3D Modeling of Flows in Intracranial Bifurcating Arteries with Saccular Aneurysms

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I. Abstract

Intracranial bifurcating arteries are sites of Saccular aneurysms. The disorder may result from congenital defects or from other conditions such as high blood pressure, atherosclerosis and less commonly, head trauma or infection. Once the aneurysm is formed, it is likely to rupture and can lead to a stroke. In this report, we first describe the development of an idealized 3D finite element model of a bifurcating artery with a saccular aneurysm using SolidWorks. We then mesh this model, using ICEM to be able to simulate its actual physiological working. A study of the hemodynamics of the model for visualizing important parameters like the velocity, pressure, etc. (in Tecplot) was performed using Fluent. Tecplot was used to analyze convergent data from Fluent in order to draw conclusions about the various hemodynamic parameters for our model. The same procedure was followed for the treated model, with a stent placed very close to the aneurysm neck to be able to divert the blood flow directly into either of the output vessels and bypass the aneurysm to a large extent, thus preventing hemorrhage. We also discuss the future possibilities in the field of using stents to ameliorate problems related to saccular aneurysms in bifurcating cranial arteries.

II. Introduction

Saccular intracranial aneurysms (IAs) account for 90% of the aneurysms that occur at bifurcations of the circle of Willis and its proximal branches (Figure 1). Saccular (or berry) aneurysms display a variety of complex sizes and shapes. The major complication of IAs is their rupture, which causes Subarachnoid Haemorrhage (SAH). The incidence of IAs has been estimated to be between 1% and 8% among western populations, with the majority of IAs being dormant over life. There is a 70-90% mortality rate in people with this kind of aneurysm. When an IA is diagnosed and a patient needs to be protected from aneurysm rupture or re-bleeding, a number of treatment options are considered to isolate the aneurysm from the cerebral circulation and to re-establish a physiological flow passage into the parent artery. These include either open surgery and clipping of the aneurysm, or endovascular treatment (EVT), performed primarily by coil embolization. Surgical treatment involves craniotomy and may be associated with vasospasm, infection, brain edema and higher health care costs. EVT is minimally invasive and involves the deployment of platinum coils into the aneurysm after the placement of a microcatheter guided through the arterial system. The objective of EVT is to shield the aneurysmal wall and reduce the blood flow into the aneurysm sac, thus progressively inducing aneurysmal flow stasis, thrombus formation and aneurysm occlusion [1].

Although aneurysms have been the target of numerous studies, precise quantitative data on the aneurysm geometry, including neck size, dome diameter, and other shape factors, remain limited. This geometric knowledge is potentially important for patient management when decisions concerning therapy must be made (eg, surgical clipping or endovascular coiling). An understanding of the geometry of aneurysms is also important for studies that seek basic information about the pathophysiology of the disease, including

Figure 1: Saccular aneurysm in an intra-cranial bifurcating artery
approaches that use computer modeling, biomathematical models, in vitro experimentation with vascular phantoms, and in vivo animal models. The need for accuracy in modeling is imperative since the results obtained are only as valid as the models themselves. Furthermore, in vitro models based on accurate geometric information are essential in the refinement of imaging techniques, for use as training tools, or for testing new therapeutic procedures, all of which have the potential to improve diagnosis and treatment [2].

Although widely accepted and increasingly used, coil embolization of IAs is not a risk-free procedure either, and may be associated with coil compaction, recanalization and aneurysmal regrowth, necessitating re-treatment. In addition, efficacy of coil embolization is limited in wide-neck aneurysms, which are also difficult to treat surgically. Intracranial stenting may be used to aid aneurysm coiling (stent-assisted coiling) providing a scaffold to hold the coils inside the aneurysm sac while being deployed and thus achieve higher packing density [1].

While coil support is the main concern during stent-assisted coiling, several studies have shown that changes in aneurismal fluid mechanics are induced by stent placement alone. Using particle image velocimetry (PIV), up to a 64% reduction was measured in intra-aneurysmal velocity magnitudes in three anatomical sidewall aneurysm models following Neuroform stent deployment. Similar reductions were measured following a Y configuration deployment in a realistic cerebral aneurysm model. Similarly, reductions were also seen after simulating multiple Neuroform stent-in-stent deployments in anatomical sidewall aneurysm models. Researchers have simulated half-Y and Y deployments in three anatomical bifurcation models and found large differences in aneurysmal velocity magnitudes between the configurations. This shows that performing multiple simulations would allow us to choose the best configuration, but this is very computationally intensive as well as time consuming [3].

