Computer Simulation of Flow Dynamics in Paraclinoidal Aneurysms

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Key words: computer simulation, flow dynamics, brain aneurysms

Summary

Endovascular treatment, which is very useful method especially for paraclinoidal aneurysms, has the limitations of coil compaction and recanalization, which are difficult to predict. We tried to understand flow dynamic features, one of the important factors of such problems, using computer flow dynamics (CFD) simulations. CFD simulations were made in paraclinoidal aneurysm model of different size and protruded directions. Flow patterns, flow velocities and pressure are analyzed. Although the pressure on the aneurismal orifice is highest in the aneurysm protruding vertically-upward, the flow velocity is highest in the superior-medial protruding one. Significant difference is not observed in either flow patterns, flow velocities or pressures on the aneurismal orifices between the sizes of aneurismal sac. Among paraclinoidal aneurysms, an aneurysm protruding to superior-medially receives the most severe haemodynamic stresses at the orifice and the aneurysm size does not cause significant differences in the aspect of flow dynamics. It should be considered in the treatment of such aneurysms.

Introduction

We have two treatment options for cerebral aneurysms, surgical clipping and endovascular embolization. Although the selection of these two treatment options is still controversial, the

latter is a valid technique especially for paraclinoidal aneurysms^{1,2,3} which are problematic for a safe surgical approach^{4,5}. One of the important limitations of the endovascular aneurysmal embolization is recanalization due to coil compaction, especially in cases of large or giant aneurysms^{6,7}. It is very important to understand the flow dynamics when treating cerebral aneurysms. Nowadays, improved computer technology and diagnostic equipment can provide haemodynamic information for individual cases using computational flow dynamic (CFD) simulations^{8,9,10}. However, individual flow simulation is hindered by the limitations of time and facilities. Therefore, at present in clinical settings, it is still practical to understand flow dynamic features classified according to basic anatomical patterns.

To date, many flow dynamic studies has been reported⁸⁻¹⁶, in most of which the vessels are considered as rigid tubes because of the difficulty in mimicking vessel wall elasticity in *in vitro* studies. The vessel wall exhibits a pulsatile motion, however, according to the cardiac cycle, and the motion is derived from the pulsation of the blood flow. This pulsatile motion should have some effect on blood flow dynamics.

In the present study, we examined the flow dynamic pattern of paraclinoidal aneurysms in relation to the size and direction of protrusion using computer flow simulations in three dimensions under the pulsatile motion of the vessel wall.

Figure 1 The aneurysm is 12 mm (upper) and 6 mm (lower) in diameter. They protrude in 4 different directions: vertically upward (0°), superior-medial (45°), inferior-medial (135°), and vertically downward (180°).

Methods

A model of the internal carotid artery (ICA) was made. The diameter of the ICA was 5 mm and the carotid siphon was a half circle 10 mm in radius, and the C2 and the C4 segments were parallel. The aneurysm was located on the distal part of the carotid siphon. Small branches, such as the ophthalmic artery or posterior communicating artery, and the distal part of the ICA were omitted for simplicity. The aneurysms were 6 mm and 12 mm in diameter, respectively. The orifices were round and 4 mm in diameter in all models. They protruded directionally as follows: vertically upward (0°), supe-

rior-medial (45°), inferior-medial (135°) and vertically downward (180°), respectively (figure 1).

At the inlet, the blood inflow velocity was based on the blood flow pattern in MCA measured by trans-cranial Doppler (figure 2), with a maximum of 0.91 m/sec and a minimum of 0.46 m/sec. The maximum velocity was seen at 0.22 sec per pulse cycle. The pulse rate was assumed to be one cycle per second for the sake of convenience. The inlet was immediately adjacent to the ICA. The blood was assumed to behave as a Newtonian fluid. Its density and viscosity was 1060 kg/m³ and 0.004 Pa.sec, respectively.

