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Analysis of the Elastic Stress for the Bifurcated Region of Blood Vessel

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***Abstract*—In recent years, application of biomechanical modeling techniques in the area of medical image analysis and surgical simulation is a commonly method. In the development of medical surgery robot technology, haptic and virtual reality systems also have attended consideration on simulators as a tool of medical training. In order to enhance the authenticity of our vascular interventional surgery and improve the safety of operation, we intend to simulate the blood flow in the blood vessels, to create a certain operating environment for our operating system. The purpose of this study was to investigate the hemodynamic effect of different initial flow velocity of the blood in the vessel. In this paper, we have established a three-dimensional model of blood vessels, a Fluid Solid Interaction analysis of the reconstructed vascular model was performed. The vascular wall was modeled as an isotropic material. In order to investigate the correlation between flow-induced wall shear stress and geometry of the vessel, the vascular wall shear stress was computed. The velocity flow pattern and elastic stress were also measured. Results showed that blood flow velocity was found to reduce when the flow changed from the main stem to the bifurcated region. The high wall shear stress values calculated at the apex of the bifurcation indicated that this location is predisposed to form thrombus. Understand the changes of blood flow velocity and pressure in the local blood vessels, provides an important reference for the treatment of vascular disease.**

Index Terms – blood flow velocity; wall shear stress; Fluid Solid Interaction; vascular interventional surgery

I. INTRODUCTION

Cardiovascular disease is a common disease in modern society. According to the World Health Organization survey statistics, the world's annual deaths from cardiovascular and cerebrovascular disease up to 15 million. In particular, cardiovascular diseases represent 49% of death disease in Europe. Cardiovascular and cerebrovascular diseases occur frequently in some of the body's specific vascular parts, such as vascular stenosis, well-recognized regions of curvature, bifurcated area, vessel branches and so on. It is helpful to understand the morphology and structure of blood vessels and to know the causes of vascular diseases. At present, it is an effective method to treat cardiovascular and cerebrovascular diseases by vascular interventional surgery [1-3]. However, there is a high degree of complexity and risk in traditional vascular interventional surgery. This method has a high demand for doctors' experience and skills [4]. Virtual surgery simulation system can simulate all kinds of physical and physiological characteristics of human body [5], and can also be used to guide clinical interns to learn and train in clinical physiology, clinical surgery and so on. The virtual surgery simulation system with multisensory and interactivity, with the advantages of customizability and repeatability, so the virtual reality technology has been widely used in many fields such as military, medical and commercial [6] [7]. Among them, the application of this technique has been widely used in the medical research of vascular interventional surgery training system [8].

In the research of virtual surgery simulation system, the study of soft tissue's deformation is the key and difficult. Deformation simulation is the process and result of using computer to simulate the deformation of objects [9]. The measure of its quality is the timeliness and accuracy. The commonly deformation simulation methods based on the physical characteristics include the mass spring method (MSM) [11] [12], the boundary element method (BEM) [13], and finite element method (FEM) [10]. Finite element model is a continuous parametric model, the modeling precision is high and can simulate complex geometry structure, it can get more precise models of mechanics by adjusting the parameters and tectonic units of different shapes appropriate [14]. However, the calculation process of the model is complex and the computation time is very large, so it is difficult to meet the real-time requirements. With the continuous optimization of the model and the improvement of the algorithm, the computational efficiency is improved, and the application of the FEM in virtual surgery is more and more widely [15] [16].

