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Numerical Study on Aneurysmal Blood Flow After Coil Embolization

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ABSTRACT

Aneurysm embolization method using coils has been widely used. When partial blocking of an aneurysm is inevitable, the locations of coils are important since they change the flow patterns inside the aneurysm, which affect the embolization process. We calculated the flow fields inside the partially blocked lateral aneurysm models for different coil locations- proximal neck, distal neck, proximal dome and distal dome. Flow into the aneurysm sac was significantly reduced in the distally blocked models, and coils at the distal neck blocked inflow more effectively comparing to those at the distal dome. This study suggests that the distal neck should be the most effective location for aneurysm embolization.

Key Words: Lateral aneurysm, Coil Embolization, Hemodynamics, CFD

1. Introduction

Aneurysm embolization using coils has been used to treat cerebral aneurysms recently. A small catheter is introduced from either femoral or carotid artery and advanced to the aneurysm sac. Micro coils are inserted via a catheter and packed inside the aneurysm sac, which induce blood stagnation and thrombosis. Various factors, such as blood elements, enzymes, arterial wall function and blood flow characteristics, have been involved in thrombus formation and embolization process. Among these factors, blood flow has been thought to be an important environmental parameter. Blood stagnation promotes thrombus formation, and velocity fields affect blood element transportation. Wall shear stress has been also thought to be as an important hemodynamic factor influencing arterial wall function.

Various studies have been performed on hemodynamic changes in aneurysms after coil embolization[1]. It is hard to fill the aneurysm sac completely with coils in case of aneurysms with wide necks and giant aneurysms, and aneurysm regrowth and rupture have been reported for partially blocked aneurysms[2]. When an aneurysm is partially filled with coils, blood flow characteristics and aneurysm embolization processes are affected by the locations of coils. Also the parent vessel shape affects the flow pattern inside the aneurysm for lateral aneurysms. Therefore the shape of a parent vessel may affect the effective coil locations for embolization. We want to study the blood field changes inside the partially blocked aneurysms for different coil locations using numerical analysis in order to suggest effective coil locations for aneurysm embolization. The effects of the parent vessel geometry are also explored.

2. Methods

2.1 Aneurysm Models

Since aneurysms were frequently formed at the lateral sides of curved vessels and at the terminal sites of branches, we selected a lateral aneurysm model occurred laterally on a carotid artery. The size and shape of aneurysms and carotid arteries were different for individuals, and we chose the average values of

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aneurysm and carotid artery geometries which were measured and statistically analyzed using angiograms of patients[3]. The aneurysm sac we selected had a pear shape. Neck width was 3.1 mm, long axis diameter was 5.3 mm, height was 5.3 mm, and parent vessel diameter was 3.6 mm. In order to study the effects of parent vessel geometry on flow fields, we constructed two models with different parent vessel curvatures – one was a lateral aneurysm model with a straight parent vessel and the other was one with a curved parent vessel. The ratio of parent vessel diameter to radius of curvature was 0.17. Inserted coils were modeled as spheres whose diameter was 0.75 times the aneurysm long axis diameter. Four different coil locations were modeled – proximal neck, distal neck, proximal dome and distal dome.

2.2 Numerical Analysis

Three dimensional incompressible laminar flow fields were obtained by solving continuity and Navier-Stokes equations computationally. Numerical analysis was performed using a commercially available CFD package, Fluent 6.0 (Fluent Inc.), which was based on the finite volume methods. We constructed lateral aneurysm models with a straight and a curved parent vessel for four different coil locations. Each model was divided into 30,500 and 26,500 tetrahedral meshes, respectively. Arterial blood flow has a pulsatile flow waveform, therefore unsteady flow fields were calculated. The pulsatile flow waveform used in this study is shown in Fig. 1. The maximum and mean of Reynolds numbers ($Re = \frac{Vd}{\upsilon}$, V: Area mean velocity, d: parent vessel diameter, ν: kinematic viscosity) based on area average velocities in a parent vessel was 900 and 200, respectively, and Womersley number ($\alpha = r\sqrt{\frac{\omega}{\upsilon}}$, r: parent vessel radius, ω:angular velocity) was 7.5. These are typical values of non-dimensional parameters occurred at common carotid arteries. A period of a flow cycle was divided into 33 intervals, and parabolic inlet velocity profiles were updated at each interval. Pressure boundary condition was given to the outlet of a model. Twenty iterations were performed at each interval, and calculations were performed for three periods. Convergence was checked. We assumed the blood vessel had a rigid wall because cerebral arteries were less elastic than other arteries[4].

