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Newtonian and non-Newtonian blood flow in coiled cerebral aneurysms

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ABSTRACT

Endovascular coiling aims to isolate the aneurysm from blood circulation by altering hemodynamics inside the aneurysm and triggering blood coagulation. Computational fluid dynamics (CFD) techniques have the potential to predict the post-operative hemodynamics and to investigate the complex interaction between blood flow and coils. The purpose of this work is to study the influence of blood viscosity on hemodynamics in coiled aneurysms. Three image-based aneurysm models were used. Each case was virtually coiled with a packing density of around 30%. CFD simulations were performed in coiled and untreated aneurysm geometries using a Newtonian and a Non-Newtonian fluid models. Newtonian fluid slightly overestimates the intra-aneurysmal velocity inside the aneurysm before and after coiling. There were numerical differences between fluid models on velocity magnitudes in coiled simulations. Moreover, the non-Newtonian fluid model produces high viscosity (>0.007 [Pa s]) at aneurysm fundus after coiling. Nonetheless, these local differences and high-viscous regions were not sufficient to alter the main flow patterns and velocity magnitudes before and after coiling. To evaluate the influence of coiling on intra-aneurysmal hemodynamics, the assumption of a Newtonian fluid can be used.

1. Introduction

The insertion of endovascular coils in a cerebral aneurysm aims to occlude and isolate it from blood circulation, which reduces the aneurysm rupture risk and subsequent subarachnoid hemorrhage (van Gijn and Rinkel, 2001). The aneurysm occlusion after coiling is achieved by increasing the resistance to flow, inducing the coagulation cascade and maintaining a stable thrombus inside the aneurysm. Nonetheless, the influence of coils on local hemodynamics, wall mechanobiological response and blood rheological changes are not fully understood, and therefore, the therapeutic outcome is difficult to predict. Hemodynamics has been associated with the success and failure of endovascular coiling (Bhatti et al., 2004). Recanalization, which is the reopening of the aneurysm to blood circulation, has been reported to occur in 25% of the cases (Sluzewski et al., 2003).

To investigate the hemodynamic alterations after coiling inside the aneurysm, computational techniques have arisen as a promising alterative to pure theory and laboratory experiments. Computational techniques have several advantages, including reduced costs, predictive capabilities, and the possibility to test different treatments options before the real intervention takes place.

Hemodynamics modeling using computational fluid dynamics (CFD) depends on the accuracy and underlying assumptions of the tool. For example, treating blood as a Newtonian fluid, i.e., with constant viscosity, is a simplification of blood rheology. Blood is composed by red and white cells, and platelets floating in plasma that make it a Non-Newtonian fluid with a shear-thinning behavior (Kim, 2002). This means that blood viscosity decreases with increasing shear strain rate. Viscosity has an important role in keeping vascular homeostasis and it is linked to the coagulation process (Chen et al., 2012; Wootton and Ku, 1999), which is desired after endovascular therapies. Thus, an accurate model of blood viscosity is required. It has been shown a good agreement between the computational modeling and experimental data when Newtonian and non-Newtonian fluids are used in large arteries (Gijsen et al., 1999).

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In the case of cerebral aneurysms and hemodynamic modeling, several studies have investigated the role of blood viscosity. Cebral et al. (2005) reported that the resulting flow fields in untreated aneurysms using a non-Newtonian model of blood flow were not greatly affected by assuming a Newtonian model. Valencia et al. (2006) found some differences in blood flow at regions with high velocity gradients, and thus high shear strain rates, but the numerical solutions using a Newtonian and a non-Newtonian models were similar at the aneurysm. Utter and Rossmann (2007) investigated the influence of aneurysm morphology on its rupture risk using a non-Newtonian blood flow model. Fisher and Rossmann (2009) showed that intra-aneurysmal hemodynamics locally depends on the non-Newtonian model that is used. Additionally, Cavazzuti et al. (2010) extended the results of the Virtual Stenting Challenge in aneurysm treated with high-porosity stents (Radaelli et al., 2008) by including a non-Newtonian fluid model.

Previous CFD studies on coiled aneurysms have used either Newtonian or non-Newtonian blood flow (Kakalis et al., 2008; Morales et al., 2011; Cebral and Löhner, 2005; Wei et al., 2009; Schirmer and Malek, 2010). None of these studies have looked into the impact of blood viscosity on intra-aneurysmal hemodynamics after coiling. In coiled aneurysms, lower velocities and higher residence time than untreated aneurysms have been observed in both experiments and CFD simulations (Sorteberg et al., 2002; Morales et al., 2011). These hemodynamic conditions have been associated with blood coagulation (Chen et al., 2012; Wootton and Ku, 1999; Rayz et al., 2008).

The purpose of this work is to investigate the role of blood viscosity on intra-aneurysmal hemodynamics after coiling and to evaluate the accuracy of using a Newtonian model in the resulting flow fields by CFD simulations. Additionally, simulations of untreated aneurysms were included as a reference for evaluating coiling outcome and to compare the numerical results with previous studies.

