혈관 유연성을 고려한 경동맥 분기부 모델 혈류역학 해석

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NUMERICAL ANALYSIS OF BLOOD FLOW DYNAMICS AND WALL MECHANICS IN A COMPLIANT CAROTID BIFURCATION MODEL

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Blood flow simulations in an idealized carotid bifurcation model with considering wall compliance were carried out to investigate the effect of wall elasticity on the wall shear stress and wall solid stress. Canonical waveforms of flowrates and pressure in the carotid arteries were imposed for the boundary conditions. Comparing to rigid wall model, generally, we could find an increased recirculation region at the carotid bulb and an overall reduced wall shear stress. Also, there was appreciable change of flowrate and pressure waveform in longitudinal direction. Solid and wall shear stress concentration occurs at the bifurcation apex.

Keywords: 전산유체역학(CFD), 혈류역학(Hemodynamics), 경동맥 분기부(Carotid Bifurcation), 혈관벽 유연성(Vessel Wall Compliance), 혈류유동(Blood Flow), 벽전단응력(Wall Shear Stress), 탄성도(Elasticity)

1. INTRODUCTION

Atherosclerosis is a kind of chronic cardiovascular disease in decades and affects harmfully to human being. It leads to form thrombus in the inside wall of blood vessel which are prone to narrow or occlude arteries outright. As a result, stroke and heart attack may occur. In the investigations on carotid arteries wall, mechanical stress concentration at the bifurcation was found 9 to 14 times to proximal circumferential wall stress and 3 to 4 times larger at sinus bulb [1]. In the study of Glagov et al.[2], it was also revealed that atherosclerotic plaques were built up often at the high concentrating tension area. This research yielded a significant role of these high stress sites on initiation and development of atherosclerotic lesion. The correlation between wall shear stress and mechanical stress was also reported [3]. In this study, the regions where low wall shear stress and high mechanical stress occur simultaneously have been proposed to associate with atherosclerotic plaque develop.

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In a continuing attempt to achieve the best computational fluid dynamic analysis, enormous affairs were carried out to get more realistic models, and also apply appropriate numerical methodology. Moreover, with advancement of in vivo magnetic resonance imagining angiogram (MRI), subject specific image-based CFD study have been increasingly conducted. [4-6]

Another significant aspect of research is to consider interaction between distensible wall and pulsatile blood flow using methodology of coupling scheme. In this coupling algorithm, equations of flow and wall motion are solved simultaneously. Studies of wall with distensibility have demonstrated that wall compliance is an important factor on influence of the vessel motion. A comparison of wall stress pattern and behavior of flow field between rigid and compliant wall case has shown that flow separation and recirculation reduce at the sinus bulb in distensible wall and oscillating wall shear stress 25% smaller than one in case of the rigid wall [4].

In the present study, an idealized carotid bifurcation model was used for carrying out fluid-structure interaction (FSI) simulation. A commercial finite element program, ADINA, is

employed to accurately solve the coupling equations. Thin shelled linearly elastic wall was assumed with imposing canonical waveforms of flowrates and pressure for the boundary conditions. For the purpose of comparison, a non-distensible wall model has been computed simultaneously.

2. METHODS

2.1. CAROTID BIFURCATION MODEL AND INFLOW CONDITIONS

Three-dimensional geometry of carotid bifurcation model was constructed using computer-aided design software (CATIA, Dassault systems, France) based on representative dimension of normal healthy carotid. The canonical pulsatile flow rates based on the combination of in vivo measurement data was specified at the common carotid artery(CCA) and the external carotid artery(ECA)[7-8]. A corresponding time-varying pressure condition based on Perktold et al.[4] was imposed at the internal carotid artery (ICA).

2.2. MECHANICAL PROPERTIES OF THE CAROTID

The carotid artery wall thickness was generally revealed to be 8-10% of the arterial diameter. In our research, the wall thickness is 0.6 mm and unchanged along the vessel. In addition, to make simplifying, an assumption of the shell model was applied in which the middle surface has a role of reference surface and represents the mechanical behavior of model. The strain is supposed to distribute in the thickness direction and corresponding stress distribution is achieved by analytical or numerical integration. The carotid wall was assumed to be an isotropic, linear, elastic solid with a Young's Modulus of $E = 2.6\times10^5$ N/m^2 and a Poisson's ratio $v = 0.45$.