FEM have been widely used for identifying mechanical properties and behavior of soft tissue. Picinbono et al. proposed the St. Venant-Kirckoff nonlinear FE model to simulate liver indentation behavior. Tendick et al. used the Mooney–Rivlin FE model to simulate tissue deformation in a training environment. The above research clearly shows that FE models are a promising approach for predicting tissue behavior and identifying tissue properties. Many works pay their attention on properties of soft tissue from the medical point of view. They focus on study hemodynamic aspects and the interaction between blood flow and vascular wall, since hemodynamics and vessel geometry play an important role in the cause of plaque formation [22]. Some other researches are focused on stenting technique [18] [19]. Huo et al. studied the effect of the compliance on the WSS computation for a porcine RCA comparing rigid wall and FSI simulations. Theodorakakos et al. investigated the effect of the myocardial motion on pulsating blood flow distribution on a narrow LAD branch coronary artery. Berthier et al. constructing CFD models based on RCA angiograms. They studied the flow differences caused by assuming different vessel cross-sectional shapes and sizes. Chaichana T et al investigated the hemodynamic effect of variations in the angulations of the left coronary artery, based on simulated and realistic coronary artery models [20]. J. Ohayon et al. analyzed the correlation between coronary artery wall stiffness and plaque formation in eight patients [21]. Most of studies neglecting the contribution of the blood flow. Therefore, the aim of this work is to evaluate how the initial velocity can affect the blood flow pattern and the wall shear stress in human vessels using FSI analysis. The main contents of this paper include: extract the brain blood vessels by using Mimics software and then reconstruct the three dimensional of cerebral vessels. Later we would focus on simulation for deformation of blood vessels.

II. MODELING OF BLOOD VESSEL

A. *Vascular reconstruction and numerical grid generation*

In the establishment of a human vascular model, it is more likely to build complex geometric models that are more similar to those of the actual blood vessels. It is common to use imaging modalities such as computed tomography (CT) and MRI to seek tumors information preoperatively. Augsburger et al. construct a hemodynamics mo del of a narrow carotid artery based on CT images. However, due to the deformability of the soft tissue during the surgical procedure, it is very difficult to accurately record the images to the intraoperative tumor locations. In this paper, we use the software Mimics to extract the blood vessels of the brain. The CAT data used in this paper is provided by Kagawa University. At present, Mimics is a standard 3D image processing and editing software based on scanning data. It is a visual interactive tool for CT, MRI images and 3D rendering objects. It can directly build the 3D model for the input scan data, then output the general CAD, FEA and RP formats, and the data can be converted on the computer. Mimics provides a number of basic modules can be directly connected with CAD, ANSYS and other software, the model can be directly imported into the professional software for further analysis and processing.

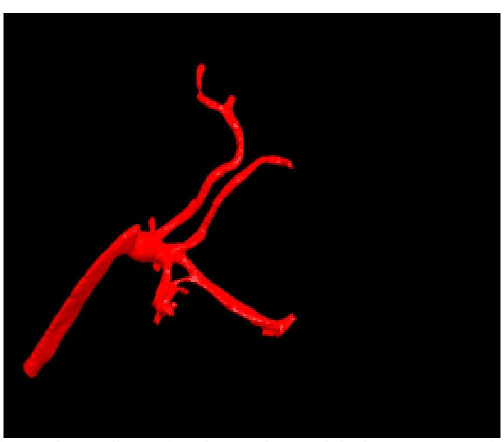


Fig. 1 Three- dimensional vascular model after reconstruction

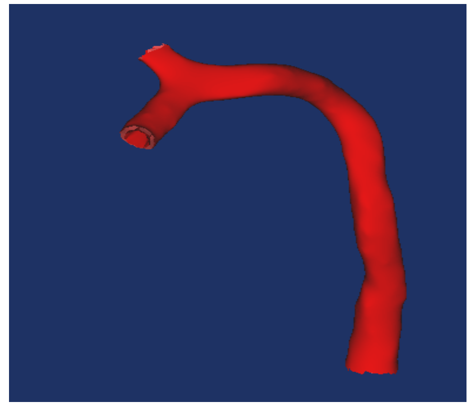


Fig. 2 Three -dimensional vascular model after optimization.

In this paper, we first import the CAT data of cerebral blood vessels into the Mimics for the basic treatment, and then reconstruct the 3D model of the blood vessel with the method of shearing deformation. In the FEA module, re-divide the grid for the generated 3D model. Then transform surface mesh model into solid mesh and reimport the body grid to FEA module to give the material properties, and then output. This can improve the quality of the model and improve the processing speed of the model in finite element analysis.

The 3D vascular model after reconstruction is shown in Fig. 1. Due to the 3D model reconstructed is rough, it also need to optimize and further processing by covering and smoothing, to improve the quality of the surface mesh model. Eventually get a more satisfactory vascular model. Then, the initial rough geometry was smoothed and the computational fluid mesh was built. In order to make a clear observation, only show parts of the optimized model. The optimized vascular model is shown in Fig.2.With the 3D vascular model reconstructed by Mimics can not only measure the vessel diameter, branch angle, can also be rotated, the observation model from multiple perspectives, to understand the anatomic structure and position of the vessel has an important role.