2. Materials and method

2.1. Materials

Three image-based aneurysm geometries were used in this study. All cases were located in the supraclinoid segment of the internal carotid artery (ICA). These geometrical models of the aneurysms and parent arteries were extracted from three-dimensional rotational angiography (3DRA) images that were acquired by an AXIOM Artis (Siemens Medical Solutions, Erlangen, Germany) system. According to patient condition, aneurysm morphology and clinical expertise, all cases were suitable for endovascular coiling.

2.2. Image and surface mesh processing

Surface meshes of the aneurysms and arterial lumens were obtained by segmentation of the 3DRA images using a geodesic active region method (Bogunović et al., 2011). After image segmentation, small vessels were removed, superficial holes were filled and the final surface of the vasculature was smoothed for further CFD analysis and post-processing of flow fields.

2.3. Virtual coiling

Coil models were created using a computational technique based on a dynamic path planning algorithm (Morales et al., 2013). This coiling technique reproduces the structure and distribution of endovascular coils inside saccular aneurysms according to measurements on histological images of real data (Morales et al., in press). Aneurysm geometries were filled up with 0.254-mm coils to around 30% of their volume, which is a high aneurysm filling in the clinical practice using standard bare coils (Sluzewski et al., 2004). The use of highly packed aneurysms avoids the dependence of the intra-aneurysmal hemodynamics on the coil configuration (Morales et al., 2011).

Table 1
Number of elements ($\times 10^6$).

Case	Untreated	Coiled
1	1.70	2.06
2	3.52	9.23
3	0.94	1.76

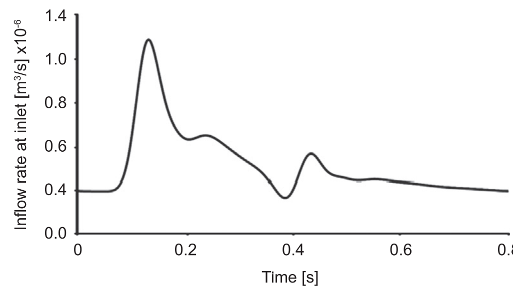

Fig. 1. Inflow rate at the inlet of case 1.

2.4. Volumetric grid

ICEM-ANSYS v12 (Ansys, Inc., Canonsburg, PA) was used to generate non-structured tetrahedral meshes in the fluid domains of coiled and untreated models. The meshes were refined near endovascular coils and independence of CFD simulations on the volumetric grid was evaluated. Element size was 0.12 mm at the vessel wall, 0.06 mm at the aneurysm wall, 0.06 mm at the coil surface and 0.24 mm elsewhere. Table 1 summarizes the number of elements used in each case.

2.5. CFD modeling

After volumetric grid generation, CFD simulations were carried out with the commercial software ANSYS-CFX v12. Blood was assumed as an incompressible fluid with density of 1066 kg/m^3 in a laminar flow regimen. Rigid wall with no-slip condition was applied at vessel and aneurysm walls, as well as at coil surfaces. Steady-state simulations were performed in all cases. An area-based mass flow was set at the inlet of each case, which was taken from physiological measurements at the ICA (Cebral et al., 2008). Zero-pressure conditions were imposed at all the outlets.

Additionally, unsteady simulations were performed in one case to evaluate the influence of pulsatile flow on the resulting intra-aneurysmal hemodynamics. Two cardiac cycles of 0.8 s were calculated and the first cycle was not taken into account to remove initial transients, which was previously reported (Cebral et al., 2005; Morales et al., 2011; Pereira et al., 2013). Time steps were set every 0.005 s. Fig. 1 presents the used pulsatile waveform that was taken from a validated one-dimensional model of the whole arterial tree (Reymond et al., 2009). Time-averaged mass flow was equal to the one imposed in the steady-state simulation of this case.

2.6. Viscous models

To evaluate the influence of blood viscosity on local hemodynamics in coiled and untreated aneurysms, a Newtonian and a non-Newtonian fluid were considered. The Newtonian fluid was set with a viscosity of 0.0035 Pa s. This value is typically used in literature for Newtonian blood flow models (Cebral et al., 2005; Morales et al., 2011; Kakalis et al., 2008). The Casson model was used for the non-Newtonian blood, which mathematically describes the dynamic viscosity μ as follow:

$$\sqrt{\mu} = \sqrt{\frac{\tau_0}{\dot{\gamma}}} + \sqrt{\mu_0} \tag{1}$$

where, $\dot{\gamma}$ is the shear strain rate and τ_0 is the yield stress equal to 0.004 Pa. The parameter μ_0 corresponds to the blood viscosity of 0.0035 Pa s. These parameters were taken from the literature (Cebral et al., 2005). Fig. 2 shows μ_0 as function of $\dot{\gamma}$ for both fluids under evaluation.