2.2. NUMERICAL METHODS

The flow domain was divided into 170,562 computational 3-D fluid elements with 64,001 nodes and the elastic wall discretization yielded 13,414 shell elements. A fine prism mesh was generated near the wall region. The coupled model results in a set of hemodynamic data including velocity, pressure, wall shear stress in flow domain and mechanical data containing wall displacement and effective stress distribution on the arterial wall. A fully coupled FSI simulation using an Arbitrary Lagrangian-

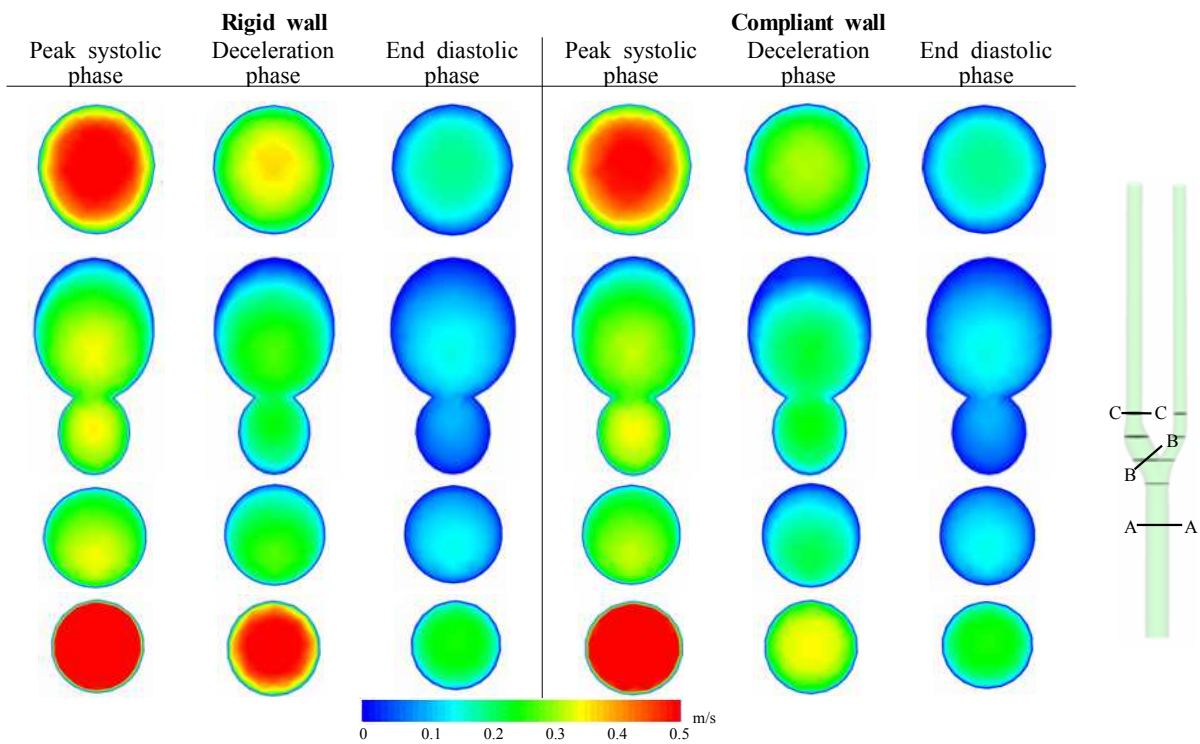

Fig. 1. Comparisons of axial velocity at four different locations

Eulerian formulation was conducted under transient flow condition.