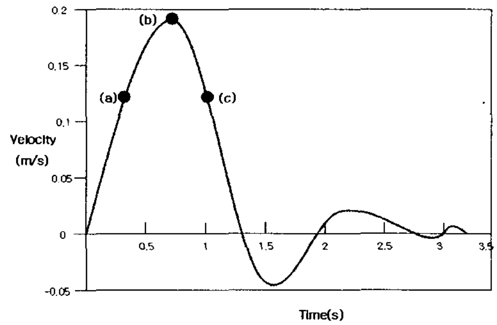

Fig. 1 Pulsatile flow waveform. (a),(b),(c) denote acceleration, peak and deceleration phase, respectively.

Blood was assumed as a Newtonian fluid since blood flow in large arteries showed negligible non-Newtonian viscosity characteristics[5].

3. Results

3.1 Straight Parent Vessel

Blood was introduced from the proximal neck during an acceleration phase. At an acceleration phase, blood introduced from the side wall of coils moved out to the distal neck in the proximal neck blocked model, and blood drawn from the proximal neck moved out via the side wall of a coil in the distal neck blocked model. In case of the dome blocked models, most of inflow drawn from the proximal neck moved out to the distal neck while some inflow formed a recirculation zone inside the aneurysm sac. As the flow accelerated, inflow from the distal neck was increased in the proximal neck blocked model while inflow from the side wall of a coil moved to the proximal neck in the distal neck blocked model. In the dome blocked models, inflow from the distal neck formed a large vortex inside the aneurysm sac and small portion of inflow moved out through the side wall of a coil to the proximal neck during a peak phase (Fig. 2). During a deceleration phase, strong inflow from the distal neck was observed in the proximal neck blocked model while inflow drawn along the side wall of a coil moved out to the proximal neck in the distal neck blocked model. A large vortex formed during athe cceleration phase persisted during the deceleration phase in the dome blocked models. Inflow from the distal neck