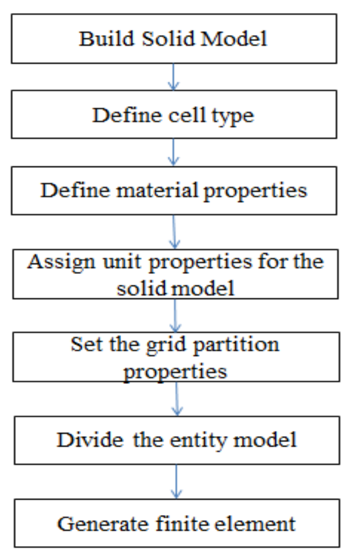


Fig. 3 The flow chart of setting up the finite element model

The finite element analysis includes three processes: pretreatment, analysis and calculation. It is important to reconstruct the model in the pre-processing. The specific process of vascular model pretreatment is shown in Fig. 3. The modeling work has been done in the Mimics software, only need to import the model into ANSYS, to set up the material property and the partition of the mesh. The biggest difference between the finite element model and the geometric model is that it is not only closer to the actual object from the appearance, but also has some biomechanical properties such as deformation and quality. This requires the establishment of material properties for the model to give the corresponding biomechanical properties.

The fluid tetrahedral mesh was generated starting from the internal shell that represents the numerical fluid solid interface domain. The final mesh was imported into the ANSYS where the FSI and rigid wall simulations were performed. The computational fluid grids used in this study possessed finally 38849 elements. For the solid grid, a full tetrahedron mesh of 7186 elements was created. Considering the complexity and irregular shape of the vascular model, this paper adopted the free mesh which is suitable for all models. Select tetrahedron as cell shape. Moreover, for reason of research purpose, the inlet and outlet diameter of the model were same.

B. The intravascular blood flow and vascular wall properties

Learn from the previous works on human arteries, the blood rheology was assumed as Newtonian. According to literature, the blood viscosity and blood density are assumed to be constant, the blood viscosity was taken as 0.0035 kg·m·s-1 and blood density 1150 kg·m-3, respectively. Since the Reynolds number based on the average arterial diameter was Re = 150, the blood flow was assumed isotropic rand incompressible under unsteady flow conditions. The vessel wall was modeled as an incompressible isotropic and homogeneous material [23]. In this study, the viscoelasticity and the intrinsic anisotropy were not considered. The parameters of the blood vessel are shown in Table 1. Based on the above assumption, the governing equation of Newton fluid is a viscous incompressible N-S equation, as in (1). Where u is velocity vector, ρ is density of the blood, t time, p pressure, μ viscosity of blood (flow). The mass conservation equation of blood is defined as in (2).

$$\begin{cases} \rho(u_t + (u \cdot \nabla)u) - \mu\Delta u + \nabla p = 0 \\ \nabla \cdot u = 0 \end{cases} \quad (1)$$

$$\partial(u)/\partial(x) + \partial(v)/\partial(y) = 0 \quad (2)$$

$$u_\Gamma = 0,\ v_\Gamma = 0 \quad (3)$$

$$\bar{v} = \text{const} \quad (4)$$

TABLE I
THE PARAMETERS OF THE BLOOD VESSEL

Young's Modulus (Pa)	Poisson's Ratio	Bulk Modulus (Pa)	Shear Modulus (Pa)
5E+05	0.45	1.6667E+06	1.7241E+05

