Measurement of flow diverter hydraulic resistance to model flow modification in and around intracranial aneurysms

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Abstract: Flow diverters (FDs) have been successfully applied in the recent decade to the treatment of intracranial aneurysms by impairing the communication between the flows in the parent artery and the aneurysm and, thus, the blood within the aneurysm sac. It would be desirable to have a simple and accurate computational method to follow the changes in the peri- and intraaneurysmal flow caused by the presence of FDs. The detailed flow simulation around the intricate wire structure of the FDs has three disadvantages: need for high amount of computational resources and highly skilled professionals to prepare the computational grid, and also the lack of validation that makes the invested effort questionable. In this paper, we propose a porous layer method to model the hydraulic resistance (HR) of one or several layers of the FDs. The basis of this proposal is twofold: first, from an application point of view, the only interesting parameter regarding the function of the FD is its HR; second, we have developed a method to measure the HR with a simple apparatus. We present the results of these measurements and demonstrate their utility in numerical simulations of patient-specific aneurysm simulations.

Keywords: intracranial aneurysm, flow diversion, computational fluid dynamics, wall shear stress, pressure

Introduction

Intracranial aneurysms are dilated sections on the large arteries supplying the brain. The rupture of an intracranial aneurysm is a common source of stroke, which in many cases results in the death or paralysis of the patient. With the help of modern medical imaging techniques, unruptured intracranial aneurysms are often found [1].

In case of large-sized aneurysms, a method of treatment is the reconstruction of the parent artery using one or multiple flow diverter stents [2] layered on each other. The flow diverters (FDs) are chosen based on their nominal diameters suiting the internal diameter of the parent vessel. These devices opened a new era providing a previously unseen high rate of complete and stable occlusion in large, giant, and wide-neck intracranial aneurysms [2–5]. There is no consensus, however, as to how much flow reduction is needed to achieve complete thrombosis.

Therefore, the flow modification effects of FDs are widely studied using computational fluid dynamics (CFD) [6–8]. These studies show that treatment results in significant alteration of the flow inside the aneurysm sac as well as cause the change in the hemodynamic forces on the parent artery. In a former study, Appanaboyina et al. [6] highlighted the elevated wall loads upstream of the aneurysm after stenting. Most studies follow a strategy in which they accurately model the fine stent structure in the numerical model [6–8]. To model the densely woven stents at this level of detail demands such a high computational effort, which might not be justified by the accuracy of the results. To reduce the computational need of the flow diverter simulations, Augsburger et al. [9] proposed a porous substitution of the stent layer. The authors used numerical simulations to find the permeability of the simulated devices. They compared the resulting flow modification with the exact modelling approach and found an agreement between
the two methods. The largest differences were found in the vicinity of the aneurysm neck and during the systolic peak in time.

Manufacturers provide the mesh density of commercial FDs, but their capacity in reducing flow is unknown and is not taken into consideration while choosing a technique for individual aneurysms. In our work, we follow a similar approach to Augsburger et al. [9] but extending the study with laboratory experiments of the devices. Also, we study the permeability of additional stent layers. We aim to develop a technique that is capable of determining the efficiency of FDs in reducing intraaneurysmal flow.

Materials and Methods

Measurements to determine the hydraulic resistance of FDs

The hydraulic resistance (HR) of the pipeline embolic device (PED) is defined as the static pressure gradient through the device. The HR of one or two layers of the woven PED (Covidien/ev3, Irvine, CA, USA) was measured in an experimental apparatus as a function of flow rate \( Q \) as depicted in Fig. 1. The apparatus consists of a flow tube with an internal diameter matching the nominal diameter of the FD to be studied. A hole is cut in the wall of the tube and is covered by the FD(s) that is (are) placed within the tube. Either one or two coaxial layers of a 3.25 × 20 mm PED were used. The flow was fed from a water tank through a throttle valve to set the flow rate. A symmetric configuration was used by letting water through from both sides of the tube to avoid errors arising from a significant parallel velocity component through the FD. The tube was immersed in a second water tank to avoid errors due to surface tension on the surface of the FDs.

The fluid exits the tube through the opening covered by the FD(s). The flow rate was measured by levelling the overflow of the tank into which the water flows through the tube. A mean through-flow velocity was calculated from the volume flow by dividing it with the cylindrical section area of the FD exposed to the exiting flow.

