COMPUTATIONAL HEMODYNAMIC STUDY OF ENDOVASCULAR STENTING
IN PATIENT-SPECIFIC CEREBRAL ANEURYSMS

by

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To my immediate and extended family (my friends)
Acknowledgments

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# Table of Contents

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>List of Tables</td>
<td>vii</td>
</tr>
<tr>
<td>List of Figures</td>
<td>viii</td>
</tr>
<tr>
<td>Abstract</td>
<td>xi</td>
</tr>
<tr>
<td>1 Introduction</td>
<td>1</td>
</tr>
<tr>
<td>1.1 Brain Vasculature</td>
<td>1</td>
</tr>
<tr>
<td>1.2 Stroke and Aneurysms</td>
<td>2</td>
</tr>
<tr>
<td>1.3 Diagnosis</td>
<td>4</td>
</tr>
<tr>
<td>1.4 Treatment</td>
<td>5</td>
</tr>
<tr>
<td>1.5 Modeling</td>
<td>5</td>
</tr>
<tr>
<td>1.6 Image-Based Patient-Specific Endovascular Stent Simulations</td>
<td>7</td>
</tr>
<tr>
<td>1.7 Original Pipeline</td>
<td>8</td>
</tr>
<tr>
<td>1.7.1 Image Processing</td>
<td>8</td>
</tr>
<tr>
<td>1.7.2 Hemodynamic Modeling</td>
<td>10</td>
</tr>
<tr>
<td>1.7.3 Boundary Conditions</td>
<td>12</td>
</tr>
<tr>
<td>1.7.4 Post-processing and Visualization</td>
<td>13</td>
</tr>
<tr>
<td>1.8 Summary</td>
<td>14</td>
</tr>
<tr>
<td>2 Embedded/Immersed Methods</td>
<td>15</td>
</tr>
<tr>
<td>2.1 Introduction</td>
<td>15</td>
</tr>
<tr>
<td>2.2 Methods</td>
<td>16</td>
</tr>
<tr>
<td>2.2.1 The Embedded Unstructured Grid Method</td>
<td>17</td>
</tr>
<tr>
<td>2.2.2 Immersed Unstructured Grid Method</td>
<td>22</td>
</tr>
<tr>
<td>2.2.3 Treatment of spheres</td>
<td>23</td>
</tr>
<tr>
<td>2.3 Results</td>
<td>25</td>
</tr>
<tr>
<td>2.3.1 Flow past a cylinder</td>
<td>25</td>
</tr>
<tr>
<td>2.3.2 Idealized stented aneurysm</td>
<td>29</td>
</tr>
<tr>
<td>2.4 Summary</td>
<td>32</td>
</tr>
<tr>
<td>3 Patient-Specific Stenting</td>
<td>34</td>
</tr>
<tr>
<td>3.1 Stent deployment</td>
<td>34</td>
</tr>
</tbody>
</table>
3.2 In-vivo evaluation ........................................... 38
3.3 Effects of stent positioning .................................. 42
3.4 Partial stent modeling ....................................... 44
3.5 Summary ...................................................... 47

4 Flow Alteration Analysis ...................................... 48
  4.1 Effects of Stent Design ..................................... 48
  4.2 Effects of Treatment Options ................................. 53
  4.3 Quantitative Methods ....................................... 59
  4.4 Summary ...................................................... 63

5 Flow alteration in side branches ............................... 64
  5.1 Boundary Condition Methods ................................. 64
  5.2 Perforators .................................................. 65
  5.3 Side branches ................................................ 71
  5.4 Summary ...................................................... 74

6 Conclusions ..................................................... 75
Bibliography ........................................................ 77
### List of Tables

<table>
<thead>
<tr>
<th>Table</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.1</td>
<td>Drag coefficient for $Re = 25$</td>
<td>26</td>
</tr>
<tr>
<td>2.2</td>
<td>Comparison of velocity reduction between body-fitted and embedded.</td>
<td>32</td>
</tr>
<tr>
<td>3.1</td>
<td>Mesh size and computation time taken to simulate flow with the full and partial stents.</td>
<td>44</td>
</tr>
<tr>
<td>4.1</td>
<td>Quantitative values for the 3 models. Model 1:figure 4.8, left column; Model 2:figure 4.8, middle column; Model 3:figure 4.8, right column.</td>
<td>60</td>
</tr>
</tbody>
</table>
List of Figures