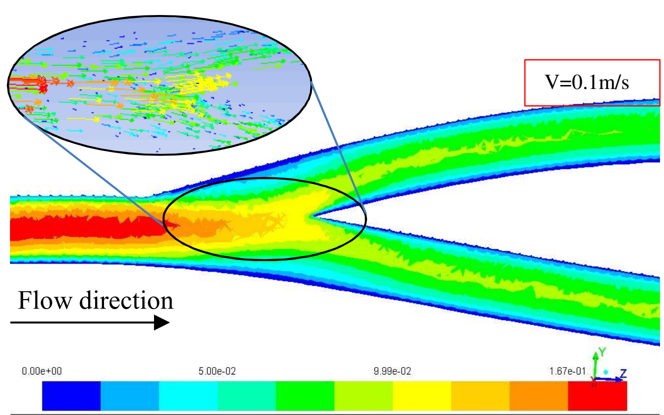
C. Boundary conditions

FSI was employed for the simulation, for describing and analyzing the flow of fluid elements at each location in certain geometry. In order to simulate realistic blood flow, a domain of interest must be defined, and boundary conditions specified. All geometrical data were put into a specialized preprocessing program for grid generation. And then, all grid data as well as flow data determined from the boundary conditions were imported into the solver. As for FSI analyses, mixed velocity–pressure condition are required in order to correctly compute flow features and structural stresses, strains and displacements. No-slip boundary condition was imposed at the surface of the vascular wall (as in (3),Γis represent vascular wall), which represented the fluid-structure interface for the FSI simulation. Pulsatile velocity was applied as an inlet boundary condition at the main stem, and pulsatile pressure was applied at the daughter channels as an outlet boundary condition. The velocity inlet boundary condition is defined by (4). The assumptions made about the nature of the flow are that it is 3-D, steady, laminar and isothermal.

III. ANALYSIS AND EVALUATION

Because of the cyclical contraction and relaxation of the heart, the blood that enters the aorta from the left ventricle affects the entire pulsating vessel in the form of pulsating waves, causing periodic changes of the blood pressure and flow in the blood vessel, and its period is equal to the cardiac cycle .The average frequency of human heat beat is 75 times per minute, heart beat cycle is 0.8s. During the calculation process, we simulate the blood flow at different initial inlet velocities, these include v=0.1 m/s, 0.15 m/s, 0.2 m/s, 0.25 m/s and 0.3 m/s. The velocity profile of the flow field in the blood vessel at the entrance velocity of 0.1 m/s and 0.3 m/s are shown in Fig.4.

The vector labels range from 1 to 10 corresponding to the 10 color levels shown in the bar. Results showed that a small region of low-velocity blood flow distributed in the model when the initial inlet velocity is small and gradually became a large region of flow separation when the inlet velocity were increased.



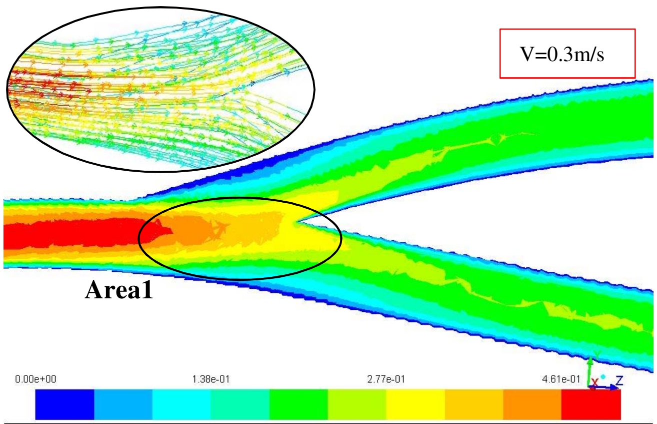


Fig.4 Velocity distribution at the inlet velocity of 0.1m/s and 0.3 m/s

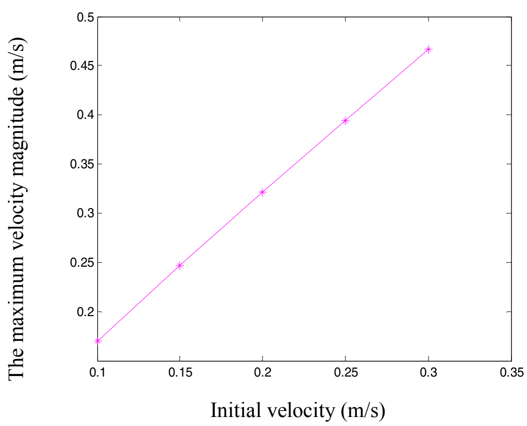


Fig.5 The relationship between the maximum velocity and the initial velocity

In this paper, the relationship between the maximum velocity and the initial velocity is obtained by studying the variation of the maximum velocity of the blood vessels with the initial velocity of the inlet, as shown in Fig.6.It can be seen that the maximum velocity and the initial velocity of the inlet presents a function or linear change. In Fig.4 the blood flow velocity is relatively large in the main stem, the velocity became slower after encountering the branch. With the increase of the initial blood flow velocity, the flow velocity and the maximum velocity of the blood flow in the branch increased. It appeared low velocity zoon at the bifurcate position, which is the best location for the formation of thrombus. The faster the blood flow velocity the more likely to form thrombus.