The pressure after the FD is the ambient pressure corrected with the small constant hydrostatic pressure due to the water column. The pressure before the FD was measured inside the tube, using two pressure taps on both sides of the tube before the hole covered by the FD. The pressure drop across the FD was recorded in relation to flow rate.

The HR of the FDs was expressed by the relation between the flow velocity and pressure gradient according to Darcy’s model [Eq. (1)]:

\[
- \frac{\Delta p}{\Delta x} = \frac{v \mu}{K} + \frac{v|v|B\rho}{2}. 
\]

In Eq. (1), \( \mu \) is the dynamic viscosity of the liquid, \( K \) is the permeability of the porous region, \( B \) is an empirically determinable constant, \( \rho \) is the density of the liquid, and \( x \) is the average direction of the velocity \( |v| \). The left hand side of Eq. (1) is the pressure gradient that is, in our case, approximated by \( \Delta p/\Delta x \). Here, \( \Delta p \) is the pressure drop across the layer described before and \( \Delta x \) is the thickness of the FD layer. Measuring the pressure loss at various flow velocities, the HR of the FD could be determined as characterized by the constants \( K \) and \( B \).

The HR of the system was measured without FD and was later subtracted from the values measured with FDs. A single FD configuration was measured twice. Measurements with two FD layers were repeated five times. Each time, the FDs were removed from the apparatus and reinserted separately, thereby randomly creating five different relative positions of the two FDs.

As the measurements were carried out with water, the results were first transformed into non-dimensional quantities as calculated with the properties of water and then dimensionalized again with the viscosity and density of blood. Although the measurements were carried out in steady flow, the results were well applicable in a pulsatile flow such as blood flow, where the rate of change of the blood flow rate is small. Differences in the

![Fig. 1.](image_url)
steady and unsteady HRs are expected for much higher accelerations or decelerations only.

Modelling of aneurysms and simulation of perianeurysmal flow

Patient-specific finite volume (FV) aneurysm models were created from three-dimensional (3D) reconstruction of rotational angiography in two human cases. An accepted indicator of the risk of an aneurysm was found to be the aspect ratio, which is the depth of the aneurysm sac compared to the neck diameter proposed by Ujiie et al. [10]. For this study, two paroophthalmic internal carotid artery aneurysms that represent typical indication of FD treatment were chosen. To mimic the impact of neck size, a low aspect ratio (LA; 1.3) and a high aspect ratio (HA; 1.8) aneurysm were selected (Fig. 2). The FV models were created using a technique previously described [11–13]. The technique is similar to other methods used to carry out patient-specific simulations [14–16].

Briefly, a 3D surface of the vessel geometry is extracted from the 3D model obtained by rotational angiography. The geometry assembly is meshed with ICEM CFD (ANSYS, Canonsburg, Pennsylvania) using tetrahedral elements. The vessels of the resulting 3D mesh are extruded at their ends to decrease the effect of boundary conditions applied on these geometric boundaries. As an inlet boundary condition, the velocity with a parabolic profile, the spatial average varying between 0.37 and 1 m/s in a cycle of 0.8 s, is prescribed. The other boundary conditions, flow and fluid parameters, were the same as described by Kulesar et al. [12].

The flow was simulated using ANSYS CFX 12.0. By simulating flow in these models, the mean intraaneurysmal flow velocity was extracted and the wall shear stress (WSS) and pressure distribution within the aneurysm and the parent vessel was analyzed both prior to and following virtual deployment of a single and a double layer of FD. For a better demonstration of the impact of FD on WSS and pressure, the post-treatment simulations were subtracted from the pre-treatment studies.

FD simulation

FDs placed within the parent artery across the entrance of the aneurysm were simulated as 200-μm thick layers of porous material. These layers are inserted using a method similar to that introduced by Appanaboyina et al. [6].