It is assumed that, depending on their design, stents will also cause a reduction of the intra-aneurysmal flow. This aspect has recently gained more interest as stents can be used alone as flow diverters. Stents are flexible, self-expanding or balloon-expandable porous tubular meshes made of stainless steel or other alloys such as Nitinol. They are expanded into the arterial lumen across the aneurysmal orifice to deviate blood flow away from the aneurysm sac. Ideally, stents should provide sufficient haemodynamical resistance to aneurysmal inflow and outflow to promote thrombosis and stabilization without the need of accessing the fragile aneurysmal cavity. Preliminary clinical experiences with stent-based flow diversion have reported progressive aneurysmal thrombosis induced by single or double stenting for wide-neck aneurysms and complex vertebrobasilar aneurysms [1].

Parent vessel geometry and stent design have been suggested to play an important role in the reduction of aneurysmal flow intensity and in the potential success of IA treatment. Extensive analyses of aneurysmal haemodynamical changes induced by stent deployment have been provided by both in vitro and numerical studies. Stent-induced aneurysmal rheological changes, characterization of aneurysmal flow activity reduction due to a stent and influence of stent porosity on aneurysmal flow have been described for idealized aneurysm models. The influence of parent vessel curvature has also been analyzed. When the same idealized aneurysm model was considered, straight and curved parent vessels provided significant differences in the potential of thrombotic occlusion created by a specific stent. These observations suggest the importance of considering patient-specific anatomy to assess the performance of a specific stent design and study its relation with clinical events and post-implant complications. Advanced image processing and geometrical modeling techniques have been recently combined with computational fluid dynamics (CFD) approaches to generate detailed haemodynamical descriptions in patient-specific anatomical models.
The method adopted in this work was purely geometrical and was intended to provide realistic stent porosity. For this purpose, we used a Neuroform stent, which is the most frequently deployed stent geometry (with the traditional 8 crowns). Improvements could include the addition of stent and vascular mechanical properties and would probably require a statistical analysis, modeling the uncontrolled initial stent orientation before deployment. This is due to the fact that current stent designs are not optimized to act as flow diverters and their performance may depend on a correct positioning during intervention. Novel stent designs aim at providing the same flow diversion irrespective of the actual positioning before deployment. Nevertheless, stenting has been reported to reduce overall Wall Shear Stress and flow activity in all cases, potentially favoring wall remodeling and reducing the risk of coil compaction and aneurysm re-growth. This report concentrates more on the velocity and pressure changes than the wall shear stress and our goal has been to try and replicate results from literature.

In our model we use straight parent vessel and curved outlet vessels to allow for the flow to develop well enough. The model used is an actual anatomical model constructed based on a CT image of an aneurysm in a bifurcated cranial artery.

### III. Materials and Methods

Initial work started with the creation of an idealized bifurcating artery saccular aneurysm model. Creation of the geometries was carried out in SolidWorks (A 3D mechanical CAD program developed by Dassault Systèmes SolidWorks corp.) by using various features within the program. The dimensions of the idealized model were made possible by the previous work carried out at the Image Processing Applications Laboratory at Arizona State University. The dimensions for the primitive model are given as below:

<table>
<thead>
<tr>
<th>Model Feature</th>
<th>Dimension in millimetre (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Inlet Vessel Diameter</td>
<td>4 mm</td>
</tr>
<tr>
<td>Outlet Vessel Diameter</td>
<td>4 mm</td>
</tr>
<tr>
<td>Aneurysm Dome Diameter</td>
<td>4 mm</td>
</tr>
<tr>
<td>Aneurysm Neck Diameter</td>
<td>2 mm</td>
</tr>
<tr>
<td>Length of the inlet vessel</td>
<td>95 mm</td>
</tr>
<tr>
<td>Length of the Outlet Vessel</td>
<td>60 mm</td>
</tr>
<tr>
<td>Angle between the Parent Vessel Axis and Outlet Axis</td>
<td>60 degrees</td>
</tr>
</tbody>
</table>

Table 1: Table showing the dimensions of each of the model features in our initial model

A few assumptions to note about the created model are that the compliance of the vessels is not taken into account, and the models created in SolidWorks are of a rigid nature for the sake of simplicity. Also the models are representative of a hollow closed network of cylindrical geometries that are brought together to create an idealized structure. The model we created in SolidWorks is shown in Figure 2:

The next process among several sequential ones is the creation of mesh
elements within the volume and along the walls of the SolidWorks model. The meshing utility that we made use of is ICEM (an ANSYS package). The importance of creating mesh elements before exporting the model into Fluent is the fact that Fluent converges to a solution after calculating the flows and pressures within each fluid volumetric element. The subdivision of the entire volume within the model to very small elements defines the meshing process and is based on giving a few input parameters to ICEM. The parameters that we made use of were an initial seed size of 0.18 mm and the meshing elements as all triangular shaped. The mesh was created with total element number as 6.1 million. Upon loading the mesh geometry into Fluent to simulate blood flow, we realized after successive attempts that our model geometry that was created in SolidWorks initially was irregular and not conducive for further analysis as it would have not led us to converging results.