<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.1 Circle of Willis.</td>
<td>2</td>
</tr>
<tr>
<td>1.2 A 3DRA image.</td>
<td>4</td>
</tr>
<tr>
<td>1.3 Original and the new extended pipeline.</td>
<td>8</td>
</tr>
<tr>
<td>2.1 Modeling of vessels using the body-fitted approach and stents using the embedded approach: the body-fitted grid and the embedded stent (left), mesh refinement around the device (center) and the first order approximation of the stent (right).</td>
<td>16</td>
</tr>
<tr>
<td>2.2 First-order (left) and second-order (right) treatment of embedded surfaces.</td>
<td>18</td>
</tr>
<tr>
<td>2.3 Bin storage for faces and search.</td>
<td>19</td>
</tr>
<tr>
<td>2.4 Cut edge fraction.</td>
<td>21</td>
</tr>
<tr>
<td>2.5 Extrapolation of velocity.</td>
<td>21</td>
</tr>
<tr>
<td>2.6 Kinetic treatment of embedded surfaces.</td>
<td>23</td>
</tr>
<tr>
<td>2.7 Stent geometry represented as a surface triangulation (left) and a series of overlapping spheres (right).</td>
<td>24</td>
</tr>
<tr>
<td>2.8 Treatment of embedded spheres.</td>
<td>24</td>
</tr>
<tr>
<td>2.9 Grids used for the flow simulation. From top to bottom: body-fitted grid, uniform mesh with no refinement, mesh with 2 levels of refinement and mesh after 4 levels of refinement.</td>
<td>27</td>
</tr>
<tr>
<td>2.10 Velocity contours for the body fitted grid (top), the embedded grid with four levels of refinement (center) and the embedded grid with six levels of refinement (bottom).</td>
<td>28</td>
</tr>
<tr>
<td>2.11 Embedded grid model (top row) and body-fitted model (bottom row).</td>
<td>30</td>
</tr>
<tr>
<td>2.12 Velocity contours (left column) and streamlines (right column), before stenting (top row), body-fitted (middle row) and embedded (bottom row). The flow is from right to left.</td>
<td>31</td>
</tr>
<tr>
<td>2.13 Position of the lines and their respective plots.</td>
<td>32</td>
</tr>
</tbody>
</table>
3.1 Steps followed in the vascular deployment of stents: the 3D rotational angiography image (top row, left column), reconstructed anatomical model (top row, right column), construction of the skeleton of the vascular model (second row, left column), initial cylindrical surface (second row, right column), cylindrical surface in the vessel skeleton (third row, left column), final cylindrical surface (third row, right column), two stent designs (fourth row) and the stents after deployment (fifth row). 37

3.2 Flow pattern (left column) and wall shear stress distribution (right column) at peak systole before (top row) and after (bottom row) stenting. 39

3.3 Conventional (gray background) and virtual (white background) prior to stenting (left panel), right after stenting (center panel) and at a one week follow up exam (right panel). 41

3.4 Streamlines and WSS results for the original stent configuration (first and third row) and the rotated configuration (second and fourth row). 43

3.5 The full stent and the partial stent for the different aneurysms. 45

3.6 Streamlines and WSS results for the different aneurysms with full and partial stents. 46

4.1 Original 3DRA images, computational models and the three stents deployed into the vascular models. 50

4.2 Streamlines for the four aneurysm models, pre-stented case (top row), with Neuroform stent (second row), left helical stent (third row) and right helical stent (bottom row). 51

4.3 Wall shear stress for the four aneurysm models, pre-stented case (top row), with Neuroform stent (second row), left helical stent (third row) and right helical stent (bottom row). 52

4.4 Patient-specific vascular models of three patients. Model with three aneurysms (top left corner), close-up view of the two MCA aneurysms (top right corner), and two basilar tip aneurysm models (bottom row). 54

4.5 Endovascular stenting options. 55

4.6 Streamlines depicting the intra-aneurysmal flow patterns before and after the different treatment options. 57

4.7 Wall shear stress distributions before and after the different stenting options. 58

4.8 Steps in the aneurysm segregation process. 61
4.9 Models, streamlines and WSS results for the pre and post cases.

5.1 Perforator attachment to the parent vessel. Smooth connection (top), Sharp connection (bottom).

5.2 WSS in the model with the perforator.

5.3 Flow through perforator. Left column: funnel type; Right column: sharp edge; Top row: pre-stent; Bottom row: 30%-50% occlusion.

5.4 Flow reduction in perforator.

5.5 Flow through opthalmic. Left column: Model 2; Right column: Model 3; Top row: model with stent; Center row: pre-stent; Bottom row: 90% occlusion.

