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Computational simulation of aortic aneurysm using FSI method: Influence of blood viscosity on aneurismal dynamic behaviors

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ABSTRACT

It is well-established that blood viscosity plays a significant role in the determination of the health of the individual. It has been reported that many cardiovascular diseases are associated with blood viscosity. In this paper, the dynamic behaviors of aortic aneurysm subject to physiological blood flow with normal and high viscosities are presented. Fluid–structure interaction (FSI) method was used in the computational simulation. The influence of blood viscosity on flow dynamics within the aneurysm sac, aneurismal diameter, cross sectional shape, wall axial displacement and wall shear stress (WSS) was studied in detail. This investigation uncovered the correlations between blood viscosity and the dynamic behaviors of aortic aneurysm, which have rarely been found in existing literatures. We believe that these findings may provide important implications for individualized endovascular treatment for patient with aortic aneurysm.

1. Introduction

The bloodstream, which is known as the “river of life”, consists of complex substances, including plasma and particles, such as leukocytes, platelets and the red blood cells. It is clear that hematocrit (HCT), red blood cell (RBC) deformability and the degree of RBC aggregation are major determinants of blood viscosity. Most et al. have indicated that the reduction in blood viscosity will cause an increase in maximal myocardial oxygen delivery distal to a moderate coronary stenosis [1]. Kwon et al. have discovered that blood viscosity has significant effect on oxygen transport in artery [2]. Results of these studies show the importance of blood viscosity to the physiological functions of tissues and organs, since oxygen plays a crucial role in the metabolism of animals.

Many diseases are associated with blood viscosity. It has been reported that blood viscosity changes have been found in human cardiovascular diseases such as hypertension, spasm and thromboembolism [3–5]. Ding et al. have indicated that blood viscosity has large effect on duration of cerebral ischemia and reperfusion [6]. Wolfgang et al. have reported that there is a positive relation between the viscosity of blood plasma and the incidence of coronary heart disease [7].

Plenty of studies have focused on factors influencing blood viscosity [8–13]. However, little is known about the effect of different blood viscosity on aneurysm dynamic behaviors, including hemodynamics within aneurysm sac, forces acting on the aneurismal wall and aneurysm motion characteristics. In fact, a deep understanding of the influence of blood viscosity on aneurysm behaviors has long been considered to have great theoretical and practical importance. Wouter et al. have found through clinical experiments that there is a close relationship between blood viscosity and risk of cerebral ischemia after surgical treatment of ruptured aneurysms [14]. The clinical management of the unruptured aneurysms is also difficult, because the mechanisms underlying aneurysm formation, growth and the risks brought by the operational intervention are still not completely understood. Khanafer et al. point out that the viscous force that is closely associated with blood viscosity is one of the major determinants of aneurysm dilation and thrombus deposition [15]. Moreover, it is known that the blood viscosity affects wall shear stress (WSS), and it has been reported that WSS is strongly correlated with aneurysm formation, progression and risk of rupture [16,17]. Therefore, study on the influence of blood viscosity on the dynamic behaviors of aneurysm is crucial. Maier et al. [18] have calculated maximum values of diameter and wall displacement for several abdominal aortic aneurysms (AAAs). They have indicated that maximum wall stress and maximum RPI (rupture potential index) criteria enable a reliable rupture risk evaluation for AAAs [18]. However, their study has focused on the mechanical quantities of current AAA states at the time of CT imaging, not considering the transient processes of the aneurysm exposed to physiological environment during the cardiac cycle.

Based on the above analysis, the influence of blood viscosity on aneurysm dynamic behaviors, including blood velocity vector, strain rate distribution pattern, time variation regularity of aneurismal wall displacement and shear stress was investigated using the numerical method. The simulation results were presented and were systematical analyzed in this paper. Finite element model of the

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aneurysm was constructed. Fluid–structure interaction (FSI) effect has been taken into account in the numerical simulation in order to present results that are more realistic [19]. We believe that this study can provide a mechanical basis for biomedical research in relevant area.

2. Method

2.1. Geometric model and material property

As shown in Fig. 1, the computational domain for this study included a portion of ascending aorta, aortic arch and descending aorta with a saccular aneurysm. The inner diameter and the outer diameter of the normal aorta were 20 and 24 mm, respectively [20,21]. The diameter of the center curvature of the aortic arch was set to 80 mm, which is within the physiological range of a healthy human being [22]. Both fluid region and solid region were meshed with structured grids. The arterial wall was composed of three layers of elements. The blood was assumed as incompressible isotropic Newtonian fluid in this study, which is a reasonable assumption in the aorta [22]. The density of the blood was set to 1050 kg/m^3 [22]. The densities of the aneurismal wall and arterial wall were set to 1120 kg/m^3 [20,23]. Young's modulus and Poisson's ratio of the normal arterial wall were set to 1.08 MPa and 0.49, respectively, which are within the physiological range [20,24]. For the aneurysm wall, Young's modulus was assumed to be 4.5 MPa, and Poisson's ratio was set to 0.45 according to relevant literature [23]. The kinematic viscosities of the blood were set to 0.0027 Pa s(μ_1) and 0.0097 Pa s(μ_2) for the comparative study [25].