On the FV models, first, the centerlines of the vessel are calculated (Fig. 3a, solid line) using an open-source algorithm from the Vascular Modelling Toolkit (www.vmtk.org). Based on the generated centerlines, the axis of the FD is manually defined by its starting and end points (Fig. 3a, dashed line) extending both proximal and distal to the aneurysm neck. For initial tube modelling, the FD is constructed with a diameter smaller than that of the parent vessel. The proximal and distal edges of the initial meshed tube surface are snapped and fixed to the vessel wall (Fig. 3b).
The initial tube geometry is then expanded until its surface touches the artery wall but without changing its original constant diameter (Fig. 3c). Sections of this tubular structure are then removed, so that the remaining surface covers only the aneurysm entrance. The wall of this remaining virtual tube is then extruded up to 200 μm (Fig. 3d). This 200-μm thick layer is considered as a porous material. The HR of the porous layer perpendicular to its axis is set according to the measured HR of the FD(s) to be simulated. The resistance of the porous volume parallel to the surface is neglected. The reason why the simulated layer thickness differs from the real FD thickness is simply that it is easier to lay a numerical mesh on a thicker layer. The thickness has no further importance; the only important parameter from the viewpoint of the flow simulation is the HR and that was ensured to remain constant.

Fig. 3. Four steps of the virtual flow diverting process: a) calculation of vessel centerlines (solid line) and definition of FD axis (dashed line); b) insertion of collapsed tube and snapping edges; c) inflation of tube surface; d) removing unnecessary parts of the inflated tube and extruding the remaining surface to form the porous layer

Fig. 4. Pressure drop of FD layers as a function of flow rate in water. A parabolic curve is given with dark grey dotted line fitted on the measurement data in case of the single layer FD measurement. In case of measurements with two FDs, a parabolic curve was fitted onto each of the five individual measurements, indicated with light grey dotted lines. The average of the five fitted curves is indicated with dark dashed line
Results

Results of the hydraulic resistance measurements

Results of HR measurements are demonstrated in Fig. 4. The data were obtained after subtracting the HR of the empty apparatus from the FD HR measurements. For measurements with a single layer, a parabolic curve, intersecting the origin, was fitted on the data points (Fig. 4, dark grey dotted line). This curve was used to calculate the HR of a single FD layer. At the same time, the best-fit curve yielded the parameters $K$ and $B$ [Eq. (1)].

The individual light grey dashed lines demonstrate that, in case of a double layer, each measurement produced different results indicating the significance of the relative position of the layers. The individual FD, on the other hand, had always the same HR as indicated by the dots in Fig. 4. It was found that the mean curve of the five individual two-layer measurements (dashed dark curve) gives twice the pressure loss of the one, fitted on the single layer measurements (dark dotted curve).

Since the measurements were carried out using water, the results were first recalculated to the liquid properties of blood used in the simulations. Then, the parameter values in Eq. (1), such as the permeability ($K$), the linear ($\mu/K$), and the quadratic velocity loss coefficients ($B$) calculated from the measurements were corrected for the simulated layer thickness so that the real and the simulated layer had the same net HR as defined by the values in the first row of Table I. The latter pair of $K$ and $B$ values was used in the simulations. For two layers, these values were doubled.

Results of the FD simulations

As an indicator of the efficiency of flow diversion, the average flow velocity within the sac was computed (Fig. 5). In the HA aneurysm, the average flow velocity was reduced by 62% following deployment of a single FD and by 75% after deployment of a second layer. In the LA aneurysm, a single layer resulted in 50% and the second layer in 56% velocity reduction, demonstrating a lower efficiency associated with LA and reduced influence of the second layer as compared to the first.

To study the spatial distribution of FD induced WSS and pressure changes over the aneurysm wall, the pre- and post-treatment WSS and pressure maps were subtracted from each other. Figure 6 demonstrates the relative change in WSS after inserting one FD into the parent artery of the HA and LA aneurysm at the systolic instant.

As a result of accelerated flow inside the vessel, increased WSS was seen within the parent artery downstream of and across the neck of the aneurysm. Although, in general, the WSS decreases on the aneurysm sac, small elevated regions of WSS may appear on the sac after FD implantation. In the systolic peak, the average WSS on the sac decreased by 45% and 40% in case of the HA and the LA aneurysm with a single FD, respectively.