5.6 Flow reduction in Model 2.

5.7 Flow reduction in Model 3.
Abstract

COMPUTATIONAL HEMODYNAMIC STUDY OF ENDOVASCULAR STENTING IN PATIENT-SPECIFIC CEREBRAL ANEURYSMS
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Stroke is the leading cause of death after heart disease and cancer and the number one cause of long-term disability in America. About 80% of hemorrhagic stroke are produced by the rupture of cerebral aneurysms. Surgical clipping and coil embolization are the most common methods of treating these aneurysms. However, both these treatments have some limitations for wide neck aneurysms. Recently, there has been an increased interest in the use of stents as flow diverters. With an effective design, the blood flowing into the aneurysm can be disrupted thereby promoting intra-aneurysmal thrombus formation and thus preventing rupture. Modeling blood flow around these endovascular devices in intracranial aneurysms is important for designing better devices and to personalize and optimize endovascular stenting procedures in the treatment of these aneurysms. Numerous studies have been conducted in the past using animal models, patient-specific in-vitro models and idealized computational models. Nevertheless, all of them have significant limitations. The main disadvantage of using animal models is that they fail to replicate the variable anatomy of diseased human arteries. And the problem with in-vitro models is that they are not suitable for large population studies. Patient-specific image-based numerical models have demonstrated to be a fast, reliable and inexpensive way of simulating blood flow inside
these aneurysms. Moreover, studies using these models have the potential to replicate the exact anatomy of specific patients in order to connect specific hemodynamic factors to clinical events. Furthermore, large patient population study is also possible. A methodology was previously developed in the CFD Lab, GMU, to conduct patient-specific studies but without any endovascular devices. It includes image processing and segmentation algorithms, unstructured 3D grid generation, finite element solver for Navier-Stokes equations, rheological models and visualization techniques. However, the main difficulty in using these models for endovascular stent simulations lies in the generation of acceptable computational grids inside the blood vessels and around these devices. An adaptive embedded gridding technique originally developed for fluid-structure interaction problems at the CFD lab, GMU, tremendously simplifies this impediment. In this doctoral thesis the computational modeling pipeline has been extended to model patient-specific hemodynamics of stented cerebral aneurysms. The methodology was evaluated and demonstrated with a number of image-based models and different stent aneurysms and applied to the study of the effects of stents on the flow in aneurysms and side arterial branches.
Chapter 1: Introduction

This chapter lays the ground work or gives the information required to understand the problem addressed in this thesis and the attempted solution. First, some general information about cerebral aneurysms such as location of their common occurrence, problems associated with their rupture, their diagnosis and the two most preferred ways of treating them is given. Then, issues related to these treatments and why there is a sudden interest in using endovascular stents to treat these aneurysms is discussed. Attempt made by others to model such stent procedures, the issues related to them and why patient-specific simulations are the way to go is also discussed. Finally, the challenges associated with conducting patient-specific stenting simulations and the work done to come up with a new pipeline to make such simulations possible and realistic is explained.

1.1 Brain Vasculature

Brain, which is the most important organ in the human body, is the control center of the central nervous system and regulates virtually all human activity. Therefore its proper functioning is crucial for the overall well-being of the individual. Just like any other organ in the human body, the brain’s survival depends on the supply of oxygen. The internal carotid arteries (ICA’s) and the vertebral arteries (VA’s) are responsible for carrying blood to the brain. The ICA’s supply to the anterior part of the brain whereas the VA’s join to form the basilar artery (BA) which supplies to the posterior part of the brain. The anterior and posterior parts are not separate and are connected by a network of arteries. The constant supply of blood to the various parts of the brain is possible with the help of a remarkable structure at the base of the brain called the Circle of Willis. This, when intact is a circle of arteries connecting the ICA’s, anterior cerebral artery and anterior communicating artery
in the front and the posterior cerebral arteries, posterior communicating arteries and the branches of the basilar artery in the back. When one part of the circle or an artery supplying to the circle becomes narrowed or blocked, blood flow from the other vessels can preserve the circulation.

1.2 Stroke and Aneurysms

Stroke is the leading cause of death after heart disease and cancer and the number one cause of long-term disability in America[1–7]. There are two types of strokes - ischemic and hemorrhagic. Ischemic stroke occurs when there is a reduction in cerebral blood flow below a certain threshold due to blockage of arteries in the brain by blood clots or by the gradual build-up of plaque and other fatty deposits. Hemorrhagic stroke or subarachnoid hemorrhage (SAH) is caused when a blood vessel in the brain breaks leaking blood into the brain. The symptoms suffered by the patient depends on where the stroke occurs in the brain and how much the brain is damaged. Some of these symptoms are disturbance of language, unable to perform skilled motor acts, sensory loss, visual field loss, incoordination, hemiplegia, akinetic mutism, bladder dyscontrol, bowel dyscontrol, vertigo and loss of
About 80% of SAH are produced by the rupture of cerebral aneurysms[9], which is a cerebrovascular disorder in which weakness in the wall of a cerebral artery or vein causes a localized dilation or ballooning of the blood vessel[1]. The circle of Willis and its arterial bifurcations are a common location for the occurrence of these aneurysms and nearly 85% of them develop in the anterior part which involve the internal carotid arteries and their major branches that supply the anterior and middle sections of the brain. The most common sites include the anterior communicating artery (30-35%), the bifurcation of the internal carotid and posterior communicating artery (30-35%), the bifurcation of the middle cerebral artery (20%), the bifurcation of the basilar artery, and the remaining posterior circulation arteries (5%). SAH due to aneurysm rupture occurs in approximately 10 per 100,000 population, with 80% occurring in the 40 to 65 age group[9]. It has fatal consequences in 2.6-9.8% of the patients and serious consequences in 10.9% due to intracranial bruise, hydrocephaly the cerebral vasospasm and its ischemic consequences[10]. Only 32% are reported to lead a normal life after SAH[9].