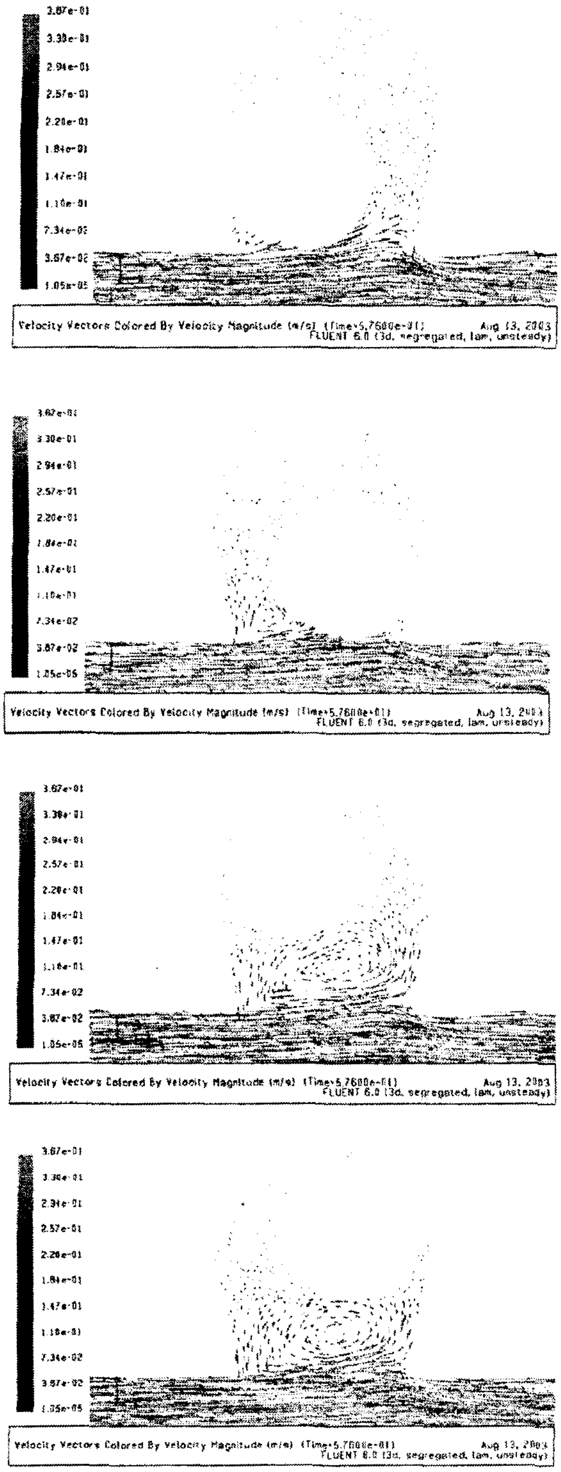

Fig. 2 Velocity fields in the midplane of the aneurysm models with a straight parent vessel at peak phase.

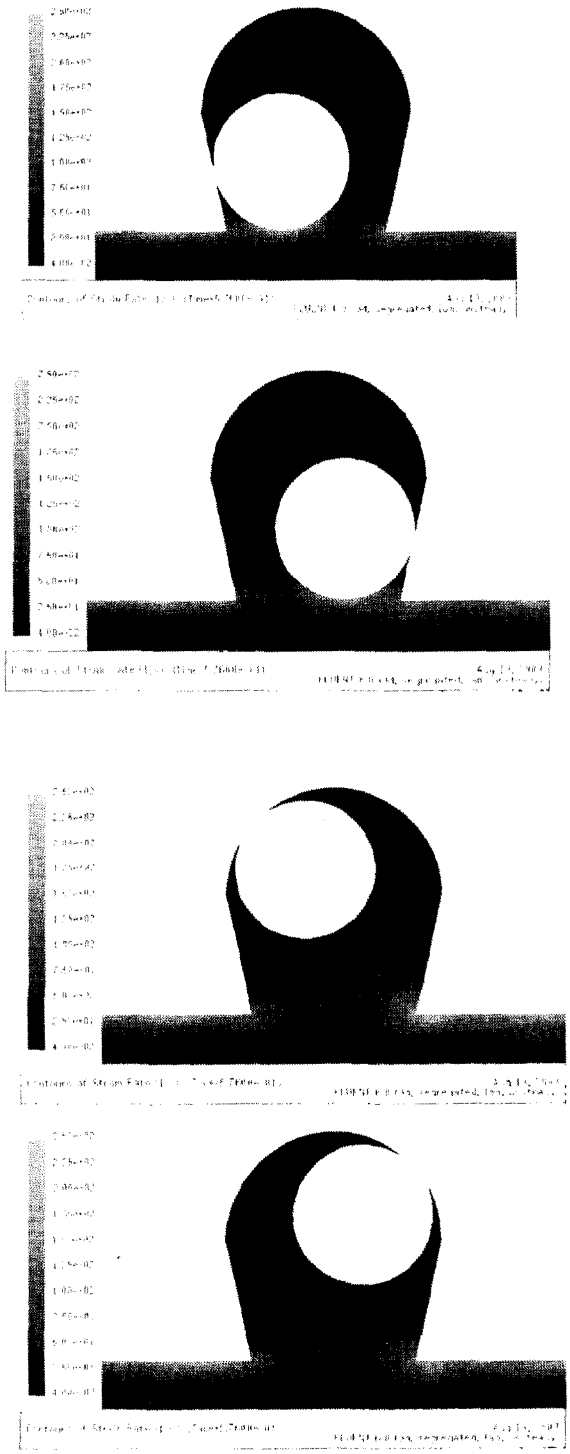

Fig. 3 Shear stress distribution in the midplane of the aneurysm models with a straight parent vessel at peak phase.