We made a few changes, and got our model feature dimensions correct as per the ratios defined in [2]. In order to make the changes we had to get back to SolidWorks and get our dimensions correct. Some of the ratios that were adjusted were the D/N (Dome diameter/Neck width ratio), H/N (Dome Height/Neck Width), D/H (Dome diameter/Dome height) and H/S (Dome Height/semi-axis height) ratios. On completing the changes as decided, we could observe that our model was much smoother with regards to fluid interaction and this was verified by the members in the IPA (Image Processing Applications) lab as well.

The modified geometry was meshed again (7 million elements, shown on the right in Figure 3) and we ran blood flow simulation in Fluent using the following parameters. The blood density was set at 1060 kg/m$^3$ and the viscosity as 3.71E-3 kg/m-s. These are the acceptable constants associated with fluid dynamics of blood flow as described in [4]. While meshing the geometry we had to specify the flow inlets and outlets. In our model we have one flow inlet which further bifurcates to give us two flow outlets. During simulation the inlet flow was set at 0.1592 m/sec and the operating pressure was set at 0 pascals (Pa) [5]. Model convergence was defined at 1e-5 m/sec of continuity, x-velocity, y-velocity and z-velocity. This means that the solution would be considered to be reached if the difference between residual velocities between any two iterations is below $10^{-5}$ m/sec.

We were able to observe a definite

![Figure 3: Modified geometry with mesh elements shown in ICEM](image)

![Figure 4: Scaled residuals depicting convergence.](image)
solution to the flow problem in the untreated aneurysm model and got the residual velocity plots as shown in Figure 4. We wanted to investigate the alteration in flow characteristics with the introduction of a stent near the neck of the aneurysm. We were able to design the 8-crown neuroform stent using SolidWorks, and able to unite the two models (Stent & Untreated model) together using a software package called GeoMagic 2010. This software allowed us to modify the stent in a manner that it fit perfectly in the cylindrical lumen in an area around the neck. This united model is hereafter referred to as the treated model in our report. The stent had to be specially modified and altered geometrically (Using GeoMagic) after its initial design in SolidWorks in order to ensure that it did not protrude out from the vessel geometries as this would give us errors in subsequent simulations when we meshed each of the objects using ICEM. For the treated model (model + stent) we had to mesh the individual objects differently, and used a seed size of 0.1 mm for meshing the surface & volume of the stent within the model so that we would be able to observe the interaction of blood with the stent in better resolution. The model geometry was meshed with the same parameters as those used for the untreated model. Figures 5 & 6 below depict the modifying process of the stent and the stent within the model geometry respectively.

Solution convergence was ensured and we got the results for each of the treated and untreated models and they are presented in the results section of our report. It is important at this point to note that Fluent uses the Navier Stokes Equation to solve for solutions to first order and second order differential equations in order to solve for both the pressure and the velocity of each of the x, y & z components within each volumetric element (Please refer to Item 1 in Appendix-2).

We have extensively made use of Tecplot (Version 360, 2009), a data plotting and visualization software that has the capabilities of processing large amounts of data and showing it in 3D space. Saved data from the fluent simulations was loaded into Tecplot. We calculated the Velocity magnitude vectors for the entire 3D space from within Tecplot and were able to represent them as results. Tecplot has the capability to show data traces (Tracking how a particular element of data moved along the model with time) and this proved to be one tool which helped us a lot in making sense of the Fluent simulations. We present various graphs and figures from Tecplot in the proceeding sections of this report.
IV. Results
A. Untreated Model

Figure 7: Pressure plot of the untreated model as obtained from Tecplot 360.

Figure 8: Velocity magnitude plot of the untreated model, as obtained from Tecplot 360.

Figure 9: Visualizing the Flow Pattern using an arrowhead and line diagram.

