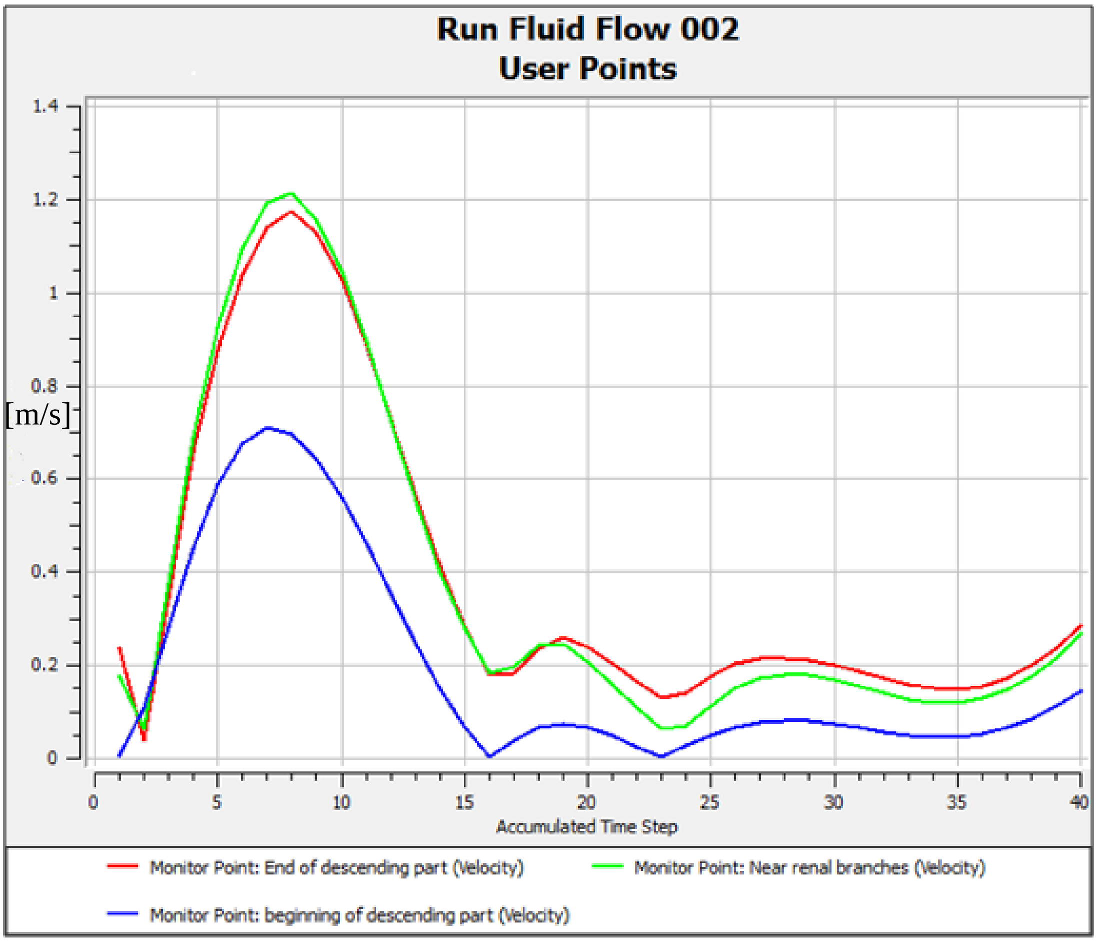

Figure 4.11. Time variant velocity pulse for  region

For the distribution of the Von-Misses stresses (equivalent stress) on the wall structure, it was observed that maximum amount of the stresses focused on the renal branches entrance curvature and at the middle of the iliac branches regions where also the highest value of the WSS were detected (Fig 4.12). Also WSS distribution for this case showed some different with previous one, but for the maximum and minimum region was almost the same. This finding shows the reliability between

WSS distribution and Von-Misses stresses. As for the effect of the pulsatile flow, it was observed that fluctuate amount of the forces were applied to the interface of the fluid and solid as seen in [11] (Fig 4.13).