The exact reasons for the genesis, growth and rupture of these aneurysms have not been clearly identified, however factors such as congenital or inherited defects weakening the arterial wall, hypertension, atherosclerosis and thrombosis, have been considered as the causative factors[10–15]. Also based on more indepth studies conducted arterial hemodynamics, wall biomechanics, mechanobiology and intracranial environment have been identified as the major factors responsible for the initiation, growth and rupture of intracranial aneurysms. And of the above given reasons hemodynamics or the intra-aneurysmal blood flow and the spatial and time variation of the wall shear stress(WSS) is considered to play a significant role[10, 12, 13, 15, 16]. However the role of hemodynamics is considered a controversial issue since there are two opposing views to this problem[14]. One view supports that low WSS on the aneurysm wall triggers a mechanobiological process such as arterial wall remodeling, which degrades and weakens the aneurysm wall and results in rupture. The opposing theory claims that mechanobiological processes associated with high WSS
are responsible for localized damage to the vessel wall that result in rupture.

\section*{1.3 Diagnosis}

The presence of an aneurysm can be diagnosed with the help of a number of sophisticated imaging techniques out of which the three-dimensional rotational angiography (3DRA) is the preferred to detect and treat cerebral aneurysms. It is an invasive technique where a contrast agent is injected intra-arterially using a catheter to enhance imaging of the blood vessels. The angiographic images are then acquired during the pass of the contrast using low dose x-rays. It is the preferred modality because these images have the highest resolution and highest contrast which also makes it the most suitable option for generating computational models to run flow calculations. Figure 1.2 shows a 3DRA image of the two ICA’s and an aneurysm on the anterior communicating artery. More details on using 3DRA images to generate geometrical models will be presented in the following sections.
1.4 Treatment

Surgical clipping and coil embolization are the most common methods of treatment[17]. Surgical clipping involves placing a metallic clip across the neck of the aneurysm thereby preventing the blood from entering into the aneurysm. Endovascular embolization is a minimally invasive technique that consists in packing the aneurysm with platinum coils. The idea here is to clot the blood inside the aneurysm and completely stop the intra-aneurysmal flow. Both procedures intend to isolate the aneurysm from the arterial circulation thus preventing aneurysm rupture and hence the hemorrhage. Surgical clipping is not always the favored choice due to the high risks involved with craniotomy and the impossibility of clipping certain aneurysms due to their location and shape. Endovascular embolization also has some serious complications. Numerous cases of coil compaction leading to the re-growth or the formation of a secondary aneurysm have been reported. Moreover, many wide neck or fusiform aneurysms cannot be treated with coils alone. In cases like these a stent - an expandable metallic mesh - is deployed in the parent vessel to hold the coils inside the aneurysm dome. Recently, there has been growing interest in the use of stents as flow diverters without coils. With an effective design, the blood flowing into the aneurysm can be disrupted thereby promoting intra-aneurysmal thrombus formation and thus preventing rupture[18–20].

1.5 Modeling

The coil compaction problem has led to an increase in the use of stents to reduce the flow of blood into the aneurysm and promote thrombus formation in the aneurysm sac. Extensive studies are being conducted in order to find the functional relationship between stent properties such as strut diameter, porosity, design etc. and rheological and hemodynamic changes in the aneurysm and the parent vessel. M. Ohta et al.[21] investigated the influence of stent placement on blood flow velocity and wall shear stress of an idealized intracranial aneurysm using a finite element modeling approach. Also to assess viscosity changes...
induced by stent placement, the rheology of blood as a non-Newtonian fluid was taken into account. S.C.M. Yu et al. [22] conducted an in vitro study with stents and springs of different porosity to analyze the flow inside the aneurysmal pouch. The experiments were conducted using Particle Image Velocimetry over a range of Reynolds number from 200 to 1600. A.K. Wakhloo et al.[23] studied the changes in local hemodynamics resulting from stent implantation with various porosities. Pulsatile flow patterns in an experimental flow apparatus were visualized using laser-induced fluorescence of rhodamine dye. K. Barath et al.[24] analyzed the influence of different stent parameters on intra-aneurysmal flow reduction. Two different neck-sized elastic sidewall aneurysm models were connected to a circulatory loop. Twenty different stents were introduced in both models to analyse the effect of the parameters such as porosity, filament diameter and permeability. G.R. Stuhne et al.[25] performed a mesh convergence analysis with a varying node spacing near the stent to come up with a mesh resolution requirement to run flow simulations with stents. M. Hirabayashi et al.[26] proposed a stent positioning factor as a characterizing tool for stent pore design in order to describe the flow reduction effect. They simulated the blood flow using the lattice Boltzmann equation. However, most of these studies have been conducted using idealized in-vitro and numerical models. Therefore, the results from these studies cannot be generalized in a straightforward manner to the patient population or used to plan the endovascular treatment of individual aneurysms. On the other hand, personalized or patient-specific models provide a more realistic representation of the in-vivo hemodynamics[15, 27–30]. However, simulating blood-flow past endovascular devices such as stents in patient-specific models poses a number of challenges[31, 32]. These include: techniques for virtual deployment of stents, meshing the vascular model and the deployed stent, developing appropriate pre- and post-stenting boundary conditions, and handling of relatively large computational meshes. This work addresses some of these technical issues using an unstructured grid embedding approach and discusses different clinical questions that can be investigated with these simulations.
1.6 Image-Based Patient-Specific Endovascular Stent Simulations

