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내부직경 변화를 고려한 혈관의 유체-구조 상호작용 해석

Fluid-Structure Interaction Analysis of Blood Vessel Considering Internal Diameter Variation

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ABSTRACT

A three-dimensional elastic blood vessel model with internal diameter variation is considered to investigate internal flow characteristics and effects of structural deformation. Also, computational analyses for both the rigid model and the fully-coupled FSI model have been conducted in order to compare the shear stress, pressure distribution, and flow velocity in detail. A 70% narrowing area of asymmetric blood vessel model was especially investigated to show the versatility of fluid-structure interaction phenomenon. The results reveal that effect of fluid-structure interaction is very important to accurately investigate the flow characteristics of the blood vessel.

1. Introduction

The study of blood vessel considering internal diameter variation is based on one of the arterial disease which is directly pertained to blood flow restriction by thickening the arterial wall (stenosis). The severity of the disease gives the adverse effect related to high velocity, negative pressure, high shear stress, flow separation, or even collapse at the narrowing area which could directly lead to stroke and heart attack [1-4]. Many investigators have conducted the research of the flow characteristics in the stenotic blood vessels in order to assess plaque cap rupture mechanism [1-2]. The exact mechanism of this phenomenon is still inconclusive owing to lack of experimental data and limitation of computational method [2]. Thus, it is needed to conduct advanced computational analyses for complex fluid-structure interaction (FSI) phenomena in order to investigate detailed mechanism of blood vessel flows.

In this paper, blood flow and FSI analyses for artery have been conducted based on the advanced computational methods such as computational fluid dynamics (CFD) and finite element method (FEM). The blood vessel with 70% narrowing area which is categorized as the case of severe stenosis is considered. The artery was modeled as a three-dimensional elastic tube with 70% asymmetric thickening wall and 50% eccentricity to make the model more relevant to the physiological case. For the present structural model, the nonlinear Mooney-Rivlin material properties are used to construct the proper model for artery wall. It is shown from the results that the there are critical flow characteristics such as the high velocity profile, negative pressure value, and high shear stress, which contribute to the collapse mechanism related to the arterial disease in reality.

2. Computational Backgrounds

For the computational analyses of special FSI problem on blood vessel model, a commercial finite element software named as ADINA (Ver.8.3) is used in this study. In FSI analyses for a flexible blood vessel, the force by blood flow can generate structural deformation and because of this inside flow characteristics can also be changed [5]. In the computational analysis, the fluid and structural models are defined in their own domains, the structural model is based on the Lagrangian coordinate system and fluid model complied with the Eulerian coordinate system. The ALE (Arbitrary Lagrangian Eulerian) formulation is imposed to exactly solve unknown variables related to the fluid-structure interaction.

In this study, the blood flow is assumed to be laminar, viscous, Newtonian, and incompressible because of its inherent flow characteristics. No-slip boundary condition is assumed to consider the flow viscosity between the fluid and the wall surface of blood vessel. Two different computational models are considered in this study. One is rigid blood vessel model and the other is elastic blood vessel model.

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For the rigid blood vessel model, the wall with infinite stiffness is specified at the outer surface of the fluid domain to confine the flow. The three dimensional fluid model with 70% severity and 50% eccentricity is considered and analyzed. The nonconservative forms of Navier-Stokes equations are used to prescribe the flow in the wall as follow:

$$\begin{aligned} &\rho \nabla \bullet v = 0 \\ &\rho \frac{\partial v}{\partial t} + \rho v \bullet \nabla v - \nabla \bullet \tau = f^B \end{aligned} \tag{1}$$

The fluid geometry (Fig.1) with 70% severity and 50% eccentricity was defined in ADINA-M. Flow analyses for rigid wall structural model have been conducted using two different kinds of mesh. The free-form mesh is generated using 4-node tetrahedral volume element for the first model and the other model using hexahedral element. Corresponding computational grids are presented in Figs.2-3. The narrowing area shape of the blood vessel is defined with the following equation:

$$H_0(z) = R_0 - S_0 R_0 \{1 - \cos[2\pi(x - x_1)/(x_2 - x_1)]\}/2,$$

$$x_1 \le x \le x_2 \tag{2}$$

where R_0 is the internal radius of the straight part of the blood vessel, x_1=3.2 cm and x_2=4.8 cm, are the beginning and the end of the narrowing area, respectively.

