CFD Study of the Blood Flow in Cerebral Aneurysms Treated with Flow Diverter Stents

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Flow diverters are high-density meshed stents designed to reduce the flow into the aneurysm and enhance the potential of aneurysmal thrombosis formation, which presents a condition expected to reduce the risk of growth and rupture of residual aneurysms. In this study, the blood flow in cerebral aneurysms with implanted stents is modeled and analyzed taking into account the design and porosity of the flow diverter, where the focus is the flow reduction through the cerebral aneurysm. For a high-density mesh flow diverter, the reduction in the flow through the aneurysm was higher compared to the low-density mesh regular stent. Thus, a regular stent may not provide enough efficiency in flow reduction inside the aneurysm, possibly causing treatment failure, and the flow diverter is a more efficient invasive treatment to reduce the risk of rupture in cerebral aneurysm. Computational fluid dynamics (CFD) simulations of intracranial aneurysm hemodynamics usually assume a Newtonian blood rheology model. The present study shows that this assumption may overestimate the wall shear stress in intracranial aneurysm domes and underestimate the risk of rupture.

1 Introduction

Cerebral aneurysm is a vascular disease characterized by the local dilatation of arterial walls in the intracranial space. The prevalence of intracranial aneurysms in the general population is approximately 2%-3% [1]. Rupture of an aneurysm can cause subarachnoid hemorrhage, associated with a high risk of mortality and morbidity [2]. Due to widely used imaging methods in clinical surgery, the detection of unruptured aneurysms becomes more and more frequent. While patient-specific (e.g. family history, smoking) and aneurysm-specific (e.g. size, location) factors increase the risk of rupture, they are not very specific. In order to reduce the risk of rupture and hemorrhage and to identify highly effective treatment options, the understanding of the hemodynamic mechanisms involved is of great importance, and numerical tools may provide excellent support of the medical treatment of cerebral aneurysms.

There are two effective approaches for the treatment of intracranial aneurysms: Clipping of the aneurysmal neck, and endovascular intervention. The latter one is the treatment of choice for cerebral aneurysms because of both its safety and efficacy. The major advantage of this treatment is the fact that there is no need to do a craniotomy, avoiding the exposure of the surface of the brain vessel.

The flow diverter stent is a promising method of endovascular reconstruction for large and complex intracranial aneurysms, with an overall porosity metallic mesh placed in the parent artery to reduce blood flow in the aneurysm to the point of stagnation and gradual aneurysmal thrombosis [3].

Computational techniques offer new capabilities in the healthcare provision for intracranial aneurysms. The availability of a simulation tool for a virtual flow diverter is extremely useful to support the decisions of treatment options by medical experts and to develop and optimize new implant designs [4]. In the present project, the development of a computational tool for the modeling and simulation of the hemodynamic effects of the endovascular stents with different porosities are developed. Both a flow diverter and a regular stent are modeled and analyzed taking into account their design, porosity, and the flow reduction through a giant sidewall cerebral aneurysm. A workflow which is able to connect the clinical data and a numerical software package to carry out CFD simulations based on patient-specific computational tomography (CT) medical images, and patient-specific boundary conditions is demonstrated. A model based on geometrical properties of braided flow diverter stents that includes the wires' location as well the length and local porosity of the stent in a patient specific vasculature is used.

The study includes the influence of the blood rheology assumption of a Newtonian versus non-Newtonian viscosity model to investigate the wall shear stress in the dome of the cerebral aneurysm, which is considered to be associated with the risk of rupture.

The simulation intends to provide realistic insight into the pathological vessel parameters and better evaluation of the risk of rupture for a given patient.

2 Computational Grid, Mathematical and Numerical Methods

An idealized giant sidewall cerebral aneurysm, cf. Fig. 1, with a diameter of 33 mm is modeled based on a three-dimensional cerebral rotational angiography image provided by the Neuroradiology Department of Heidelberg Medical School. The spherical aneurysm is located at distance 2.0 mm above a straight cylindrical artery of diameter 4.5 mm. The length of the artery inlet to the aneurysm proximal is 52 mm and the length of the aneurysm distal to the artery outlet is also 52 mm.