Wall shear stress (WSS) plays a central role in atherogenesis for vessel hemodynamic. Wahle et al. studied the correlation between WSS and plaque distribution in a set of 48 in vivo vessel segments based on a 3D reconstruction of intravascular ultrasound images. Their findings confirmed that relatively lower WSS is associated with early plaque development. Fig.5 shows the wall shear stress contours value .As it is apparent, WSS on the bifurcation apex is taking very high values compared to the rest of the vascular walls. Due to the bifurcation of the blood vessel, the vascular geometry has changed, resulting in a sudden change in the hemodynamic effect at this location. Concentrating our attention to the bifurcation (see Fig.6), the shear stress of the wall has a sudden change at the bifurcation position. And with the increase of the initial velocity, the mutational range also increased. When the initial velocity increased from 0.1 m/s to 0.3 m/s, corresponding to the WSS value at the bifurcation is increased from 2.39 Pa to 12.15 Pa.

In Fig.7 is shown a comparison of the maximal shear stress between different initial velocities. The analysis of blood velocity and WSS has been recognized to provide early biomarkers of the atherosclerotic plaque formation and growth. Shear stress is regarded as an important factor leading to the focal distribution of atherosclerotic plaques. Some studies have been performed using other imaging modalities to further verify the correlation between WSS and plaque development, such as extraction of vessel center line from coronary CT angiography, reconstruction of side branch with coronary CT angiography data for investigation of 3D shear stress distribution in coronary bifurcations [17].

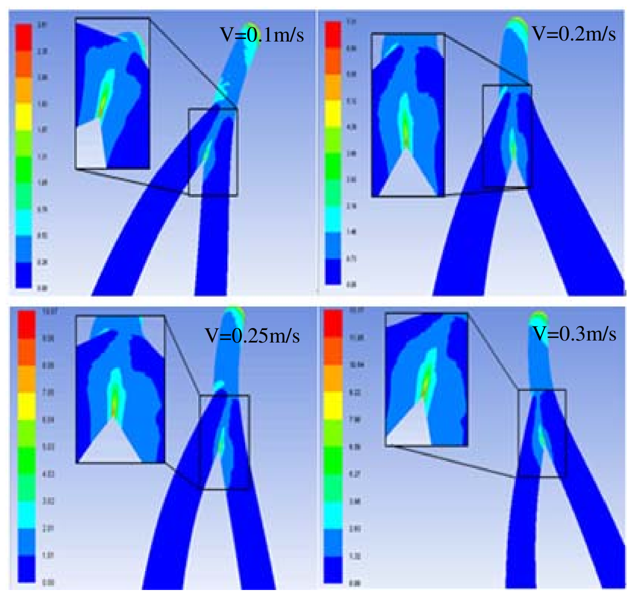


Fig. 6 Contour plots of the wall shear stress at different initial velocities.

