Pressure variation after stent intervention for a giant aneurysm complicated by a stenosis

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Abstract

This paper investigates the pressure variation in a stented aneurysm complicated by a stenosis. Computational Fluid Dynamics analyses were performed on a giant aneurysm with and without a stenosis. Three models were constructed to compare the pressure variation. Model M1 is the one with an aneurysm (no stenosis and no stent), Model M2 is the one with a preaneurysm moderate stenosis (50%), and Model M3 is the one with an aneurysm implanted with a stent. Pressure increase in the aneurismal sac caused by the stenosis is about 10.3 mmHg at peak systole (comparison between M2 and M1). The pressure increase in the aneurismal sac is about 7.8 mmHg at peak systole (comparison between M3 and M2). The geometry of the parent vessel and its aneurysmal/stenotic disease do have influence on pressure variation.

Keywords: Aneurysm, Stenosis, Hemodynamics, Computational Fluid Dynamics

1 INTRODUCTION

Four articles published [1-4] in "American Journal of Neuroradiology" raised a drastic debate about that how much is the pressure increase in a segment of aneurysm after stent intervention treatment for an aneurysm complicated by a proximal stenosis. The study [1] demonstrated a 20mm Hg increase of pressure (the first model) in intra-aneurysmal sac after treatment. However, Fiorella thinks that the pressure increase was overestimated by at least a factor of 2 compared with traditional fluid mechanics calculations and experimental measurements[5,6]. Flow resistance will decrease according to the Poiseuille law when a 50% stenosis of a vessel is removed. But the Poiseuille law can only be used for straight tubes. The geometry of the model used by Cebral [1] is tortuous and it cannot be simply solved by the Poiseuille law. So more work needs to be done to explore this problem.

The purpose of this paper is to perform a numerical simulation study by using Computational Fluid Dynamics (CFD) approach and investigate whether a sharp increase of pressure occurs in an intra-aneurysmal sac because of the existence of stenosis after the stent is inserted to the aneurismal region.

2 METHODOLOGY

For the sake of comparison, the internal carotid aneurysm model (M1) without stent in the reference [7] is used in this study. Based on M1, a segmental vessel with moderate stenosis

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about 50% was artificially constructed. Stenosis locates at the proximal neck of the aneurysm in M1. In this way, the second model (M2) was constructed. The third model (M3) is equivalent to the implantation of a stent into M2 and the stenosis in M2 was expanded. These three models are shown in Fig.1.

Figure.1 Aneurismal models (from the left to the right: M1, M2, M3). Lower end is the inlet, and the upper end is the outlet.

The results obtained from the papers [8] and [9] were used to judge the reliability of the present numerical calculation. So numerical simulation was carried out based on the model used in reference [8,9] firstly. The model is two dimensional in the reference [8,9], while the model in the present study is three dimensional. The diameter at the inlet, the diameter at the outlet and the longitudinal length of arterial vessel are 0.38cm, 0.36cm and 5.2cm respectively.

Validation between their experiment and our CFD calculation was carried out. In this indirect way, the reliability of numerical simulation used in this study was tested. Hemodynamic modeling used in this paper is the same as that in the references [8,9]. Comparison of the pressure drop between the measured values and calculated CFD results was shown in Fig.2. The average values of the pressure drop measured from the experiment and obtained from this study are 0.59 mmHg and 0.69 mmHg respectively in a cardiac cycle. The relative error between the experiment in the references [8,9] and this numerical calculation is less than 15%, which means that the method for the CFD calculation used in this study is reliable. The maximum of the pressure drop, about 5.6 mmHg, occurs at times between 0.08s and 0.09s in the present study. The maximum of this pressure drop, 4.5 mm Hg, occurs at 0.11s in the experiment [8,9]. There is a phase difference about 0.02s between the two studies. The main reason is that the vessel wall is assumed to be rigid in the present study and the wall is elastic in the experiment [8,9].