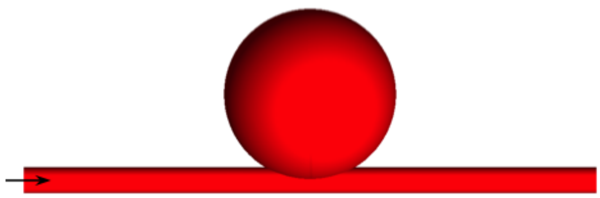
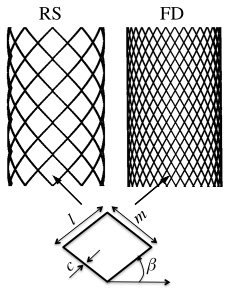


Figure 1: Geometry of the giant cerebral aneurysm

Two stents with meshes made of cylindrical wires are considered which fit the shape of the parent artery. Their geometrical configuration and properties are summarized in Fig. 2 in terms of a stent unit cell [5]. The stent with a low metal coverage proportion and high porosity is called regular stent (RS) and that with a high metal coverage proportion and low porosity is known as flow diverter stent (FD).

Patient-specific geometries for CFD are created from 3D rotational angiogram (3DRA) for



Dimension	RS	FD
l [mm]	0.96	0.30
m [mm]	0.94	0.30
c [µm]	60	42
β [°]	49	55
ϵ (porosity) [-]	0.88	0.73

Figure 2: Left: geometrical configuration; Right: properties of the regular stent (RS) and flow diverter (FD)

every patient. The CT images are imported into the software ImageLab to repair and delete artifacts, segmentation and smoothing of the region of interest, as shown by the Fig. 3, left and center parts. A terminal middle cerebral aneurysm with a diameter of 9.0 mm, was provided by the Heidelberg Medical School, and an internal carotid artery (ICA) aneurysm shown in Fig. 4 (left) with a diameter of 4.25 mm was obtained from the Toronto Western Hospital, Canada.

The computational grids for the simulations were generated using the software ICEM-CFD v.11 (Ansys Inc.). The tetrahedral mesh number is increased until the flow field independent of the number of grid cells is guaranteed. For the untreated aneurysm (UA) model, 6.14 million grid

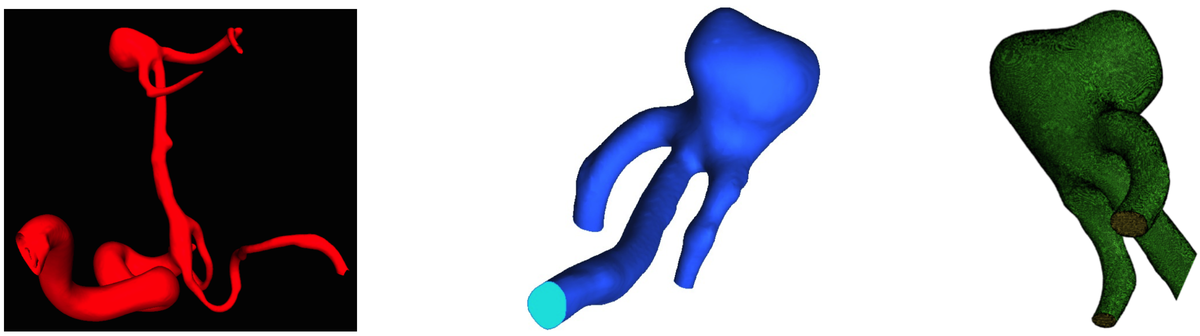


Figure 3: Terminal middle cerebral aneurysm: segmentation (left), geometry (center) and mesh (right)

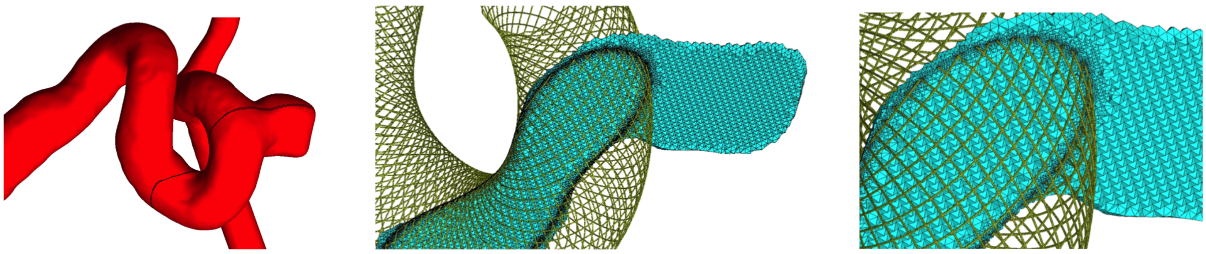


Figure 4: ICA aneurysm with flow diverter: geometry (left), meshing cut plane (center), and image zoom of the meshing cut plane (right)

cells are used, and for the aneurysm merged with the RS, 11.39 million cells are used, whereas for the aneurysm combined with the FD, 11.93 million grid cells are necessary.