In order to successfully conduct patient-specific endovascular stenting simulations the following tools and techniques are required -

- Patient-specific modeling of cerebral aneurysms,
- A meshing technique to mesh the parent vessel with endovascular stents, and
- A stent-parent vessel mapping algorithm.

Figure 1.3 shows the original pipeline already developed at the CFD Lab, GMU, to model cerebral aneurysms from 3DRA images. Various studies have been performed to show the validity of using such models to realistically predict the intra-aneurysmal flow pattern in cerebral aneurysms [10, 33]. The issue with regards to generating the computational mesh inside the patient-specific model and around the stent is that it is a very difficult and error prone process. However, an adaptive unstructured embedded approach solves this problem [31, 32]. As part of the doctoral thesis, analysis, evaluation and the final application of this embedded gridding technique in the context of modeling blood flows around stents was carried out. Also a stent-parent vessel mapping algorithm was developed to deploy the stents into the aneurysm model. The following list gives a break-up of the work done and the respective chapter numbers where the details of the work have been discussed.

- Evaluate the embedded gridding technique with the traditionally used body-fitted approach and experiments (Chapter 2).
- Develop an algorithm to deploy stents in patient-specific models (Chapter 3).
- In-vivo evaluation of the pre and post-stenting simulations (Chapter 3).
- Conduct several studies with regards to endovascular stenting in cerebral aneurysms (Chapter 4 & 5).
The extended pipeline in Figure 1.3 is the new extension which was developed to make patient-specific virtual stenting simulations possible. The original pipeline is explained in the following section and specific stages of the extended pipeline such as the stent deployment and changes to the original flow solver to make embedded simulations possible is explained in their respective chapters.

### 1.7 Original Pipeline

In this section the methodology used to study the hemodynamics of endovascular stenting in patient-specific image-based computational models is presented. The computational modeling pipeline includes image acquisition, methods for image processing, segmentation techniques, stent deployment, grid generation and numerical simulations. Each part is discussed in some detail, paying particular attention to the assumptions of the models and the advantages and disadvantages of the methods.

#### 1.7.1 Image Processing

The steps to construct an anatomical model of the cerebral vasculature from a 3DRA image consists of the segmentation of the region of interest, construction of a surface triangulation and the generation of a three dimensional grid of tetrahedra. This process is semiautomatic
and is carried out using an in-house developed software, ZMD, that combines several image processing and geometry processing tools with a common graphical interface\cite{34, 35}.

To segment the region of interest, first, the images are smoothed by applying the blurring and sharpening operations alternatively to reduce noise. This operation tends to smooth the image without enlarging or shrinking the vasculature\cite{10}. A seeded region growing algorithm is then used to segment the vessels and an initial surface model is obtained by iso-surface extraction. The intensity level for the region growing segmentation and the iso-surface extraction are selected by trial and error. The extracted iso-surface is then used as a geometric deformable model\cite{10, 36, 37}, to allow the nodes of the triangulation to deform under the action of internal smoothing forces and external forces computed from the image intensity gradients. Performing this operation moves the surface nodes to the edge of the vasculature structure in the unprocessed image. The final model is then smoothed and interactively cut and extruded. To obtain a good quality surface triangulation, edge collapsing and diagonal swapping algorithms are applied to delete very small or stretched elements and to minimize the maximum internal angle\cite{34}. Finally if some small-scale imperfections are to be removed a non-shrinking surface-smoothing algorithm is used\cite{38}.

If the above mentioned methodology could not segment small vessels from the medical image then a tubular deformable model technique is used to extract that geometry\cite{35, 39}. In order to do this the axis of each vessel is manually selected on cross-sectional view. A cylindrical surface triangulation is then constructed along the vessel axis. The nodes on this surface are allowed to move in the radial direction using the forces mentioned in the geometric deformable model. The image gradients are computed by convolution with the derivatives of a Gaussian kernel. The final surface model is obtained by merging this tubular model with the rest of the surface\cite{40}. The merging is done using a surface-merging algorithm that first creates a background grid covering the entire computational domain. Then, from each point in the background grid the shortest distance to any reconstructed surface is calculated and assigned positive or negative signs based on the location of the point i.e., if the point lies inside or outside the closest surface. In order to increase the
resolution of the algorithm the background grid is adaptively refined in regions close to the surfaces. Finally, a zero-level iso-surface is then extracted from the signed distance map and if required processed based on the above mentioned techniques.