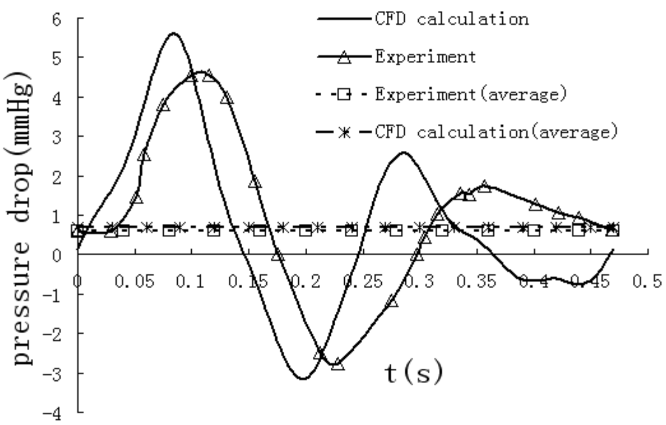

Figure.2 Temporal variation of pressure drop between inlet and outlet in a cardiac cycle

Blood flow is controlled by the three-dimensional incompressible Navier-Stokes equations. The viscosity and density of blood are 0.04 Poise and 1.0 g/cm^3 respectively. Compared with the large aorta, the elasticity in cerebral arteries is small. Therefore vascular wall was assumed to be rigid. No-slip boundary conditions were applied to the walls. Studies using Newtonian fluid and non-Newtonian fluid on the same cerebral aneurysm models find that the main flow features are not greatly affected by the viscosity model[10], so blood is assumed to be a Newtonian fluid. The maximum of Reynolds number based on the entrance flow velocity and the diameter at the inlet is 704, so the flow is laminar.

As noted in the reference [1,2], the blood flow rate was not measured from the patient from which the model M1 was constructed. So the blood flow rate in the reference [11] was used in the present study. Velocity at the internal carotid artery in different individuals changes greatly. According to the clinical medical statistics data from the literatures [12,13], the velocity range at the internal carotid artery in normal individual is from 20.1 cm/s to 112 cm/s. This range is from 24.1 cm/s to 209.8 cm/s when the aortic stenosis is existed and the degree of stenosis is 10-55%. The maximum velocity is 58.7 cm/s at the inlet in a cardiac cycle in the three models used in the present study and the average velocity is 38 cm/s. Blood pressure was set to 0Pa at the output. The same boundary conditions were used in the 3 models. Based on the consideration of computational accuracy and computational efficiency, the time step of 0.005s was selected for this study. The discrete form of the differential governing equations of blood flow follows the upwind scheme with second order accuracy. ANSYS CFX 12.1 was employed to perform the numerical simulation of Computational Fluid Dynamics (CFD). The solution approach of implicit finite volumes was used. Parallel calculation was used to reduce the running time. The residual convergence criteria of mass and momentum were set to 10^{-5}. The CFD simulations were carried out for 3 cardiac cycles. Results in the third cycle were analyzed.

3 RESULTS AND CONCLUSIONS

As shown in Fig.3, a plane which locates on aneurismal region was assigned to calculate the pressure drop between this plane and the outlet (also between inlet and outlet). Plane positions of inlet, preaneurysm plane and outlet are same for the 3 models. To clearly show the position of this defined plane, streamlines were drawn in Fig.3. The plane of preaneurysm is perpendicular to the centerline of artery model. The average value of pressure on a plane (inlet, plane of preaneurysm, outlet) was calculated at different moments in a cardiac cycle. The average value of pressure on a plane (inlet, plane of preaneurysm, outlet) in a cardiac cycle was calculated too. The results are presented in Table 1.