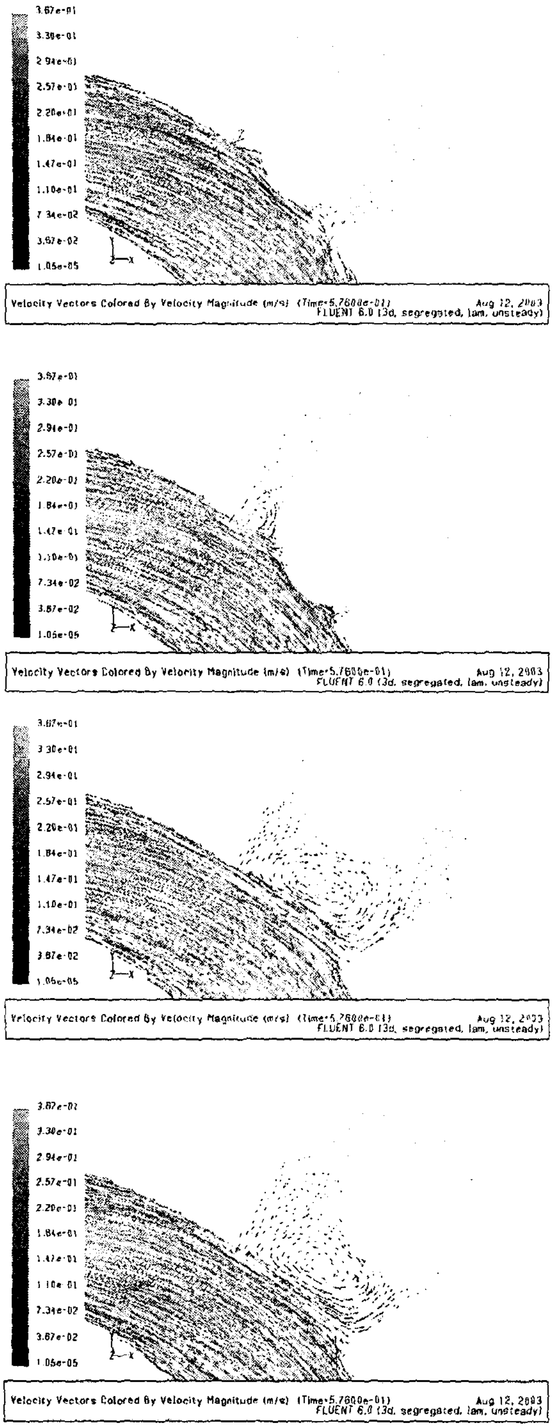

Fig. 4 Velocity fields in the midplane of the aneurysm models with a curved parent vessel at peak phase

