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Clinical Study

A growth model of saccular aneurysms based on hemodynamic and morphologic discriminant parameters for risk of rupture

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ABSTRACT

The aim of this study was to derive a model describing the evolution of bifurcation type cerebral aneurysms based on morphological and hemodynamic parameters. Idealized bifurcation models were constructed based on the two morphological parameters of aspect ratio (AR) and size ratio (SR). Aneurysm development was investigated according to the following four patterns: R1, increasing SR with constant AR; R2, increasing AR with constant SR; R3, increasing SR and increasing AR; R4, increasing AR with constant parent artery diameter. Relationships were obtained between energy loss (EL) and morphological parameters (EL-SR and EL-AR curves). The curves were validated by mapping the growth of a ruptured patient-specific bifurcation aneurysm at three stages of follow-up. EL increased in parallel with growth patterns R1 and R3, whereas growth pattern R2 showed a decrease in EL. No significant changes were observed in EL when the growth of the aneurysm was associated only with changes in aneurysm size and independent of changes in parent artery diameter and main flow (R4). Changes in parent artery diameter of bifurcation aneurysms resulted in significant variation in EL. Mapping the growth of a follow-up aneurysm onto the EL-AR curve demonstrated that aneurysms with increasing EL during the observation period are at higher risk of rupture than aneurysms with decreasing EL. Based on the proposed growth model, assessment of morphological (AR and SR) and hemodynamic (EL) parameters may provide quantifiable information on the risk of bifurcation aneurysm rupture during clinical patient follow-up.

1. Introduction

A brain aneurysm is an abnormal widening or dilatation of an artery due to weakening of the vessel wall [1]. Introduction of medical imaging techniques such as CT scan and MRI has increased detection of unruptured aneurysms from 2% to 5% [2]. However, further advanced methods are required to detect the risk of rupture, by understanding the mechanism of aneurysm growth and rupture. These methods will benefit clinicians, specifically neurosurgeons, enabling them to evaluate patient clinical status and assess the treatment course for unruptured aneurysms.

Aspect ratio (AR) has been accepted as a valid clinical measurement to predict aneurysm rupture [3]. AR has been shown to determine intra-aneurysmal flow pattern better than other geometrical indices [3]. However, Raghavan et al. found that shape indices are more effective in predicting the risk of rupture than AR and size of the aneurysm [4]. Dhar et al. defined a new geometric parameter to relate aneurysm size and its parent vessel as the size ratio (SR), calculated as maximum aneurysm height/average parent vessel diameter [5]. Although it was found that among all morphological parameters SR discriminates well between ruptured and unruptured aneurysms [6], neither AR nor SR can explain the growth of an aneurysm without considering the effect of hemodynamic parameters.

Ruptured aneurysms have been found to have more complex flow patterns, a smaller impingement region [7] and a larger kinetic energy ratio (ratio between the kinetic energy of the aneurysm and the kinetic energy in the near-parent artery) [8] compared to unruptured aneurysms. A study of four ruptured and 26 unruptured internal carotid artery–posterior communicating artery aneurysms showed that the energy loss (EL) of ruptured aneurysms was five times higher than that of unruptured aneurysms [9].

For improved rupture prediction, it is of interest to quantify how hemodynamic parameters are affected by morphological parameters, as the high dependency of EL to the variation of daughter artery ratio was shown in a previous study [10]. The aim of this study was to investigate the correlation between two

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morphological parameters, AR and SR, and a hemodynamic parameter, EL, using computational fluid dynamics (CFD) analyses and idealized models. Based on this relationship, a new growth model was explored to predict the risk of rupture. The idealized growth model was validated by mapping the evolution of a ruptured follow-up aneurysm at three sequential stages.

2. Method

2.1. Idealized aneurysm bifurcation model

The following definitions for geometric variables of bifurcation aneurysm based on anatomy of the middle cerebral artery (MCA) were used for each model: D_p, diameter of the parent artery; H, maximal height from the center of the neck to the dome; N, aneurysm neck width; AR, the aspect ratio (H/N); and SR, the size ratio (H/D_p). In all models, the distal outlets were 2.0 mm in diameter and N was 4.0 mm. The bifurcation angle was 180° for all simulations. D_p and H were set across in ranges from 3.0 mm to 4.2 mm and 4.2 mm to 9.0 mm, respectively. The calculated AR and SR are shown in Table 1. The geometry of the idealized model is shown in Figure 1. Due to the unknown relation between daughter artery ratio, AR and SR, this parameter was assumed to be constant in this study.

It has been shown that bifurcation aneurysms with 0° angle relative to the plane of the parent artery at the apex of bifurcation are highly unstable [11] and sensitive to the type of blood flow model (Newtonian *versus* non-Newtonian) [12]. Accordingly, in this study, the angle of the aneurysm dome relative to the parent artery was assumed to be 30° (Fig. 1d).

The outlet boundary was located 100 mm from the apex of the bifurcation to eliminate the effect of boundary condition locations on arterial flow and achieve sufficient recovery of the blood pressure at the branches [13]. An inlet was located 80 mm from the bifurcation apex so flow would be fully developed at the region of entry [14,15].