### Table I

| Thickness | $K$ [$\mu m^2$] | Linear loss coefficient ($\mu/K$) [kg mm$^2$/s$^2$] | Quadratic loss coefficient ($B$) $[1/\mu m]$ |
|-----------|----------------|-----------------------------------------------|-------------------------------------------|
| Real (31.75 $\mu m$) | 179.48 | 16.715 | 1.6267 |
| Simulated (200 $\mu m$) | 1130.58 | 2.6535 | 0.2582 |

Fig. 5. Average velocity in the systolic peak flow instance inside the aneurysms without and with one or two FD layers as a ratio to the average velocity before FD introduction.
The pressure only decreased slightly, by less than 2% in both cases. As the magnitude of any pressure changes was insignificant over the aneurysm wall, no pressure distribution map is shown here.

**Discussion**

We presented a method to study the effects of the flow diversion treatment technique involving densely woven stents referred as FDs. This technique has a growing interest among physicians due to promising results of the first clinical experiences [2]. In case of large aneurysm bulges, the technique proves to be an effective treatment method.

We studied these medical devices both experimentally and numerically. The key factor in the accurate simulation of the flow in the vicinity of FDs is the accurate determination of the HR. The function of an FD is to put a resistance in the neck of the aneurysm to hinder the communication between the flow in the parent artery and the aneurysm sac, since the energy supply of the flow in the aneurysms sac comes from the parent artery. This is achieved by the flow resistance posed by the FD that effectively creates a pressure loss between parent artery and the aneurysm sac, and as a consequence, the flow slows down in the aneurysm sac leading to thrombosis. The physical essence of this process is the flow resistance or HR.

To this end, one can follow three paths: to determine the HR experimentally or to find an equation on the basis of an analogous flow in the literature or to simulate the flow around the exact stent geometry [9]. The first approach is more accurate in case of laboratory experiments, but without measurements, the second can be more practical. Determination of the permeability using CFD technique is with questionable benefits since, with the same effort, one might carry out an accurate modeling of the stent structure.

The presented novel technique to measure the HR of an FD by measuring pressure drop over the thickness of the FD as a function of the flow rate yields a parabolic curve crossing the origin. This is in concordance with the numerical findings of Augsburger et al. [9]. Two parameters determine these curves completely which could be used as a general characterization of FDs and may help compare existing models, as well as serve as aids in the development phase of new FDs. It is actually somewhat surprising to have found a parabolic relationship. If the velocity is low, its square is second order small so that it is negligible. The quadratic term cannot be deemed a priori negligible because it depends on a lot of additional factors. Fluid mechanical intuition would have suggested that the velocity is low enough but experimental evidence proved clearly that this was not the case.

Using two FDs involves a random factor, namely, their relative position. Having performed the same measurement several times, we found that the mean HR curve of two layers is twice of that of one layer. In a simplistic way of thinking, one might consider a layer as one hydraulic resistance and the second layer as an identical resistance in series with the first one. Analogously to electrical circuits, the total resistance is simply the sum of both. In many cases in fluid mechanics,
however, this way of thinking can be misleading since the flow distortion caused by one resistance influences the other resistance both backward and forward (in upstream and downstream direction). Our measurements suggest, however, that the mean resistance of the two layers is exactly the sum of the two resistances, that is, the two resistances can be viewed as completely independent of each other.

If the measurement of the FD resistance is not possible, a linear approximation can be used, neglecting the quadratic term in Eq. (1). In this case, the permeability of porous structures can be estimated as follows [17]:

\[ K = \frac{D^2 \phi^2}{\varepsilon (1 - \phi)^2}, \]

where \( \phi \) is the porosity (the ratio between the empty and the whole volume), \( D \) is the characteristic size of the porous structure (which is the mean wire diameter in our case), and \( \varepsilon \) is a constant. This equation expresses the influence of the porosity on the permeability proportionally to the specific surface (ratio of exposed surface of the structure to the bulk volume) of the porous volume. \( \phi \) and \( D \) can be obtained from manufacturing data and \( \varepsilon \) can be calibrated with measurements. In our case, \( \phi = 0.7 \), \( D = 32 \mu m \), and the calibration yielded \( \varepsilon = 21.46 \).

The limitations of the formula were studied, and it was found that neglecting the second order term of the HR of the FD causes 20% or less deviation from the original HR if the velocity of the blood flow through the FD is under 0.06 m/s. Additionally, for velocity values under 0.02 m/s, the deviation of the measured and the linear approximation of the HR is less than 10%. The ratio of the through-flow velocity magnitudes lower than 0.02 m/s and lower than 0.06 m/s to all the velocity values evaluated both in time and space on the exposed FD surface is given in Table II, for both HA and LA cases.