Considering the patient-specific aneurysms, 8.3 million grid cells for terminal aneurysm as shown by the Fig. 3 (right), and 31 million grid cells for the ICA aneurysm deployed with the flow diverter as shown in Fig. 4 center and right parts.

The laminar flow is described by three-dimensional incompressible transport equations, which are solved using the finite volume based software platform OpenFoam v3.1. A uniform velocity profile of 0.56 m/s for the peak systole is prescribed at the inlet [6], and the pressure gradient at the outlet is zero. All the vascular walls are assumed rigid with a no-slip boundary condition, and the elasticity of the stent is neglected.

The blood flow is not homogeneous and its rheological properties are mainly dependent on the hematocrit or the volume fraction of red blood cells in the blood, cholesterol, fibrinogen, for instance. In the present study, the most commonly used power-law model of Carreau-Yasuda [7] is used for the non-Newtonian fluid.

For the UA, the high performance computing resource used was 64 processors during 36 hours clock time. For the RS and FD, 256 processors were used with 4 and 7 days clock time, respectively. In the patient-specific flow diverter situation, the simulation was performed during two weeks clock time using 256 processors.

3 Results and Discussion

The terminal aneurysm shows a slowly recirculating secondary vortex near the dome. The non-Newtonian and Newtonian models predict similar velocity and wall shear stress (WSS) distributions in the parent vessels of the aneurysm. However, as shown in the Fig. 5, large discrepancies in the WSS between the predictions of the different rheology models are found in the dome area of the terminal aneurysm, where the flow is relatively stagnant. In this region, the non-Newtonian model predicts lower shear rates and WSS values as well as a higher blood viscosity compared to the Newtonian model. Thus, the assumption of a Newtonian blood flow may underestimate the risk of rupture in the aneurysms.

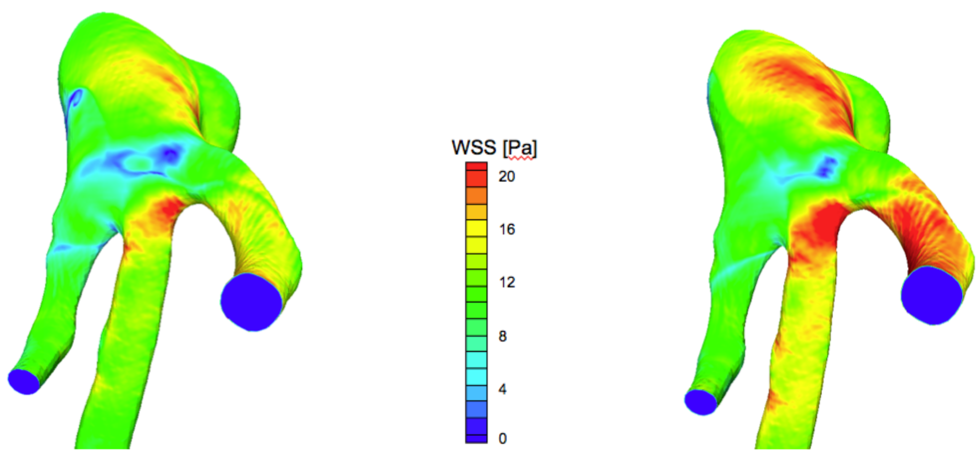


Figure 5: Wall shear stress distribution in front view: non-Newtonian (left) and Newtonian (right) blood flow