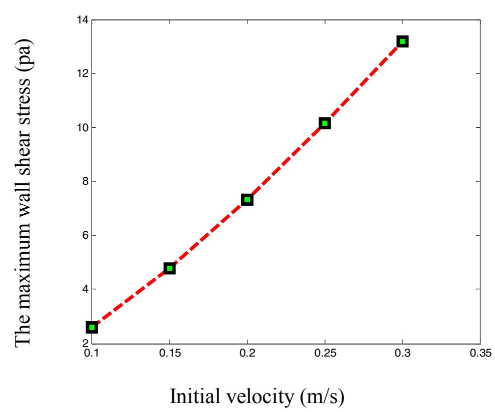


Fig. 7 The maximal wall shear stress value at different initial velocities

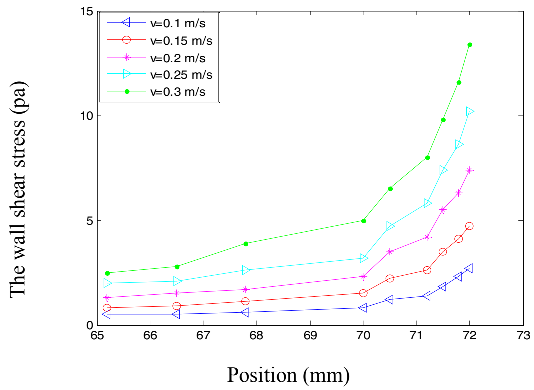


Fig.8 The wall shear stress distribution at area 1 with the different velocities

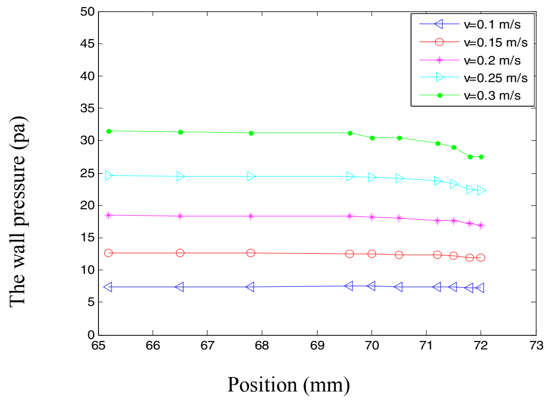
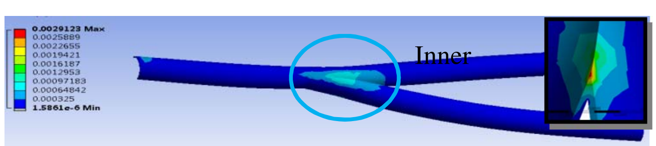
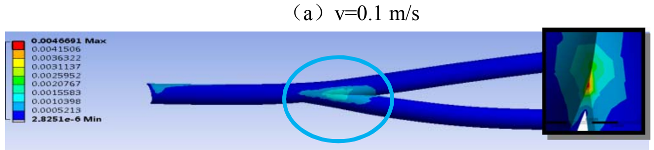


Fig.9 The wall pressure distribution at area 1 with the different velocities

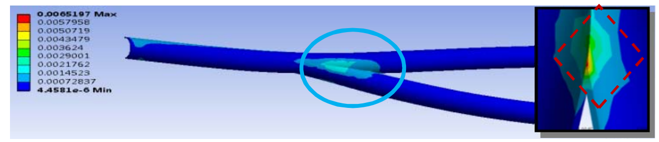
Study on the influence of different inlet velocities on wall shear stress and wall pressure .The distributions of wall shear stress and wall pressure near zone 1 are obtained, as shown in Fig. 8 and Fig.9.The results show that the more close to the bifurcation the greater the wall shear stress, while the wall pressure is gradually reduced. It can be seen from Fig.7 that the wall shear stress increases first and then decreases along the extension direction of the vessel. With the increase of inlet velocity, the wall shear stress and wall pressure in the vicinity of this area are increasing on the whole. This is because the velocity will suddenly change the direction in bifurcation, the blood must be subjected to a force that changes the velocity, the reaction of the force is the wall shear stress and wall pressure. From the medical or physiological point of view, the wall shear stress and wall pressure value is high, will directly damage the endothelium and promote inflammation, which increase the probability of the formation of thrombosis.