Table II shows that, with one stent, a large aneurysm with HA, 60% of all perpendicular velocity magnitudes appearing on the layer surface during the whole cardiac cycle are below 0.02 m/s and 91% of the velocity values are below 0.06 m/s. The same aneurysm with two stents shows even higher percentages resulting in a less than 20% error of the linear approximation in 95% of all the occurring velocity variations. With this associated deviation from the measured HR of stent being relatively low in a dominant portion of the velocities in HA cases, the second order effects might be neglected. In the best LA case, 20% or larger error is associated with more than 35% of the occurring velocities, which is too high to assume only the permeability of the stents.

Subsequently, the linear approximation is an acceptable method if permeability measurements are not available. However, it may introduce a non-negligible error into the porous material simulation of FD effect on LA aneurysms.

In order to predict the effect of flow diversion on individual aneurysms, we developed a technique to simulate the effect of woven FDs. We used the porous material technique because woven wire FDs have a complicated thin structure that makes their geometrically accurate modelling extremely cumbersome, probably without a significant benefit. Appanaboyina et al. [6] showed that the positioning of the stent yields no or little difference in the exact modelling; therefore, the overall HR is important. Furthermore, the scale of the FD wires is in the order of the surface roughness of the vessel walls due to the endothelial cells, which is neglected along other fluid dynamical effects in this scale arising from the inhomogeneous nature of blood at small scales.

Besides, in case of applying multiple layers, the random orientation of FD struts relative to each other is impossible to simulate accurately. To mimic the effect of random orientation, we repeatedly assembled and dismantled the two FDs in the experimental apparatus. In the patient-specific aneurysm models, the mean values of five different HR calculations were used. The porous layer simulation complemented with accurate laboratory measurements produced flow conditions similar to those reported on clinical angiographic studies [2]. The results are also very similar to those obtained with more time-consuming simulation techniques [18].

Our porous layer FD simulation study confirmed that, by using FDs, significant intraaneurysmal flow reduction could be achieved in HA aneurysms. In the LA aneurysm, the flow velocity reduction was less significant. The numerical results show that the first inserted FD renders the internal flow significantly slower and a second layer produces just a fraction of the change caused by the first one. It was also found that, in the HA aneurysm, the relative effect of the FD implantation is more significant than in the LA case. All of these factors should be considered while planning FD treatment for individual aneurysms.

Delayed rupture following flow diverter treatment has been reported. In special vascular geometries, an increase of intraaneurysmal pressure was shown as a result of flow diversion and was claimed to cause the rupture [18]. In a larger series of such complications, besides the detrimental effect of intraaneurysmal thrombosis, they raised the possible role of remaining jet flow into the aneurysm sac.
after flow diverter treatment [19]. In Fig. 6, the small area of increased WSS after placing a single layer of FD across its neck corresponds to this latter phenomenon magnified by the relative representation. This highlights the need of effective flow diversion, either by placing more FD layers (as it was done in this case) or placing coils within the aneurysm simultaneously with flow diversion.

**Conclusion**

Flow diversion for aneurysms can be simulated on patient specific FV models using CFD with a porous approach. This technique is more economic compared with the accurate geometrical simulation of FDS but requires the knowledge of the HR of the FD to be used. We developed a novel technique to measure the HR of FDS and found a quadratic relationship between the pressure drop and the flow rate. This also means that an arbitrary FD can be characterized by two (in case of low velocities only one) parameter(s). The mean HR of two layers out of several trials was found twice as much as that of one layer. In case no measurement apparatus is available, a linear relationship can be used that provides a good approximation for low velocities but becomes less reliable for higher velocities. In patient-specific FV models, flow reduction was found to be more significant in an HA aneurysm as compared to LA one. A second FD layer causes significantly less flow reduction than the first layer. Besides the overall reduction of flow velocity and WSS within the aneurysm sac, focal increase of WSS and pressure may occur over the aneurysm wall after treatment. The technique described here may prove useful in both treatment planning and in designing new FD devices.

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