2.2. Numerical model for FSI simulation and input conditions

The geometry and grids of the aneurysm model were generated using CFD-GEOM. Then the geometrical model was imported into ANSYS software package which is considered as a suitable platform for solving the blood–fluid interaction problem by previous authors [26–28]. The key toward the solution of the coupled fluid structure problem is the proper numerical technique among the fluid and solid parts [29]. The arbitrary Lagrangian–Eulerian (ALE) algorithm or method, which was first proposed by Hirt et al. in 1974 [30], provides an effective way of bring into relation the Lagrangian coordinate system usually used in solving solid problem with the Eulerian coordinate system usually used in solving fluid problem. We used ANSYS structure code to deal with the solid problem. The transient deforming fluid domain of the aneurysm model was solved by CFX code, which is based on ALE algorithm. During the calculation, corresponding physical quantities transferred across fluid–structure interfaces through the coupling of the two sets of codes until convergence criterion (10^{-4}) was reached for each time step. Minimum residual was 10^{-18}. A constant time step of 0.005 s was employed in this study. Second order central difference scheme was used. During the simulation, element grids of the structural wall were auto-remeshed to satisfy the boundary condition at blood–wall interfaces. Three cycles were required before the periodical solutions were obtained. Governing equations used in the program for solving fluid and solid domains are presented as follows:

Fluid continuity equation

$$\frac{\partial u_i}{\partial x_i} = 0 \tag{1}$$

Fluid momentum equation

$$\frac{\partial u_i}{\partial t} + u_j \frac{\partial u_i}{\partial x_j} = -\frac{1}{\rho_f}\frac{\partial p}{\partial x_i} + \frac{1}{\rho_f}\frac{\partial \tau_{ij}}{\partial x_j} \quad \text{in } {}^F\Omega(t) \tag{2}$$

f and p stand for fluid and structure wall, respectively. u_i is the velocity vector, p is fluid pressure, ρ_f is fluid density, τ_{ij} is fluid stress tensor, ${}^F\Omega(t)$ is the moving spatial domain upon which the fluid is described.

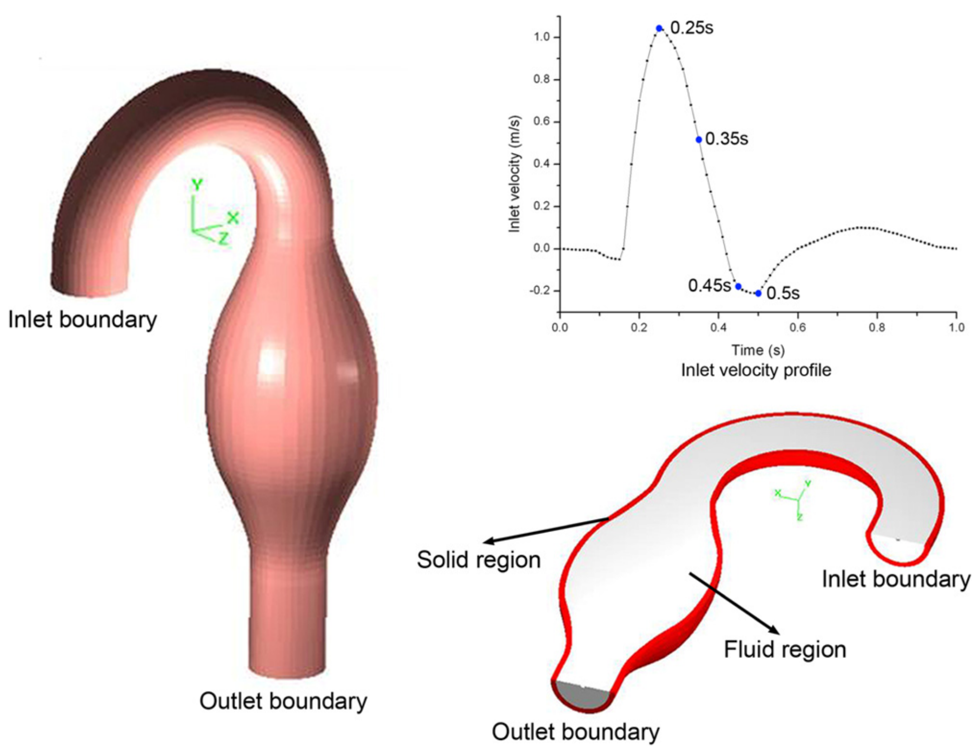

Fig. 1. Geometry and boundary condition.