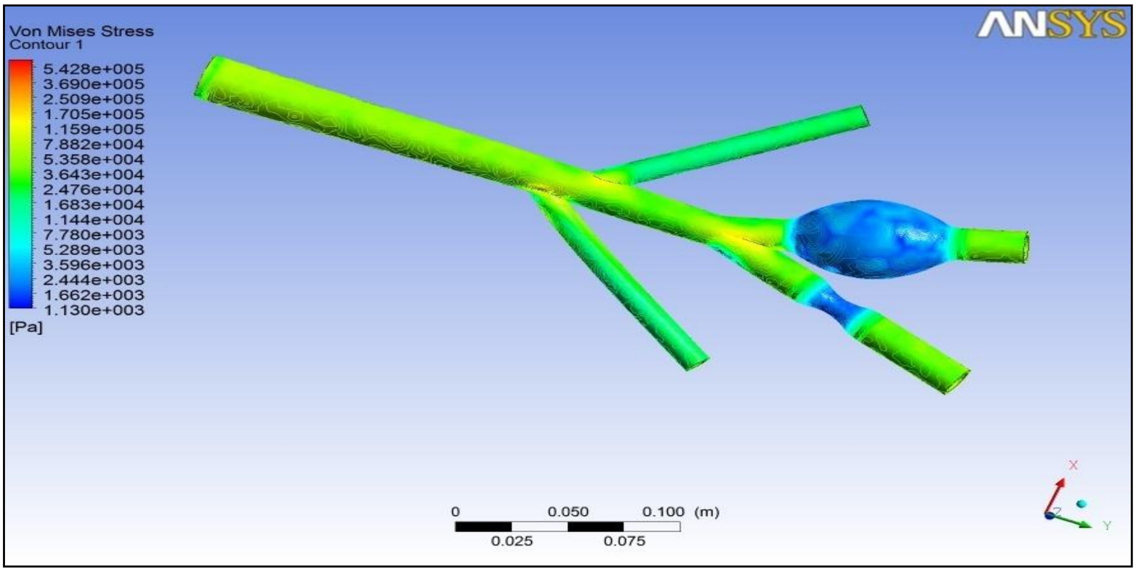


Figure 4.12. Sample of Von Misses stress contour

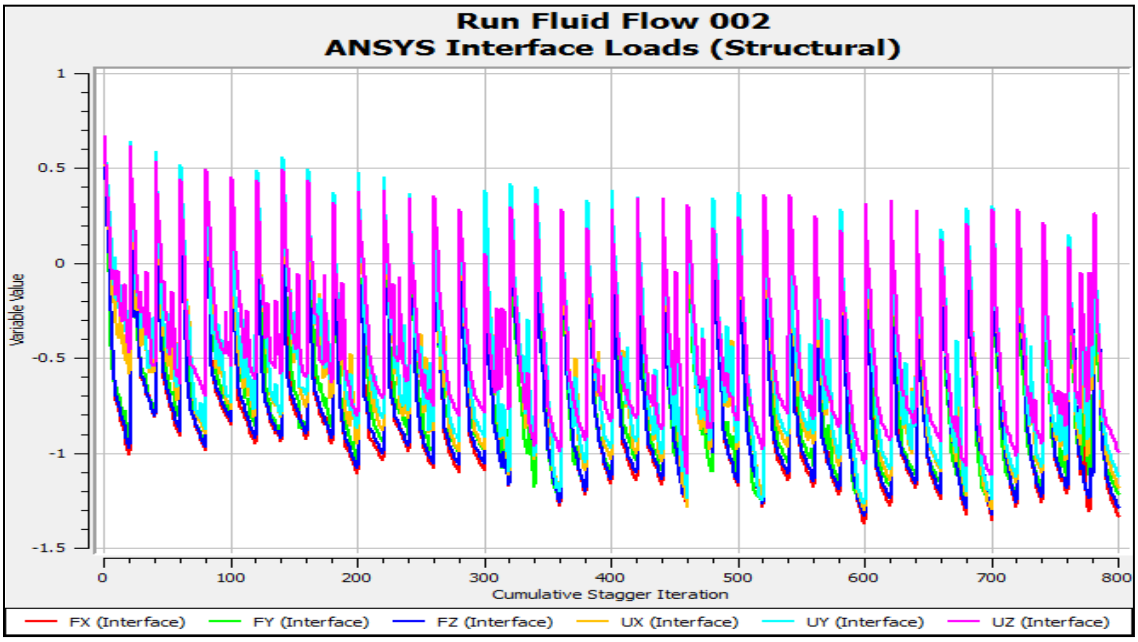


Figure 4.13. Wall interface force from fluid

# Chapter 5

# CONCLUSSION

Three-dimensional blood flow simulation has been performed in this study. In order to find the influence of geometry and wall structure, 4 geometries are constructed with and without branches considering whether they include vessel wall or diseases. A laminar pulsatile flow is analyzed considering non-Newtonian rheology for each case. For the exploration of the vessel wall effect, non-linear isotropic elastic material is introduced to the simulation.

The results showed the importance of considering the branches for geometry and their influence on flow behavior to do an accurate simulation. Observation of the vortices and magnitudes of them as well as time variant property of the velocity and pressure were the most important aspects of the blood flow analysis in branch-included model. The distribution of the WSS, whose highest value was at the beginning of branches with lowest value at the vortex region, on the wall revealed the importance of the bifurcation angles, geometry and presence of vortex flows. Presence of these parameters (WSS, vortex flow, pulsatile nature) demonstrates the possibility of the disease occurrence in the region where these parameters are localized. Although all these parameters can be affected by sex and age, the overall understanding of the results from these simulations and the stress distribution of the wall can be useful for the arteries plaque removal surgery and designing artificial vessel. Also observation of the vortices in simulation can lead to choose an appropriate turbulent model such as Large-eddy or K-ω for the future work and also as it was seen that the geometry and the bifurcations affect on flow behavior a lot, it is likely to choose MRI data to reconstruct the geometry for accurate analysis.

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