The simulations in this study were performed using the CFD (computational fluid dynamics) software package RFLOW (Rflow Co., Ltd., Saitama, Japan), which solves the incompressible and unsteady Navier-Stokes equations and continuity equation using the finite volume approach. No turbulence model was used in this study because the blood flow is of low Reynolds numbers and hence laminar, as described later. The governing equations are as follows:

$$\rho(\partial \mathbf{v} / \partial t + (\mathbf{v} \cdot \nabla) \mathbf{v}) = -\nabla p + \mu \nabla^2 \mathbf{v} \quad \text{---- (1)}$$

$$\nabla \cdot \mathbf{v} = 0 \quad \text{---- (2)}$$

where ρ is the density of blood, $\mathbf{v}$ is the flow velocity vector, p is the pressure and μ is the viscosity of blood. The convection and diffusion terms are discretized by the 2nd-order accurate TVD (total variation diminishing) and central difference schemes separately.

A non-orthogonal BFC (boundary-fitted curvilinear coordinate) and block-structured grid were generated for the computational model. The dilation and shrinkage of arteries and aneurysm were considered in this study. Therefore two sets of grids were generated at the smallest and largest sizes of aneurysm and arteries, which correspond to the minimal and maximal inflow velocities at the inlet, respectively. At the time of maximal inflow velocity, the aneurysm was dilated to 1.2 times and the arteries to 1.1 times in diameter (1.44 times and 1.21 times in the cross sectional area) compared to the sizes at the time of minimal inflow velocity (figure 3). The deformation of the grid was performed between two sets of grids with the variations in diameters of aneurysm and arteries being proportional to blood inflow velocity at the inlet. The dilation and shrinkage of arteries and aneurysm were repeated periodically, similar to the previous study¹⁷, which reported the pulsatile motion of intracranial aneurysms and

Figure 2 Simulated flow velocity at the inlet. Maximum and minimum velocity are 0.91 and 0.46 m/s, respectively. Maximum velocity comes at 0.22 sec in the pulse cycle.

parent arteries measured as the mean 1.53- and 1.20- fold increase “expansibility” in the cross-sectional area by echogram. These are the same method as we reported the differences of flow dynamical feature between rigid and pulsatile models before¹⁸.

The flow pattern (flow velocity vectors, m/s) and pressure (Pa) in the aneurysm orifice are analyzed and discussed in the following section.

Results

The Reynolds and Womersley numbers in the parent artery was 610 – 1306 and 6.45 – 7.10, respectively. The Dean number of the carotid siphon was 152.5-331.5. They were not constant due to the pulsatile flow and motion of the vessel wall.

In the parent artery, the flow pattern formed a large secondary flow, not simply a Hagen-Poiseuille flow, due to the curvature of the carotid siphon (figure 4).

The pressure and flow velocity of the orifice were considered in each case. In the vertically-upward cases, relatively high pressure was observed in the entire area as well as the distal end of the orifice, and the inflow zone was large. However, the flow velocity was not high (figure 5). In the superior-medial cases, the highest pressure zone was limited in the distal part of the orifice, and the pressure gradually decreased to ward the lower part, which formed a “pressure gradient”. The inflow zone was large, similar to the vertically-upward cases, and the inflow velocity was much higher

Figure 3 The aneurysm and parent vessel at minimum (A) and maximum (B) flow velocity are shown. The aneurysm and parent vessel have pulsatile motion in proportion to flow velocity. The aneurysm expands 1.44 times and arteries expand 1.21 times in the cross sectional area at maximum blood flow.

Table 1 Summary of the results.

	direction of aneurysm protrusion			
	0°	45°	135°	180°
inflow zone	large	large	small	small
inflow velocity	middle	high	high	small
outflow velocity	middle	high	middle	small
pressure	high	high	small	small

(figure 6). In the inferior-medial cases, the pressure at the orifice was low. Although the inflow velocity was high, the inflow zone was much smaller than the former cases (figure 7). In the vertically-downward cases, the pressure, inflow velocity, and inflow zone were all small (figure 8). The results were summarized in table 1.