The preceding equation is the shape equation for the symmetric narrowing area. In order to create the asymmetric model with the eccentricity, the centerline of the narrowing section of symmetric model was moved to:

$$z_1 = d(x) = d_{axix} \{1 - \cos[2\pi(x - x_1)/(x_2 - x_1)]\}/2,$$

$$x_1 \le x \le x_2, z_2 = 0 \tag{3}$$

Here, S_0 and e is the stenosis severity and eccentricity respectively which are defined as:

$$S_0 = (R_0 - R_{throat})/R_0 \times 100\% \tag{4}$$

$$e = d_{axis}/(R_0 \cdot S_0) \times 100\% \tag{5}$$

In this study, the fluid properties are assumed as water with density, 1 g/cm^3, and the viscosity is 0.04 g/cm s. The inlet and outlet pressure is prescribed at the beginning and the end of fluid model like artery blood flows, as following:

$$p|_{x=0} = p_1, \quad p|_{x=l} = p_2, \quad p_e = 0 \tag{6}$$

$$u|_\Gamma = (u, v, w)|_\Gamma = (0,0,0) \tag{7}$$

where u, v, and w are the radial, angular, and axial components of the flow velocity respectively, p_e is the external pressure, p_1 and p_2 are the inlet and the outlet pressure. The inlet pressure value is 100 mmHg and outlet pressure is 10 mmHg. Normal-traction is imposed at the inlet and the outlet of fluid model.