The last stage in the model development process is the generation of the 3-dimensional grid defining the computational domain. This process is carried out using an in-house developed software, GEN3D, that uses an advancing front technique to generate a new surface triangulation and then fill the interior volume with tetrahedra[35,41]. The generated mesh is then improved using methods like edge collapsing and diagonal swapping by the same code. Background grids or/and sources are used to set the element size in the grid. The former generates the mesh with uniform element size whereas the latter refines the mesh in regions of large curvature.

1.7.2 Hemodynamic Modeling

Blood has been typically considered a Newtonian incompressible fluid and has been shown that blood behaves as a Newtonian fluid in large arteries[42]. Also, hemodynamic analyses of cerebral aneurysm models with realistic anatomies using Newtonian and non-Newtonian approximations have indicated that the main flow characteristics are not significantly affected by the viscosity model. However, non-Newtonian effects can become important in the slow flows encountered in stented aneurysms. This issue deserves further investigation. Based on these assumptions, the governing equations are the unsteady Navier-Stokes equations in three dimensions[43]:

\[ \nabla \cdot \mathbf{v} = 0 \] (1.1)

\[ \frac{\partial \mathbf{v}}{\partial t} + \mathbf{v} \cdot \nabla \mathbf{v} = -\frac{1}{\rho} \nabla p + \nu \nabla^2 \mathbf{v} \] (1.2)

where \( \mathbf{v} \) is velocity, \( p \) pressure, \( \rho \) density and \( \nu \) kinematic viscosity. In the simulations, values of \( \rho = 1.0 \, gr/cm^3 \) and \( \nu = 0.04 \, cm^2/sec \) were used. These equations are discretized
in time using an implicit scheme of the form:

$$\nabla \cdot v^{n+\theta} = 0 \quad (1.3)$$

$$\frac{v^{n+\theta} - v^n}{\theta \Delta t} + v^{n+\theta} \cdot \nabla v^{n+\theta} + \frac{1}{\rho} \nabla p^{n+\theta} = \nabla \nu \cdot \nabla v^{n+\theta} \quad . \quad (1.4)$$

The parameter $\theta$ selects the time integration scheme used in the simulation. If $\theta = 1$ the first order backward Euler scheme is recovered, and if $\theta = 1/2$ the second order Crank-Nicholson is obtained. Moving the first term to the right hand side, this scheme can be interpreted as the steady-state solution of the pseudo-time system:

$$\frac{\partial v^{n+\theta}}{\partial \tau} + v^{n+\theta} \cdot \nabla v^{n+\theta} + \frac{1}{\rho} \nabla p^{n+\theta} = \nabla \nu \cdot \nabla v^{n+\theta} - \frac{v^{n+\theta} - v^n}{\theta \Delta t} \quad . \quad (1.5)$$

This is similar to the original Navier-Stokes equation but with a source term on the right hand side. The solution is then advanced in time by solving a steady-state problem in pseudo-time $\tau$ at each timestep. A linear (tetrahedral) finite element discretization of space is used. The discrete system is obtained via the Galerkin weighted residual method. An edge-based formulation is used for efficiency[44]. The resulting discrete system is of the form:

$$K^{ij} \Delta u^j = \sum_{ijn} C^{ij} (F_i + F_j) \quad , \quad (1.6)$$

where $K^{ij}, C^{ij}, F_i, \Delta u^i$ denote the implicit left-hand side matrix, the explicit right-hand side matrix entries, nodal fluxes and increments in the nodal unknowns. The numerical scheme becomes unstable when a Galerkin finite element approximation is done on the advection terms. Therefore, an edge-based upwind finite element approximation is employed to discretize the equations in space[44,45]. The coupled system is solved iteratively using a
fractional step scheme with Lower-Upper Symmetric Gauss-Seidel (LU-SGS) relaxation for the advection parts and a preconditioned conjugate gradient solver for the pseudo-Laplacian of the pressure. The scheme has been optimized over many years, and has been detailed elsewhere[32].