Figure 4 These cross sectional planes at the orifice of aneurysm show marked secondary flow generated from the curvature of the carotid siphon in any of the protruding directions of aneurysm (A-D). The theme of the secondary flow patterns is shown (E).

Figure 5 Inflow / outflow patterns and pressure at the orifice are shown. The left side is the proximal end and the right side is the distal end of the orifice of the aneurysm. The phases in the cardiac cycle are early systolic (0.11 sec), maximum systolic (0.22 sec; maximum blood flow), early diastolic (0.44 sec), and late diastolic (0.88 sec), respectively, upper to the lower. In the models of vertically upward protrusion, relatively high pressure is seen in the whole area of the aneurysm, but inflow velocity is not so high, and the inflow zone is large.

Figure 6 In the models of superior-medial protrusion, relatively high pressure stresses in a limited area around distal end of the orifice. The velocities of the inflow and outflow are high and the inflow zone is large.

Figure 7 In the models of inferior-medial protrusion, the pressure at the orifice is considerably lower. Though the inflow velocity is still high, the inflow zone is much smaller.

No significant differences in the haemodynamic factors were observed between the aneurysmal sizes, 6 mm vs. 12 mm in diameter.

Discussion

In this study, all simulations were performed under the pulsatile condition of the vessel wall as well as the flow pattern. Although it is natural for an elastic vessel wall to exhibit pulsatile motion according to the pulse cycle, this has been one of the barriers to *in-vitro* flow dynamic analysis because of its complexity. However, the development of computer technology has enabled serial animation imaging and real-time haemodynamic simulations. Although the elastic behavior of a vessel wall is not monotonous, experimentally, aneurysms expand proportionally to blood pressure in the normal range¹⁹, and blood flow is derived from that pressure. Thus, the vessels in our study are assumed to expand and contract in proportion to the blood flow. Although some previous studies reported the effects from vessel wall pulsation were not highly significant^{20,21}, the vessel wall does move according to the cardiac cycle and therefore we believe the flow dynamic analysis under pulsatile motion of the vessels as more similar to the clinical situation, especially considering the wall pressure as we reported before¹⁸.

Although endovascular treatment modalities are often preferred for paraclinoidal aneurysms¹⁻³, coil compaction and recanalization are sometimes experienced^{1,2}, especially in cases of large or giant aneurysms¹. The haemodynamic factors in the vessels are given as shear stress, pressure,

Figure 8 In the models of vertically inferior protrusion, the pressure at the orifice and the inflow velocity are both small. The inflow zone is small as well.

inflow rate, and flow impingement forces²². However, the immediate cause for the coil compaction is controversial. According to one report, coil compaction occurred in the part with the highest speed inflow⁹ and clinical experience that flow alteration from parent artery occlusion is an effective treatment modality for the giant aneurysms^{23,24,25}, it may be reasonable that coil compaction is mainly caused by the flow impingement force rather than simply by the wall pressure.

In our study, the highest inflow / outflow and large inflow zone of the orifice were seen in the aneurysm protruding to the superior-medial, not the vertical-upward one. The reasons for this difference are the pressure gradient on the aneurysm orifice and the secondary flow from the bends of carotid siphon. With the 6 mm aneurysm models, although the pressure at the aneurysm orifice is highest in the vertical-upward cases the difference between the highest and lowest pressure was greater in the superior medial case at the time of maximum velocity (the difference between highest and lowest pressure is vertical-upward = 51.5 Pa vs. superior-medial = 103.8 Pa, respectively) and diastolic phase (19.5 Pa vs. 39.1 Pa). Of the 12 mm aneurysm models, this difference is similar between these two protrusions at the time of the systolic phase (79.6 Pa vs. 80.8 Pa). However, at the diastolic phase, the difference is more apparent in the superior-medial case (22.4 Pa vs. 38.4 Pa) (figure 9). Considering these findings, a requirement for the condition of high inflow / outflow may be the existence of a pressure gradient at the orifice in any phase of cardiac cycle. In the vertical-upward cases the inflow / outflow might be obstructed by the small pressure gradient in the diastolic phase, whereas, in the superior-medial cases, the inflow and outflow zone could be kept constant from the superior medial part to the inferior part of the orifice at any phase. Comparing the inferior-medial and vertical-downward cases, the former had a higher speed inflow because a part of the orifice is adjacent to the area of the high speed flow in the parent artery which originated from a secondary flow as well as the pressure gradient in the systolic phase, although the