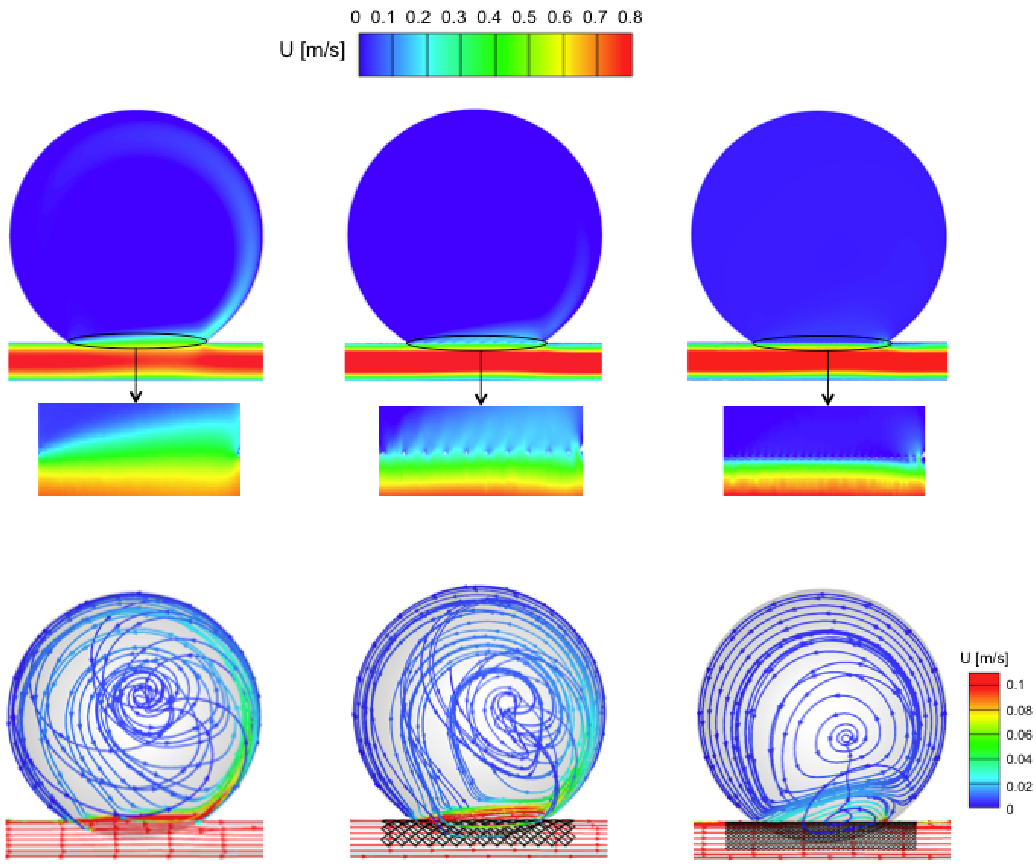


Figure 6: Velocity contour at the mid plane (top) and velocity streamlines (bottom) of the untreated aneurysm (left), regular stent (center) and flow diverter (right) during the peak systole

The magnitude of the flow velocity inside the aneurysm is strongly reduced due to the use of the RS and even more due to the FD, as shown in Fig. 6. The reduction of the aneurysm inflow and the flow activity after the implantation of a stent or flow diverter is a key factor for thrombus formation inside the aneurysm.

In both the pure aneurysm and the aneurysm treated with the regular stent, the flow pattern at the neck of the cerebral aneurysm consists of a main clockwise distal inflow jet. A possible physical mechanism to describe this phenomenon involves both shear stress and pressure [8]. The shear stress transmitted at the cerebral aneurysmal neck, induced by a strong velocity gradient in the parent artery, drives a counter-clockwise intra-aneurysmal flow. The use of the flow diverter causes the flow to transfer into a clockwise streamline from the proximal to distal side of the neck, driving a low motion swirl to the aneurysm's dome. The implantation of the FD impedes the flow at the aneurysm's neck, resulting in both a lower shear stress transmission and an increase of pressure gradient along the parent artery. The latter pushes the circulating fluid inward and outward the cerebral aneurysm at the proximal and distal sides of the neck, respectively.

4 Conclusions

The flow diverter effectively reduces the blood flow velocity into cerebral aneurysms and provides a structure to support endothelialization and reconstruction of the parent artery. The use of the flow diverter is superior compared to the regular stent, and flow separation at the distal side of the neck is observed, promoting aneurysmal thrombosis. The Newtonian fluid assumption may underestimate the flow viscosity and overestimate the WSS in regions of stasis or slow recirculation typically found in the dome in complex-shaped aneurysms as well as in aneurysms following endovascular treatment.

Acknowledgements

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