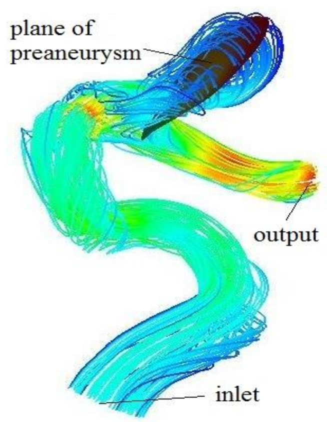

Figure.3 Positions of plane for pressure drop measurement

Table 1: Pressure drop

Model	A[a] (mmHg)	B[b] (mmHg)
M1	5.1	10.9
M2	9.4	21.2
M3	6.4	13.4

a. A indicates “The average of the pressure drop between the preaneurysm and inlet in a cardiac cycle”.
b. B indicates “The pressure drop in peak systole between the preaneurysm and inlet”.

Comparing M1 with M2, the intra-aneurysmal pressure at peak systole increases about 10.3 mmHg. Pressure drop between the inlet and the preaneurysm plane in a cardiac cycle is calculated for the 3 models and their changes with time are shown in Fig.4. The pressure drop for the M2 is the biggest, this value is moderate for the M3, and it is the minimal for the M1. Because pressure change on aneurismal wall before and after stent implantation is our concern, the difference of the pressure drop (preaneurysm plane and inlet) between M2 and M3 is calculated.

The model used in the references [8,9] is a straight tube with a mild taper. The vessel diameters at the inlet and the outlet are 0.38 cm and 0.36 cm respectively. The axial distance between the inlet and the outlet is 5.2 cm. The maximum velocity at the inlet is 49 cm/s. Pressure drop measured by Banerjee and Back at the cyclic peak is approximately 4.3 mmHg [8,9]. The axial distance along the centerline in the 3 models in this study is also 5.2 cm. It is the same with the model in the references [8,9]. The nominal mean inlet diameter in the models M1 M2 M3 is 0.56 cm. It is bigger than the model used in the references [8,9]. The maximum velocity at the inlet in this study is 46 cm/s. According to the Poiseuilles law for blood flow, pressure difference is related to the length of vascular, the dynamic viscosity of blood, the volumetric flow rate and the radius of the vascular vessel. If the Poiseuilles law could be applied to a patient-specific model, the pressure difference calculated from the model M2 in the present study will not be 21 mmHg. It should be about 3.52 mmHg. This shows that Poiseuilles law cannot be simply applied to a patient-specific model of vascular vessels with geometric characteristics of bent, taper and aneurysm. It also shows that flow resistance increases more than six folds in the model M2 used in the present study compared with the model with a straight tube used in the references [8,9].

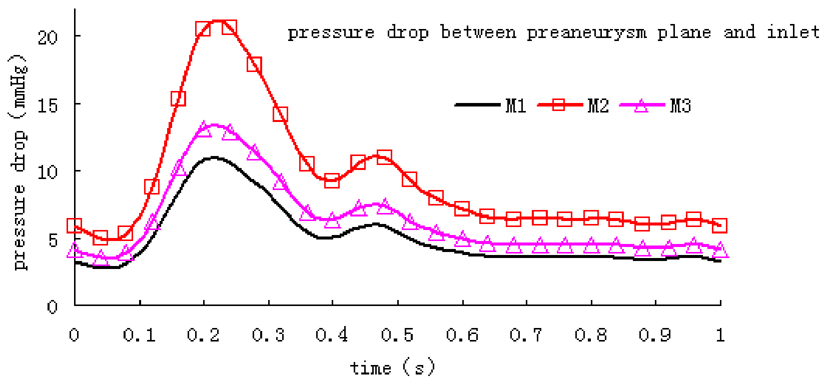

Figure.4 Temporal variation of pressure drop between inlet and preaneurysm plane in a cardiac cycle

The present study demonstrates that a moderate stenosis cannot result in a sharp pressure increase when stent intervention is applied to the treatment of an aneurysm complicated by a 50% stenosis harbored on a tortuous intracranial artery. The geometry of the parent vessel and its aneurysmal/stenotic disease do have influence on pressure variation.

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