Figure 9 The contour of the orifice of the aneurysm is marked by 44 dots. The proximal end of the orifice is Nos.39, 40 and the distal end is by Nos. 17, 18, respectively (A). The pressure gradient at the orifice at the maximum systolic phase is large in both 6 mm (B) and 12 mm (C) aneurysm models. Note this pressure gradient is remained most in the late diastolic phase in the superior-medial protrusion models.

Figure 10 No remarkable differences are seen between 6 mm and 12 mm aneurysm models in relation to the pressure on the orifice.

mean pressure of the former is lower than that of the latter. However, the pressure of the orifice was not so high that the inflow zone was small and limited to only a distal part of the orifice.

The flow impingement force should be influenced not only by flow speed but flow volume, i.e., the inflow zone in this study. From these results, considering that haemodynamic stresses cause coil compaction, they are most likely to occur in the aneurysms protruding to the superior-medial, not to the vertical-upward.

In this study, the flow dynamic patterns did not significantly differ between aneurysmal sizes, 6 mm and 12 mm in diameter in each direction of their protrusion (figure 10). In large aneurysms, the flow in the dome has been featured as a stagnated area or secondary vortex¹⁴. In the present study, a stagnated area was also seen in the larger ones even in the superior-medial type of highest inflow / outflow. However, the haemodynamic features at the orifice did not differ greatly between the sizes of aneurysmal dome. The flow dynamic pattern, i.e., haemodynamic stresses in the aneurysmal orifice, should be determined not from the sizes of the aneurysmal sac but from its protruding directions and the

diameter of the orifice. Therefore, the reason for the high frequency of coil compaction in larger aneurysms^{6,7} can not be determined from the aspects of haemodynamic stresses in this study. It may be due to the difficulty in obtaining the good volume embolization ratios in cases of large or giant aneurysms^{26,27}.

Conclusions

Among paraclinoidal aneurysms, an aneurysm protruding to superior-medially receives the most severe haemodynamic stresses at the orifice. The haemodynamic pattern is mostly determined from the direction of its protrusion, not from the size of the aneurysmal sac. Therefore, in the clinical setting, it is very important to consider the differences in the haemodynamic patterns in relation to the protrusion direction when deciding endovascular treatment.

The success rate of the endovascular treatment for large aneurysms may be improved with increased understanding of the haemodynamic features of aneurysms and the development of a new generation of detachable coils such as hydrocoils, which allow a more dense packing of aneurysm.

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EDITORIAL COMMENT

The authors report the topography and direction of aneurysms in relation to strict flow conditions. Such analysis although not necessarily pertinent for some aneurysms is certainly of great interest in so called flow related aneurysms (associated to brain AVM). If in most other etiologies the nature of the vessel wall by itself is certainly impaired and inappropriately responding to normal haemodynamic conditions, therefore flow related aneurysms in brain AVM are peculiar. This later type of aneurysms is thought to be generated by the increased diastolic flow in relation to the lack of resistance at the level of the shunt. In addition the topography of the aneurysms associated to brain AVM is different from the topography of saccular aneurysms in the same age group. This suggests that the responding capacities and remodeling responses may be different in different areas of the arterial tree expressing therefore the classically recognized geotropism of certain diseases for certain arterial segments, following haemodynamic stresses.*

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