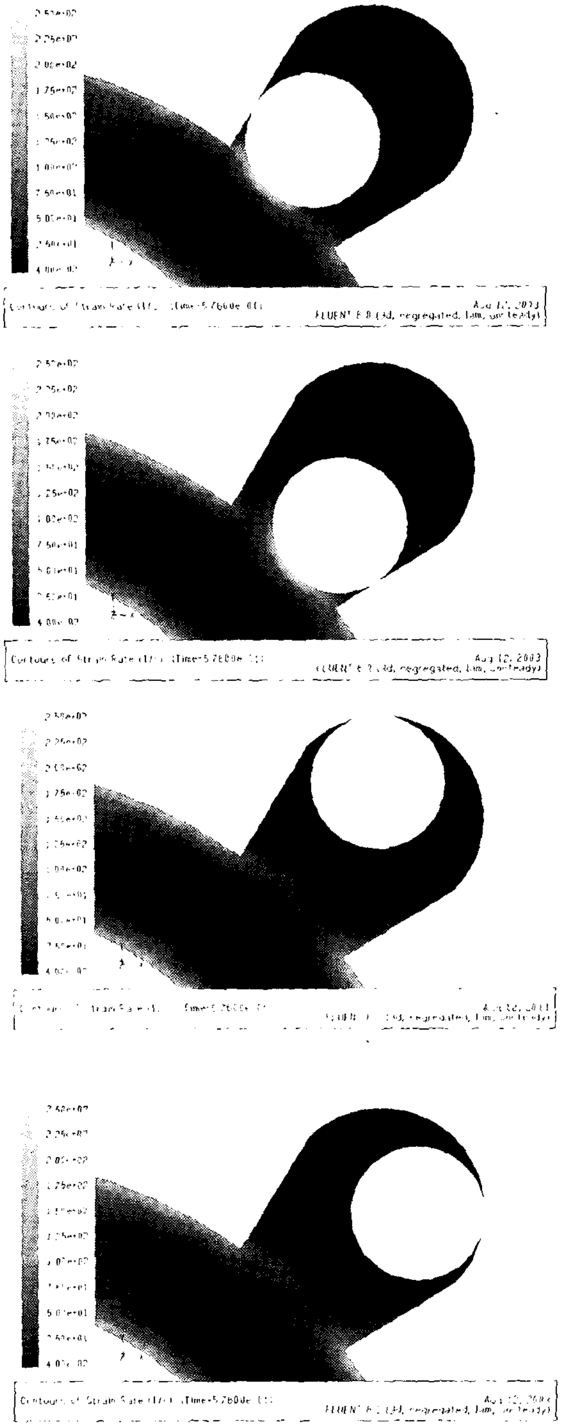

Fig. 5 Shear stress distribution in the midplane of the aneurysm models with a curved parent vessel at peak phase.

was stronger in the proximal dome blocked model comparing to that in the distal
dome blocked model. Wall shear stresses were higher near the distal neck at the mid-half plane of each model. Wall shear stresses at the distal neck were lower in the distal neck blocked model comparing to other models (Fig. 3).

3.2 Curved Parent Vessel

During an acceleration phase, flow fields in the lateral aneurysm model with a curved parent vessel were similar to those with a straight parent vessel in general. But stronger inflow from the proximal neck was observed in the proximal neck blocked model. As the flow accelerated, some inflow was introduced from the side wall of a coil in the proximal neck blocked model but its amount was small. Inflow from proximal neck was observed in the distal neck blocked model. Inflow from the distal neck formed a large vortex inside of the aneurysm sac and moved our through the side wall of a coil in the dome blocked model (Fig. 4). During the deceleration phase, inflow from the distal neck, which was stronger comparing to that in the aneurysm model with a straight parent vessel, was observed in the proximal neck blocked model. Inflow from the side wall of a coil moved out to the proximal neck in the distal neck blocked model, but the inflow was weak. A large vortex formed during the acceleration phase persisted during deceleration phase, and some flow moved out to the proximal neck in the dome blocked models. Wall shear stresses were higher near the proximal neck, and wall shear stresses inside the aneurysm sac were higher in the dome blocked models comparing to the neck blocked models (Fig. 5).

4. Conclusions

We analyzed the flow fields in the partially blocked aneurysm models using computational methods in order to observe the flow field changes affected by coil locations. Inflow was reduced and intraaneurysmal flow was diminished in the neck blocked models comparing to the dome blocked models. A coil located at the distal neck blocked inflow more effectively. Also, the shear stress distributions were lower near the distal neck in the distal neck blocked model, which implied that aneurysm rupture or regrowth was less probable[6]. Stronger inflow was observed in the aneurysm models with a curved parent vessel because of centrifugal force. Therefore inflow blocking by coil insertion was less effective in the aneurysm occurred on the lateral wall of a curved vessel. We suggested that optimal coil location for aneurysm embolization should be distal neck since inflow and intraaneurysmal flow were effectively reduced at this location.

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