1.7.3 Boundary Conditions

To make the simulations simple, vessel walls were assumed rigid in this work. Although in reality this is not true and sensitivity analyses conducted using vessel deformations measured with dynamic imaging technique show that flow characteristics such as size and location of the flow impaction zone and intra-aneurysmal flow patterns do not change significantly in comparison to simulations conducted using rigid walls[46]. No-slip boundary conditions, $v = 0$, are prescribed at the vessel walls. Pulsatile physiologic flow boundary conditions are used in the simulations. Volumetric flow rate curves measured with phase-contrast magnetic resonance techniques on a number of normal subjects are used to obtain generic flow waveforms for different arteries and scaled to the patient-specific vessel diameter[35]. The inflow velocity distribution is obtained as a superposition of Womersley profiles for each of the Fourier modes of the prescribed flow rate curve[47,48]. The boundary conditions assume fully developed flows and no secondary flows at the inlet boundaries. Previous studies have shown that if subject-specific models are truncated too close to the aneurysm neck fully developed flow conditions can have a significant effect on the intra-aneurysmal hemodynamics[49]. However, for anatomical models with longer upstream portions of the parent artery these assumptions have less influence because they allow for the development of appropriate secondary flows due to the curvature and tortuosity of the parent vessel. Traction-free boundary conditions are imposed at the outflow boundaries. This assumes that the corresponding distal vascular beds have similar flow impedances and hence the flow divisions among the arterial branches is determined by the geometry of the anatomical model. However, it is known that the flow divisions are actually determined by the impedance of of the
distal arterial tree. More realistic approaches involve imposing impedance boundary conditions from morphometric data[50], modeling the distal vascular bed as a block of porous material[51], or coupling to 1D models of the systemic circulation[52]. But even in these approaches the values of the vascular bed parameters (flow rate, impedances, porosities, etc.) have to be tuned in order to avoid unrealistic pressure drops or jumps in the wall shear stress in small vessels included in the vascular models. A study conducted by altering the flow divisions to the different branches showed that there was no major change in the intra-aneurysmal flow characterization[10].

In order, to conduct a realistic patient-specific simulation of a vascular system not only accurate geometry, but also realistic inflow and outflow boundary conditions, arterial wall motion and rheology must be taken into account. Nevertheless, sensitivity analysis has shown that although changes in the velocity fields can be observed, the fundamental characterization of the intra-aneurysmal flow patterns is not altered when the mean input flow, the flow division, the viscosity model, or mesh resolution are changed. The variable that has a greater impact on the computed flow fields is the geometry of the vascular structures[10].

1.7.4 Post-processing and Visualization

All the visualizations presented in this dissertation were generated using ZFEM, the visualization code developed in the CFD Lab, GMU. The variables of interest were the velocity and wall shear stress (WSS). The velocity was visualized by using a cut plane through the region of interest, vector plots and streamlines. The WSS was visualized as surface shading of the model. All the visualizations were done with the results obtained at peak systole of the cardiac cycle.
1.8 Summary

The most common location of intra-cranial cerebral aneurysms is the Circle of Willis and their rupture leads to SAH which in most cases is fatal. The preferred forms of treatment are surgical clipping and endovascular coil embolization. However, these two forms of treatment have some serious complications. Recently physicians have started showing interest in using endovascular stents in the treatment of aneurysms. These devices treat the aneurysms by initiating thrombus formation inside the aneurysm sac or reducing the complexity of the intra-aneurysmal flow pattern. Modeling this endovascular stenting is crucial to design better stents and optimize the treatment procedure. However, most of the previous work was conducted with animal models or idealized in-vitro or computational models. Such studies have some serious shortcomings, namely, they are impractical for large population studies, not useful for patient-specific studies and do not replicate the anatomy of diseased human arteries. Patient-specific models provide a more fast, reliable and inexpensive way to study the problem. However, conducting patient-specific simulations with endovascular devices such as stents has its own challenges. Such as deploying a stent into the model, generating a good quality mesh inside the domain and around these devices, developing appropriate pre- and post-stenting boundary conditions, and handling of relatively large computational meshes. This doctoral thesis is an attempt to address these issues so that patient-specific endovascular stent simulations can be conducted in a realistic way which can ultimately be used to design better stents as flow diverters and optimize the endovascular stenting procedure used to treat cerebral aneurysms.
Chapter 2: Embedded/Immersed Methods

In this chapter the analysis and evaluation of the embedded/immersed methods with the traditional body-fitted method and experimental data is shown. In the first section the problem associated with using the body-fitted technique to generate grids inside patient-specific models of cerebral aneurysms with endovascular stents is explained. Also in the same section a brief description of the embedded/immersed methods as a solution to the above mentioned problem is illustrated. The next section explains the embedded and the immersed method in detail and also the treatment of endovascular stents with respect to these methods. Finally the embedded/immersed method is evaluated with the body-fitted method and the experimental data by comparing results from flow past a circular cylinder and an idealized stented aneurysm.