Figure 10: Visualizing flow pattern in only one output vessel using a slightly different perspective.
B. Treated Model

Figure 11: Velocity Magnitude plot of Treated Model (left) Vs. Untreated Model (right)

Figure 12: Alternate slice view of the velocity magnitude showing function of stent as checker/diverter of blood flow
V. **Discussions and Conclusions**

A. Untreated Model

After meshing and simulating the fluid dynamics in the model using ICEM and Fluent respectively, as explained in the previous section, the graphical results were interpreted in Tecplot 360 (2009). All the figures shown above, in the results section, were obtained by using a cut plane view of the aneurysm (to visualize flows and sidewall interactions better), the front/back plane view (along the z-axis) of the model is visible in these figures.

Figure 7 is a pressure plot of the untreated model, as obtained from Tecplot. As can be seen and what is pretty much intuitive from the legend at the side of the figure is that there is high pressure at the parent vessel inlet (110 Pascal) and this pressure gradually eases out as the blood traverses the length of the vessel. This continues until it reaches the neck of the aneurysm, where, as we can see, the pressure increases again, owing to interaction with the not so smooth i.e. irregular structure of the aneurysm. The turbulence that we see in the pattern of pressure (denoted by the black curves/lines)
around the aneurysm is owing to this interaction of the blood with the aneurysm neck and the initial part of the outlet vessel sidewalls. As we can see, the pressure pattern eases out again and reaches a steady, low level of 10 Pa after it travels some distance along the two outlet vessels.

Figure 8 is a velocity magnitude plot for the untreated model, which shows that the velocity at the inlet vessel (a boundary condition set in Fluent at 0.1592m/sec, as previously mentioned) increases in the beginning along the length of the parent vessel. It then gradually changes its course and hence reduces in magnitude, as it gets split into two streams at the bifurcation. We also see that the blood interacts with the aneurysm and has a kind of circular flow pattern through it (as is seen in Fig. 9). It is also observed that the velocity magnitude decreases from the initial part (neck) of the aneurysm to its circumference (notice change in color corresponding to the scale on the legend). This means that it is safe to assume that the blood is less turbulent; however, as seen from the increase in pressure in Fig. 7, it is still capable of causing a significant amount of wall shear stress in the aneurysm bulb, owing to its comparatively smaller and more restricted area of flow. This finding is consistent with the literature in [3].

Figure 9 shows a flow pattern of blood with an arrowhead (to depict direction of flow) and lines diagram. This clearly shows the entry of blood from the inlet parent vessel and then its splitting into two separate streams that lead to the outlet vessels. The pattern of blood flow to the aneurysm bulb is very interesting since it goes into the vessel (with the velocity magnitude decreasing all the time) and then circulates and flows out of the aneurysm and into the two outlet vessels and essentially joins the main path of blood flow at the bifurcation.

Figure 10 again shows the flow pattern, but from a different perspective, which allows us to view the flow from the inlet to the aneurysm and finally to only one output vessel. Again, notice the circular pattern of blood flow and the decrease in velocity magnitude in the aneurysm bulb.

B. Treated Model

The treated model with the Neuroform stent in place at the bifurcation and oriented towards one of the outlet vessels can be seen in the Velocity Magnitude plot in Figure 11, which also shows the untreated velocity magnitude plot for better comparison. As can be clearly seen from this figure, the model to the left, i.e. the treated model, definitely shows interaction of the blood with the stent and a consequent diversion of the blood away from the aneurysm. Notice that in the untreated model to the right, the blood spreads uniformly over the area of the stent, albeit with a decrease in velocity magnitude. In the treated model however, we can see that the blood does not spread uniformly and in fact there is a considerably lesser amount of blood flowing into the aneurysm bulb as compared to the untreated model. We are of the opinion that the blood gets diverted into the (left) output vessel, through interaction with the stent and on careful observation, we can also see that the far edge of the stent (which is closest to the right output vessel) in fact does not allow blood to enter the aneurysm (or allows minimal blood flow) and diverts all the blood trying to get into the aneurysm on the right hand side, towards the right output vessel, thus reducing wall shear stress considerably.

A considerably lower flow of blood is seen getting through the stent and going into the aneurysm bulb towards the left hand side. This blood again follows a circular pattern and goes back to the left outlet vessel.

Figure 12 shows a different slice of the model and comprehensively backs our findings from the previous figure. As can be seen, the stent definitely reduces the amount of blood flowing into the
aneurysm. What can also be seen here is that after interaction with the stent, blood flowing towards the left output vessel reduces in magnitude and once it is away from the stent, it starts picking up in magnitude until it becomes constant again. This may be an indication of the way the stent interacts with the blood (as a flow checker/diverter).