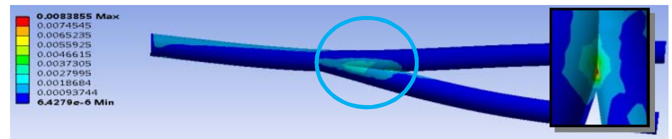
（a）v=0.1 m/s



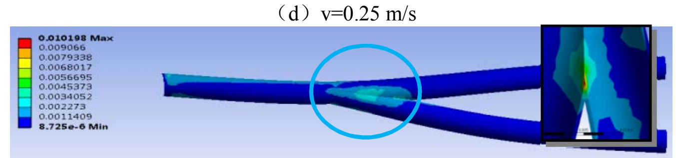
（b）v=0.15m/s



（c）v=0.2 m/s

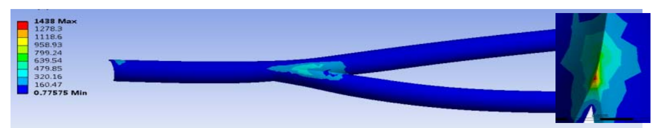


（d）v=0.25 m/s

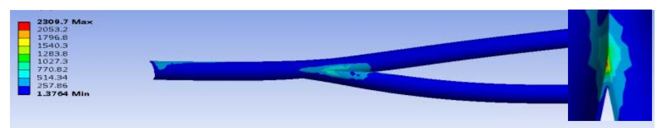


（e）v=0.3 m/s

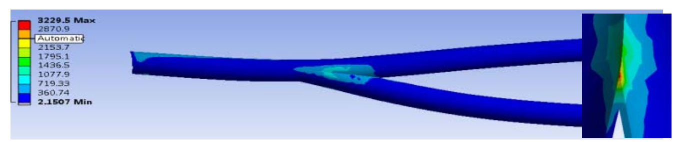
Fig.10 Equivalent elastic strain distribution of the vascular observed with variable initial velocities.



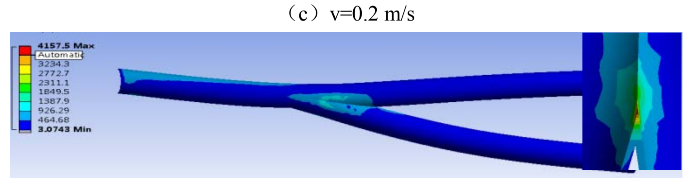
（a）v=0.1 m/s



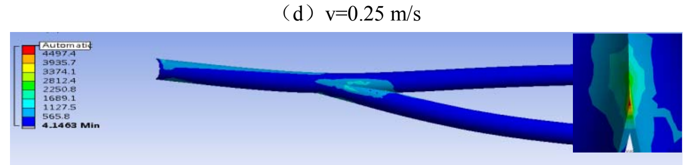
（b）v=0.15m/s



（c）v=0.2 m/s



（d）v=0.25 m/s



（e）v=0.3 m/s

Fig. 11 Equivalent elastic stress distribution of the vascular observed with variable initial velocities.

In Fig.10 and Fig.11 the equivalent elastic strain distribution and equivalent elastic stress distribution are shown at selected time .The maximal strain and stress located in a small region of the bifurcation as can be seen from some literature. However, it has to be noted that the values we found take into account the circumferential strain starting from the zero-pressure configuration, which at the time of t=0 s. Low strain region were found low-deformation, region with the increased strain was easy to form large deformation and cause damage to vascular wall.

IV. CONCLUSION

In this paper, the purpose of our study is to analyze the dominant hemodynamic flow parameters [wall shear stress (WSS), and velocity] and the pressure waveforms of the vascular wall. A three-dimensional model of blood vessels, based on average human data set extracted from medical software Mimics, was adopted for finite element analysis. From the results can be seen that in the process of blood flow, the blood flow velocity has changed abruptly in the blood vessel bifurcation position, a low velocity appears at the bifurcated blood vessel. At the same time, the wall shear stress on the vessel wall near the region is also changed abruptly. This region is regard as the best place to form thrombus. And then becomes slow after entering the branches. The results also showed remarkable qualitative discrepancies in the WSS distribution. These results confirm that as the initial velocity increases, the most suitable position for formation of thrombus increases, the wall shear stress at the same place also increases.

The novelty of this paper is to reconstruct the 3D model of blood vessels by using Mimics medical software and optimize it, which greatly improves the accuracy and modeling speed of the reconstructed model. In addition, the use of fluid-solid coupling method to analyze the vascular model, not only can provide a variety of fluid-related hemodynamic parameters, and can more accurately provide the distribution of vascular wall stress and displacement, which can provide a reference for the study of mechanical characteristics of blood vessels.

The limitation of our present research is neglect the effect of branch angles on the blood vessels, several modifications of the present model would be necessary. In addition, the influence of the elastic modulus of the vessel wall and the thickness of the vessel wall on the vascular deformation is also considered.

In the future, we can use the above analysis as a theoretical guide for our vascular intervention surgery system to create a specific operating environment, increasing the authenticity of surgical operation and improving the safety of our operating system.

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