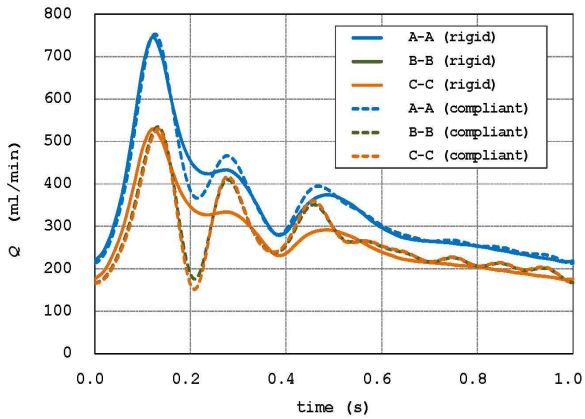

Fig. 2. Comparison of time-varying flowrate waveform

3. RESUTLS AND DISCUSSION

3.1 Axial velocity and time-varying flowrates

Axial velocity distributions at four different locations were compared between rigid and compliant wall simulations in Fig. 1. Overall, only minor difference can be seen, but in the deceleration phase, low velocity region was increased with applying wall compliance. Time-varying flowrates were computed at three locations indicated in the CCA and the ICA in Fig. 1. It was recognized the pulsatility of flowrate waveform increased in compliant wall model compared to rigid model and it was most pronounced at the deceleration phase. This was also amplified along with longitudinal direction, particularly within the

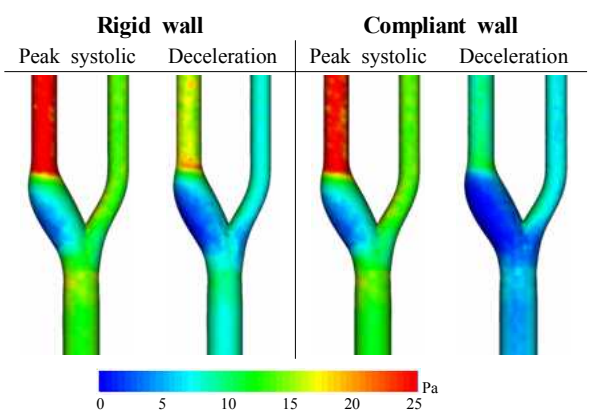

Fig. 3. Comparison of wall shear stress distributions

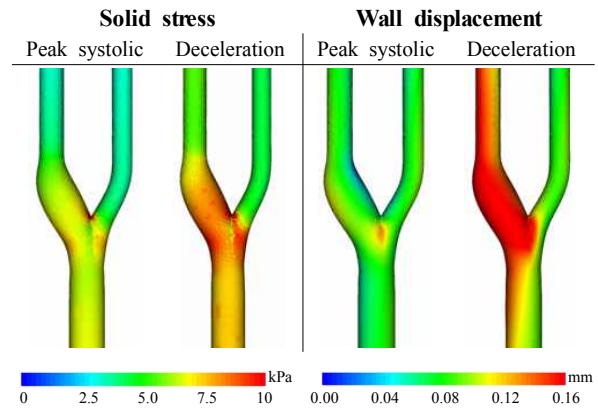

Fig. 4. Comparison of effective solid stress and wall displacement

ICA distal to bifurcation apex.

3.2 Wall shear stress

Distributions of the representative hemodynamic wall parameters, wall shear stress(WSS), which has long been implicated in biological response of vascular wall, were computed at the two different cardiac cycle phases including peak systole and deceleration phase(Fig. 3). Although only small difference was seen at the peak systolic phase, considerable difference was observed for the deceleration phase. This may be attributed to the relatively low flow rates at the deceleration phase by increased pulsatility due to wall compliance as shown in Fig. 2.

3.3 Solid stress and wall displacement

Effective solid stress and wall displacement at the corresponding cardiac cycle phases was shown in Fig. 4. Relatively higher structural stress was observed around ICA bulb with the highest value at the bifurcation apex. In addition, this highest wall displacement occurred at the deceleration phase which is actually a time point when the maximum pressure is created.

4. CONCLUSION

By conducting FSI simulations for an idealized carotid bifurcation model with elastic wall assumption, we could find an increased recirculation region at the carotid bulb and an overall reduced wall shear stress. Also, there was an appreciable change of the flowrate and pressure waveforms in longitudinal direction.

Solid and wall shear stress concentration occurs at the bifurcation apex.

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