Figure 13 shows the arrowhead and line flow pattern of the blood in the treated model (left and right), again, in comparison to the untreated model (lower bottom). Here, as can be seen, the number of lines of flow i.e. their density is much lesser in the treated model as compared to the untreated model, again underlining the finding that the stent does help in checking blood flow and diverting it away from the aneurysm and into the output vessels. Please note that the left and right figures are left and right half representations of the flow patterns from the parent vessel into the left and right output vessels respectively. That is the reason why we can visualize only partial flows in the contralateral vessel.

From the preceding discussion, it would be safe to conclude that the stent does help to reduce blood flow into the aneurysm and consequently in reducing the wall shear stress (although not measured directly) in the aneurysm bulb (and neck). This in turn would allow us to reduce chances of hemorrhage. There is the possibility that viewing from different perspectives or different slices may distort our reading of the results, however, we do not think it would grossly affect our findings.

We also believe that different orientations of the stent would allow us to view which position of the stent may be best suited as a treatment measure. Also, using concentric stents / successive stents along the neck or one output vessel may lend more strength and efficacy to the stent when it comes to diverting blood flow [3]. This is something that can be pursued as a separate project. We need to keep in mind however, that the correct level of flexibility needs to be maintained so that the artery can continue to maintain its structure and allow for free flow of blood through it.
Bibliography:


Appendix – 1

A few figures placed in the Appendix are not presented in the report or are presented in the report in low quality.

**Figure 5: Meshing of the stent with a seed size of 0.1 mm**

**Figure 6: Our modified model showing the mesh with approximately 7 million triangular elements.**
Figure 7: Scaled residuals of velocity and continuity for the untreated model.

Figure 8: Scaled residuals of velocity & continuity for the treated model (not presented in the text above)
Figure 9: Streamtraces in Tecplot showing uniform flow in our untreated model. Symmetric Geometry.

Figure 10: Left half input flow depicting lower flow into the aneurysm. The velocity magnitude within the aneurysm is much lower as compared to Figure 5.
Figure 11: Right half input flow depicting very low velocity flows in the aneurysm region.

Figure 12: Velocity Magnitude contour corroborating partial diverted flow towards the outlet vessel devoid of stent.
Figure 13: Meshed geometry for our treated model. The stent placement can be seen within the cylindrical vessels and the neck region.
1) We present the general form of Navier Stokes Equation that is used by the Fluent solver to resolve Pressure and Velocity components (X, Y & Z directional) for each volumetric element within the model (nearly 7 million such elements for our revised model discussed in Methods & Materials)

\[\rho \left( \frac{\partial v_x}{\partial t} + v_x \frac{\partial v_x}{\partial x} + v_y \frac{\partial v_x}{\partial y} + v_z \frac{\partial v_x}{\partial z} \right) = -\frac{\partial P}{\partial x} + \rho g_x + \mu \left( \frac{\partial^2 v_x}{\partial x^2} + \frac{\partial^2 v_x}{\partial y^2} + \frac{\partial^2 v_x}{\partial z^2} \right)\]

\[\rho \left( \frac{\partial v_y}{\partial t} + v_x \frac{\partial v_y}{\partial x} + v_y \frac{\partial v_y}{\partial y} + v_z \frac{\partial v_y}{\partial z} \right) = -\frac{\partial P}{\partial y} + \rho g_y + \mu \left( \frac{\partial^2 v_y}{\partial x^2} + \frac{\partial^2 v_y}{\partial y^2} + \frac{\partial^2 v_y}{\partial z^2} \right)\]

\[\rho \left( \frac{\partial v_z}{\partial t} + v_x \frac{\partial v_z}{\partial x} + v_y \frac{\partial v_z}{\partial y} + v_z \frac{\partial v_z}{\partial z} \right) = -\frac{\partial P}{\partial z} + \rho g_z + \mu \left( \frac{\partial^2 v_z}{\partial x^2} + \frac{\partial^2 v_z}{\partial y^2} + \frac{\partial^2 v_z}{\partial z^2} \right)\]

Figure 1: Generalized form of the Navier Stokes equation for each X, Y & Z co-ordinates.

The left hand side terms are the material terms of the equation and relate to the density and forces generated due to the viscous nature of the fluid. The first term on the right relates to the pressure of the fluid. The last term on the right indicates the forces generated due to the shear stresses and viscous drag forces.