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Hemodynamic effect of bypass geometry on intracranial aneurysm: A numerical investigation

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ABSTRACT

Background and objective: Hemodynamic analyzes are used in the clinical investigation and treatment of cardiovascular diseases. In the present study, the effect of bypass geometry on intracranial aneurysm hemodynamics was investigated numerically. Pressure, wall shear stress (WSS) and velocity distribution causing the aneurysm to grow and rupture were investigated and the best conditions were tried to be determined in case of bypassing between basilar (BA) and left/right posterior arteries (LPCA/RPCA) for different values of parameters.

Methods: The finite volume method was used for numerical solutions and calculations were performed with the ANSYS-Fluent software. The SIMPLE algorithm was used to solve the discretized conservation equations. Second Order Upwind method was preferred for finding intermediate point values in the computational domain. As the blood flow velocity changes with time, the blood viscosity value also changes. For this reason, the Carreu model was used in determining the viscosity depending on the velocity.

Results: Numerical study results showed that when bypassed, pressure and wall shear stresses reduced in the range of 40–70% in the aneurysm. Numerical results obtained are presented in graphs including the variation of pressure, wall shear stress and velocity streamlines in the aneurysm.

Conclusion: Considering the numerical results for all parameter values, it is seen that the most important factors affecting the pressure and WSS values in bypassing are the bypass position on the basilar artery (L_b) and the diameter of the bypass vessel (d). Pressure and wall shear stress reduced in the range of 40–70% in the aneurysm in the case of bypass for all parameters. This demonstrates that pressure and WSS values can be greatly reduced in aneurysm treatment by bypassing in cases where clipping or coil embolization methods can not be applied.

1. Introduction

Ballooning due to the weakness of the muscle layer in the brain vessels is called cerebral aneurysm. This balloon causes thinning and weakening of the vessel wall. The intracerebral hemorrhage that occurs as a result of the rupturing of this vessel from its weakened position is called subarachnoid hemorrhage. Subarachnoid hemorrhage (SAH) is a devastating event that often causes of mortality and morbidity [1]. The precise cause of brain aneurysms is unknown. However, some factors are considered to contribute to the formation of brain aneurysms. These factors are high blood pressure, smoking, genetic predisposition, damage to blood vessels, some infections. Currently, available treatment options are medical, surgical (clipping) and endovascular therapy (coiling embolization) [2]. Although microsurgical clipping is widely used in the treatment of aneurysms, it is not always appropriate due to its high recurrence rate and its inability to be applied in giant intracranial aneurysms [3,4]. In addition, coil embolization is not preferred in aneurysms with wide neck regions [5]. Extracranial–intracranial (EC–IC) arterial bypass is commonly used in such cases [6,7]. Cerebral bypass surgery is performed to correct or revascularize the blood flow via conventional extracranial to intracranial (EC–IC) or intracranial to intracranial (IC–IC) methods in the brain [8–12].

Numerical hemodynamic analyzes can be used in determining the risk of an aneurysm and in evaluating the above mentioned intracranial cerebral aneurysm treatment methods. Numerous studies have been carried out in the literature to examine the distribution of pressure and wall shear stress on the aneurysm surface with blood flow. The effect of aneurysm and neck region sizes and coil embolization method on blood flow and, pressure and shear stress values on the aneurysm surface for basilar tip aneurysm (BTA) has been demonstrated numerically by Nair et al. [13,14]. The investigation of the pressure and surface tension

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on the aneurysm surface due to the internal carotid artery (ICA) and posterior communication artery (PCOMA) aneurysm geometry was carried out by Qing and Wei-zhe with a computational study [15]. Another numerical study by Xiao-ning et al. emphasized that pressure and shear stress values on the aneurysm surface resulting from the use of stents in bifurcated artery aneurysms can be reduced [16]. Shojima et al. developed a mathematical model for the aneurysm of the middle cerebral artery using 3-D tomographic angiography method and investigated the distribution of wall shear stress in the aneurysm region using CFD calculation [17]. Numerical studies showed that wall shear stress is a factor causing the aneurysm to grow and rupture, and that maximum wall shear stress values occur near the aneurysm neck region. Zhao et al. compared the values of wall shear stress in the case of the aneurysm wall was accepted to be elastic and rigid in their numerical study [18]. The obtained results indicated that the elastic and rigid wall approach gave close results when considered for the entire aneurysm wall, and that it gave different results for the neck region of the aneurysm. Babiker et al. investigated the influence of different stent configurations (half-Y, Y and cross-bar) for cerebral aneurysm treatment on blood flow velocity in the aneurysm, numerically [19]. Numerical results showed that the Y configuration minimized the blood velocity in the neck region of the aneurysm, while the cross-bar configuration minimized the blood velocity in the aneurysm sac. Augsburger et al. used an innovative intracranial stent device (diverter) to reduce the blood flow rate in the aneurysm sac [20]. In the study, steady and pulsatile blood flow conditions were studied for different diverter porosity. The mean blood flow velocity in the aneurysm sac was reduced by 80% and 88% for steady and pulsatile flow with the lowest porosity device, respectively. The effect of the use of coils in the different packing densities for cerebral aneurysm treatment on the blood flow rate studied by Babiker et al. [21]. The inlet blood flow velocity into the aneurysm was reduced by 31.6% for the wide-necked aneurysm type at 28.4% coil packing density, and 49.6% for the narrow-neck aneurysm type at coil packing density of 36.5%.