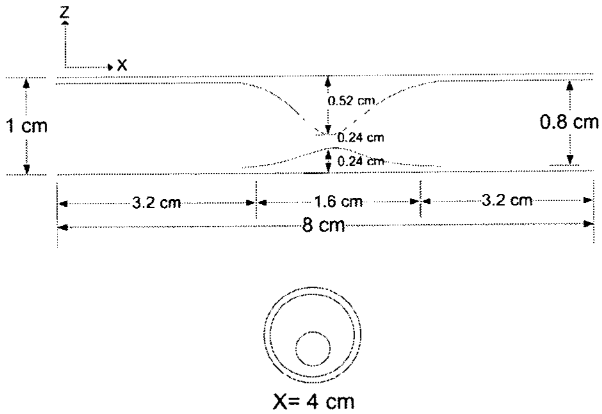

Fig.1 Geometric configuration of the blood vessel model with a stenosis

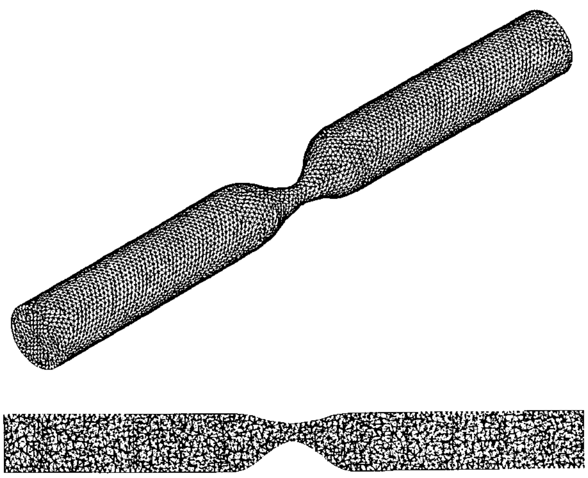

Fig.2 Computational grid for the fluid model using tetrahedral element

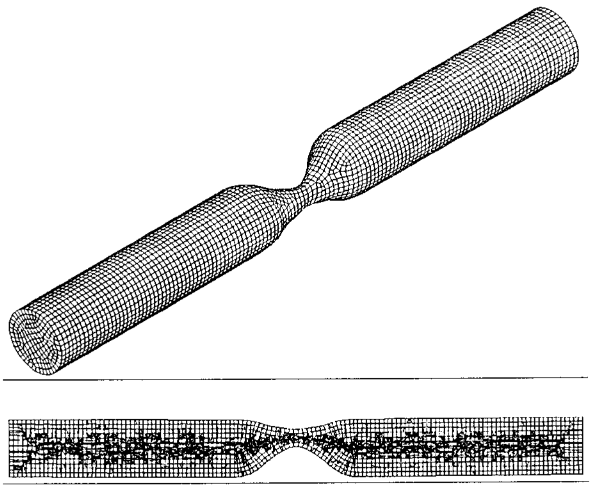

Fig.3 Computational grid for the fluid model using hexahedral element

For the elastic blood vessel model, the structure of blood vessel is assumed as a flexible pipe with nonlinear stress-strain relation. Also, the rigid wall boundary condition is replaced by the fluid-structure boundary condition. The FSI boundary is defined at both the outer surface of the fluid geometry and the corresponding inner wall surface of the elastic structure model.

The arterial wall is modeled as isotropic, hyperelastic, and incompressible. The Mooney-Rivlin material is considered to be appropriate with the mechanical behavior of arterial wall. A large displacement kinematics assumption was defined and deployed by Total Lagrangian (TL) formulation.

Delfino [6] has characterized the artery wall by strain energy function, W:

$$W = \frac{a}{b}\left[\exp\left(\frac{b}{2}(I_1 - 3)\right) - 1\right] \tag{8}$$

where a, b are material constant, and I_1 is the strain invariant.

This preceding equation is modified to prescribe the strain-energy density of hyperelastic part of Mooney-Rivlin material [2] as follows:

$$W = C_1(I_1 - 3) + C_2(I_2 - 3) + D_1[\exp(D_2(I_1 - 3) - 1)] \tag{9}$$

where C_1, C_2, and D_1 are the material constants, I_1 and I_2 are the strain invariants. The principal stress is given by the following equation:

$$\sigma = \frac{\partial W}{\partial \lambda} = c_1(2\lambda - 2\lambda^{-2}) + c_2(2 - 2\lambda^{-3}) + D_1 D_2(2\lambda - 2\lambda^{-2})\exp[D_2(\lambda^2 + 2\lambda^{-1} - 3)] \tag{10}$$

For an incompressible material, the uniaxial stress/stretch relation is

$$\lambda_1 = \lambda, \lambda_2 = \lambda_3 = \lambda^{-\frac{1}{2}} \tag{11}$$

where λ_1, λ_2, and λ_3 are stretch ratio in (x,y,z) direction, respectively. The stress/strain relations for Mooney-Rivlin material is presented in Fig. 4.

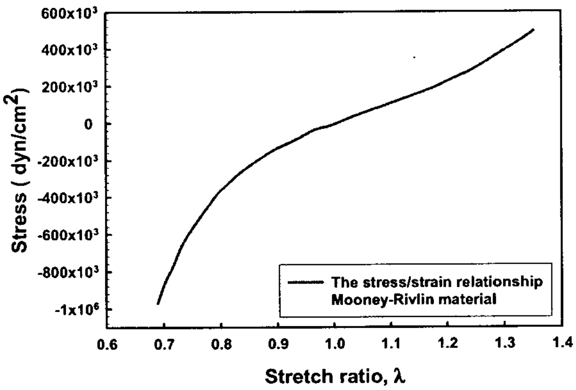

Fig.4 The stress-strain relation of Mooney-Rivlin material

The elastic wall of blood vessel is defined in ADINA-M and discretized by using three-dimensional solid element such as 8-node hexahedral element (Fig.5). At the beginning and the end of the tube, no displacement is allowed in any direction to prevent the tube in slipping away.