2.2. Computational modelling

The commercial software ANSYS ICEM 13.0 (ANSYS, Canonsburg, PA, USA) was used for grid generation and segmentation of the follow-up and idealized aneurysms. Each model was constructed with unstructured tetrahedral and prismatic mesh with a minimum 1.5×10^6 grid number according to the grid independency test based on convergency of the EL at the aneurysm neck. The distance between the wall and the first layer of the prismatic mesh was fixed at 0.01 mm. Commercial finite volume software ANSYS CFX.13.0 (ANSYS) was used to perform CFD simulations.

In this study, a total of three simulations were carried out for each model under conditions of low (125 ml/minute), medium (164 ml/minute) and high (218 ml/minute) steady state flow at

Table 1
Morphological characteristic of idealized bifurcation model for aneurysm size ranging from 4.2 mm to 9 mm

Aneurysm size H (mm)	AR	SR D_p (4.2 mm)	SR D_p (4 mm)	SR D_p (3.5 mm)	SR D_p (3 mm)
4.2	1.1	1.0	1.1	1.2	1.4
5.2	1.3	1.2	1.3	1.5	1.8
6.3	1.6	1.5	1.6	1.8	2.1
7.3	1.8	1.7	1.8	2.1	2.5
8.4	2.1	2.0	2.1	2.4	2.8
9.0	2.3	2.1	2.3	2.6	3.0

AR = aspect ratio, D_p = diameter of the parent artery, EL = energy loss, H = maximal height from the center of the neck to the dome, SR = size ratio (H/D_p).

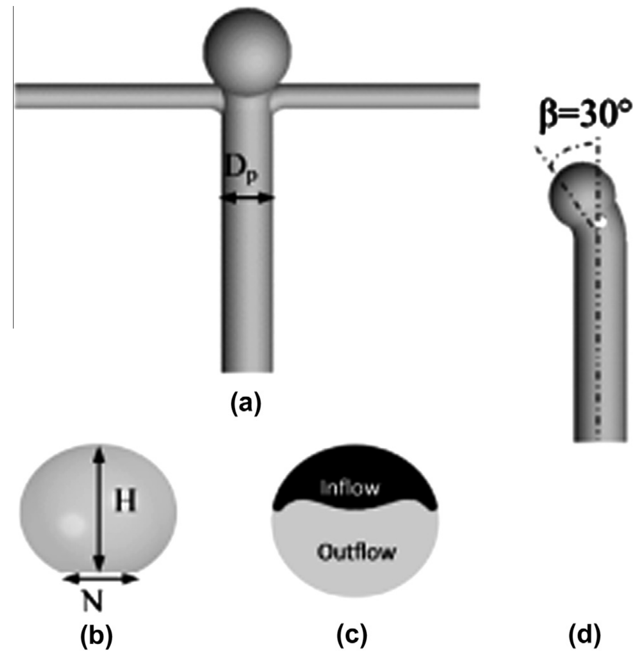

Fig. 1. Diagram of the configuration of (a) bifurcation aneurysm, (b) aneurysm morphology, and (c) aneurysm neck. (d) The right view of a bifurcation aneurysm shows the aneurysm angle relative to the parent artery was assumed to be 30°. D_p = diameter of the parent artery, H = maximal height from the center of the neck to the dome, N = aneurysm neck width.

the parent artery based on mean and peak flow rate in the MCA acquired via phase contrast MRI [16]. An incompressible, steady-state laminar flow model was utilized in the simulation. Due to negligible non-Newtonian behaviour of the blood flow in large arteries (diameter >0.5 mm), the blood was assumed to be a Newtonian fluid [17,18] with blood flow density and dynamic viscosity of 1050 kg/m^3 and 0.0035 Pa.s, respectively [19]. Traction free boundary conditions were assumed for outlets by adopting the conventional assumption of a lack of resistance in the cerebral circulation [20]. A no-slip flow boundary condition was imposed along the artery and aneurysm wall, where the vessel's inner lumen and arteries were assumed to be rigid.

2.3. Calculation of hemodynamic parameters

EL was defined as the difference between energy transport to the aneurysm by influx and energy transport from the aneurysm by outflow. The aneurysm neck in a plane perpendicular to the parent artery is subjected to both inflow to and outflow from the aneurysm. Thus, the neck section area was considered for EL calculation by assuming that positive and negative velocities describe the inflow and outflow (Fig. 1c). EL can be calculated by the following equation:

$$\text{Energy loss(EL)} = E_{inlet} - E_{outlet} \tag{1}$$

$$EL = \underbrace{\sum \left(P_i + \rho \frac{1}{2} v_i^2\right) Q_i}_{E_{inlet}} - \underbrace{\sum \left(P_o + \rho \frac{1}{2} v_o^2\right) Q_o}_{E_{outlet}} \tag{2}$$

where P, v are the static pressure and velocity, respectively; i indicates the inflow to the aneurysm through the aneurysm neck; and o the outflow of the aneurysm through the aneurysm neck. E_{inflow} and $E_{outflow}$ are the spatially averaged energy values over the cross-section of the neck.

It is still unclear how elevation or reduction of wall shear stress (WSS) can affect the growth or rupture of an aneurysm. In the cur-