In the studies available in the literature, pressure, wall shear stresses and blood flow velocity have been investigated depending on parameters and devices such as the size of the aneurysm, using of different stent type and using of coil packing which has different densities. In the present study, however, it was aimed to investigate the pressure and wall shear stress (WSS) values at the aneurysm surface depending on different parameter values in case of bypassing between the basilar and posterior arteries for the basilar tip intracranial cerebral aneurysm. In this direction, an idealized basilar tip aneurysm model (IBTA) was created and the pressure and WSS values, which vary depending on the position of the bypass connection on the arteries and the diameter of the bypass vessels, were numerically investigated to establish the most appropriate conditions.

2. Numerical study

2.1. Problem specification and mathematical model

A simplified idealized Basilar Tip Aneurysm (IBTA) model used for numerical study is given in Fig. 1a. The geometry is shown in Fig. 1a was based on the model obtained by using patient-specific clinical data in a study by Xiao-ning et al. [16]. The length of the basilar artery was taken to be 10 times greater than the diameter of the basilar artery in order for the flow to be fully developed. The diameters of the left posterior cerebral artery (LPCA), right posterior cerebral artery (RPCA) and basilar artery are equal (4 mm). Fig. 1b shows the bypass geometry applied to the right and left posterior cerebral arteries from the basilar artery. Numerical analyzes were performed for different diameter values of the bypass vessels and for different localization positions on the left / right posterior cerebral and main arteries of the bypass connection. Numerical analyzes were carried out for different values of the parameters given in Fig. 1. The parameter values used are given in Table 1.

Table 1
Parameter values.

R (mm)	L_b (mm)	L_p (mm)	d (mm)
30	25	5	2,5
40	45	15	3
80	55	25	4

Mass and momentum conservation equations for blood flow under laminar conditions are given by Eqs. (1) and (2), respectively,

$$\nabla . v = 0 \tag{1}$$

$$\rho\left(\frac{dv}{dt} + v.\nabla v\right) = -\nabla P + \mu \nabla^2 v \tag{2}$$

Where v is the velocity vector, ρ (1060 kg/m^3) is the blood density, μ is the dynamic viscosity of blood and P is the pressure. Viscosity is not a constant and it is a function of shear rate ($\dot{\gamma}$). The dynamic viscosity is defined by Eq. (3) (Carreu model),

$$\mu(\dot{\gamma}) = \mu_\infty + (\mu_0 - \mu_\infty)[1 + (\lambda\dot{\gamma})^2]^{\frac{n-1}{2}} \tag{3}$$

For blood flow, the coefficients in Eq. (3) are given by the following equations [22],

$$\begin{aligned} \mu_0 &= 0,056\ (\mathrm{kg/m\,s}) \\ \mu_\infty &= 0.0035\ (\mathrm{kg/m\,s}) \\ \lambda &= 3.313\ (\mathrm{s}) \\ n &= 0.3568 \end{aligned}$$

Where μ_0 is the dynamic viscosity at zero shear rate, μ_∞ is the dynamic viscosity at a high shear rate, λ and n are the coefficients.

The non-slip condition on blood vessel walls was assumed. Compared with the cerebral arteries and the aorta, the elasticity of the cerebral arteries is negligible [16]. Thus, the blood vessels are considered to be rigid for the simplification of the study. In the numerical analyzes, patient-measured velocity values were used for the time-dependent inlet velocity profile of the blood flow. The flow velocity profile obtained from the patient is shown in Fig. 2.

In order to incorporate the velocity profile seen in Fig. 3 into the numerical analyzes, the approximate velocity profile curve is obtained using the following equations.

$$v_{inlet}(t) = \begin{pmatrix} 0.18, & t \le 0.05 \\ 0.4\sin[3.5\pi(t + 0.0015)], & 0.05 < t < 0.21 \\ -0.5389t^3 + 1.1966t^2 & \\ \quad -0.9189t + 0.4393, & t \ge 0.21 \end{pmatrix} \tag{4}$$

In the Eq. (4), v_{inlet} is the inlet velocity of blood and t is the time.

2.2. Numerical method

The finite volume method was used for numerical solutions and calculations were performed with the ANSYS-Fluent software [23]. The SIMPLE algorithm is used to solve the discretized conservation equations. Second Order Upwind method was preferred for finding intermediate point values in the computational domain. As the blood flow velocity changes with time, the blood viscosity value also changes. For this reason, the Carreu model Eq. (3) was used in