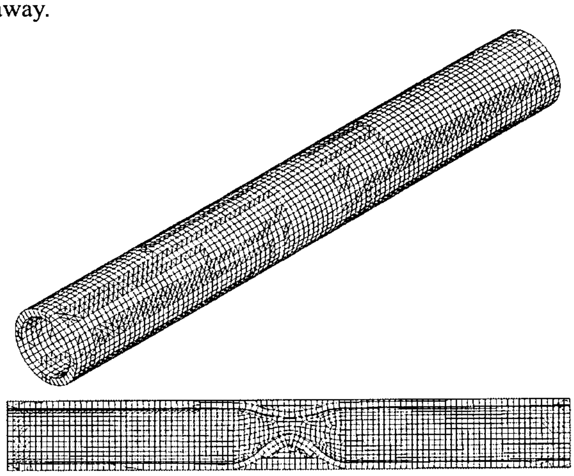

Fig.5 The finite element structural mesh for the blood vessel

3. Results and Discussion

The simulations have been performed using server computer with 3.1 GHz CPU and 2 GB RAM. It consumed 25.8 hours computational time to obtain the convergence solution for the present FSI model. The number of time steps needed to complete the solution is 120. For flow analysis using 4-node tetrahedral element, the computational time is 6.97 hours with 110 time step, but the analysis using hexahedral element took 23.7 hours and using 276 time steps. The number of fluid element used in this simulation is 79,846 and the number of solid element is 12,645. For the rigid structural model using hexahedral element, the number of element using in this analysis is 58,066, which is comprised of 26,422 4-node hexahedral elements, 19,801 8-node hexahedral elements. The convergence tolerance was set to 0.001.

For both models, the maximum shear stress occurs at narrowing area for 0^0 line angle. It also must be noticed that the maximum velocity still attach line $\theta = 0^0$ of the section after stenosis. Shear stress reachs the maximum value of 408.7 dyn/cm^2 and 519.0 dyn/cm^2 for tetrahedral element and hexahedral element, respectively. The higher shear stress value for hexahedral element is reasonable because the control volume upwinding technique gives more accurate result in fluid analysis.

A 70% narrowing internal area of blood vessel model shows that there is flow recirculation at the area after stenosis which is given by Fig.6. The maximum velocity for the rigid wall model is 443.1 cm/s for tetrahedral element and 439.3 cm/s for the FSI model. The result apparently shows that velocity value has no big difference between both analyses. Thus, the mechanical properties of elastic wall model which is less stiffness than rigid structural model make the result for velocity value reasonable.

Fig.7 shows that negative pressure also occurs for both of the analyses at the narrowing area of the blood vessel. The pressure changes drastically at narrowing area. Minimum pressure value for the rigid structural wall model is -10.52 mmHg and for the elastic structural model using FSI analysis is -3.04 mmHg. The comparison result of pressure distribution along 0^0 line angle is plotted in Fig.9.

The structural deformation in Fig.8 shows the large displacement along radial direction for the arterial wall. Asymmetric contour plot of displacement magnitude indicates the asymmetric stress distribution along the tube due to the fluid force. The maximum displacement value is 0.4026 cm which is taken place at the proximal side of the tube. The shear stress distribution which is plotted for both analyses in Figs.10-11 show that maximum shear stress occurs at narrowing area for 0^0 line angle. For the rigid structural model, shear stress reach the maximum value of 408.648 dyn/cm^2 at 0^0 line angle. The maximum value of shear stress for the FSI analysis is 503.03 dyn/cm^2 which is present at the same line angle. For 90^0 line angle in FSI analysis, there is a significant different pattern of shear stress distribution compare to rigid structural model. It shows large oscillatory shear stress which indicates larger area of flow separation which is caused by the strong couple interaction between fluid and wall at the area after stenosis. The viscosity gives large contribution to the oscillatory pattern of shear stress in FSI analysis. This condition could be related to increase the plaque formation at narrowing area.

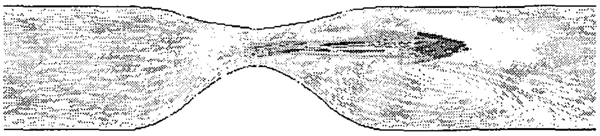

Fig.6 The velocity profile along the tube

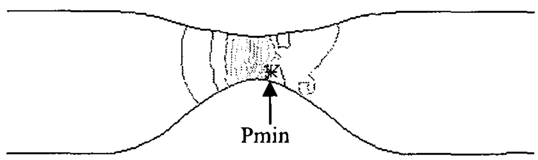

Fig.7 Contour plot of pressure distribution along the tube

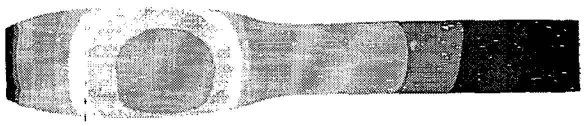

Fig.8 Contour plot of structural displacement of FSI model

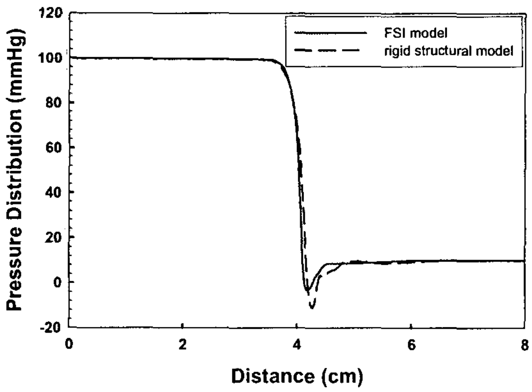

Fig.9 The pressure graph distribution ($\theta = 0^0$ line)

The shear stress distribution in our result gives the lower value compare to the averaged value of the previous investigation conducted by another author [1-2] usually from (700-1200 dyn/cm^2) which maybe because of the geometry idealization in this analysis and the mesh quality near the stenosis area. The unsmooth plotting line in the shear stress graph indicates the insufficiency of finite element mesh to capture all of the complex flow phenomena in the artery disease, such as turbulence which was indicated by the negative shear stress value.

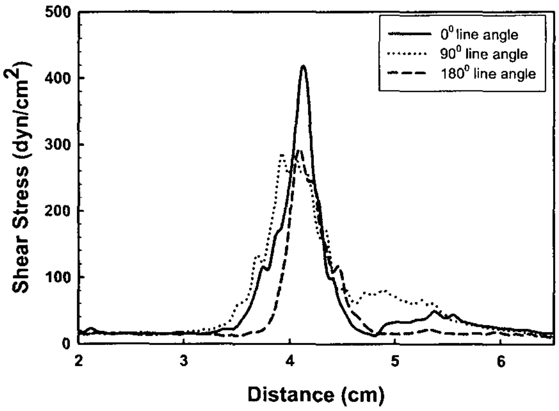

Fig.10 The shear stress distribution along line θ = 0^0, 90^0, 180^0 for the rigid structural model using 4-node tetrahedral element

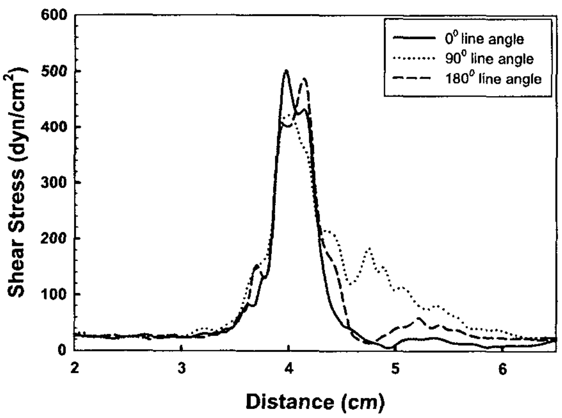

Fig.11 The shear stress distribution along line θ = 0^0, 90^0, 180^0 for the FSI analysis model

Further more, we just emphasize the important role of FSI in blood vessel analysis. Because of that reason, the focused on examining the wall compression and the collapse mechanism at the section after narrowing area was set aside.

4. Concluding Remarks

A comparison study between the rigid structural fluid model and the elastic structural model using fully-coupled FSI analysis was conducted to study the necessity of fluid structure interaction in blood vessel analysis. The future work to examine the collapse mechanism and artery compression is needed to fully understand the fluid structure interaction in the blood vessel.

Acknowledgements

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