2.1 Introduction

The main difficulty in performing patient-specific simulations with stents lies in the construction of acceptable computational grids inside the blood vessel and around these devices. Two types of grids are most commonly used for computational fluid dynamics (CFD) simulations: body-conforming and embedded. In the case of body-conforming grids the external mesh faces match up with the surface (vessel walls, surfaces of the stent) of the domain. Whereas in the embedded approach (also known as fictitious domain, immersed boundary or Cartesian method), the surface is placed inside a large mesh (typically a regular parallelepiped) with special treatment of the elements close to the surfaces. The difficulty with using body-conforming grids is that the description of the surface of the computational domain must be given by a watertight assembly of analytical or discrete patches. For complex geometries and complex endovascular devices such as stents or coils, it can be a tedious
Figure 2.1: Modeling of vessels using the body-fitted approach and stents using the embedded approach: the body-fitted grid and the embedded stent (left), mesh refinement around the device (center) and the first order approximation of the stent (right).

and error prone process to construct such a surface description\cite{25, 53, 54}. So an alternative is to use grids that are not body conforming, and simply embed the surfaces of the medical devices in them. The basic idea of this approach is schematically illustrated in Figure 2.1. Briefly, a model of the vasculature is constructed (e.g., from anatomical images) and meshed; Figure 2.1, left. The edges of the elements of this grid that are cut by the surface of the stent are then removed from the flow calculation and appropriate boundary conditions are applied; Figure 2.1, center. In order to increase the mesh resolution and accuracy, the grid is adaptively refined in the vicinity of the surface of the endovascular device; Figure 2.1, right.

In this chapter this adaptive unstructured embedded/immersed grid approach for modeling blood flow past stents is analyzed \cite{31, 32}. The technique uses a hybrid approach wherein the vessel walls are treated with body-fitted unstructured grids and the stents are embedded in this grid. Adaptive mesh refinement is used to increase the mesh resolution around the device. This technique is compared with the body-fitted approach and with experimental data reported in the literature.

### 2.2 Methods

In this section the embedded and the immersed methods are explained in detail and also the treatment of stents, which is represented as a collection of spheres, with respect to these
methods.

2.2.1 The Embedded Unstructured Grid Method

In the embedded mesh technique the surface of the object embedded inside the domain is denoted by computational structural dynamics (CSD) faces. This information is either obtained from a CAD package as a surface triangulation or generated from remote sensing or medical image data. Once the surface triangulation is embedded inside the domain, appropriate kinematic boundary conditions are imposed to the fluid nodes close to the embedded surface. Depending on the required order of accuracy and simplicity, a first-order or second-order (higher-order) scheme may be chosen to apply the boundary conditions.

A first-order scheme can be achieved by -

- elimination of the edges crossing the embedded surface,
- forming boundary coefficients to achieve flux balance,
- applying boundary conditions for the endpoints of the crossed edges based on the normals of the embedded surface.

A second-order scheme can be achieved by -

- duplicating the edges crossing the embedded surface,
- duplicating the endpoints of crossed edges,
- applying boundary conditions for the endpoints of the crossed edges based on the normals of the embedded surface.

Figure 2.2 illustrates the difference between the two approaches.

In this section only the first-order scheme is discussed. This scheme is implemented by first identifying and eliminating the edges crossing the embedded surface. Next the boundary coefficients are formed to achieve flux balance. Finally, boundary conditions are applied to the endpoints of the crossed edges based on the normals of the embedded surface. Each of these steps are further explained below.
Determination of crossed edges

The first step in the determination of crossed edges is to find the CFD edges cut by the CSD faces. This is done by using a fast spatial search data structure, such as an octree or a bin. An octree is a type of data structure that emulates a tree structure with a set of linked nodes and in which each internal node has up to eight children. They are most often used to partition a three-dimensional space by recursively subdividing it into eight octants. Using bins is another way of reducing search overheads for spatial proximity. The domain in which the data (e.g. points, edges, faces, elements, etc.) falls is subdivided into a regular mesh or bricks. Then the required data can be obtained from these bricks or bins.

Assuming that a bin is used for the CSD faces, a (parallel) loop is performed over the CFD edges and a bounding box is built for each edge. All the faces in the region of the bounding box are found from the bin. This is followed by an in-depth test to determine which faces cross the given edge. From this the face crossing closest to each of the edge end-nodes is stored. Once the faces crossing the edges are found, the face closest to the endpoints of crossed edges is also stored. This information is required to apply boundary conditions for the points close to the embedded surface. Figure 2.3 illustrates these steps. In the cases where the embedded surface cuts only a small portion of the CFD edges, a considerable speedup may be realized by removing from the list of edges tested all those that fall outside the global bounding box of the CSD faces. The resulting list of edges to be
tested in depth may be further reduced by removing all the edges whose bounding boxes do
not fall into an octree or bin filled with faces covering that spatial region. By using these
two filters, the list of edges to be tested in detail can be reduced by an order of magnitude.

First-order treatment

Given a CSD triangulation and the CFD mesh, the edges cut by the CSD faces are found and
deactivated. Considering an arbitrary field point \(i\), the time advancement of the unknowns
\(u^i\) for an explicit edge-based time integration scheme is given by

\[
M^i \Delta u^i = \Delta t \sum_{ij} C^{ij}(F_i + F_j),
\]

where \(C\), \(F\), \(M\) denote the edge coefficients for an edge \(ij\), fluxes and mass matrix respectively and \(\Omega\) denotes the domain. If an edge is crossed by a CSD face then the coefficients
\(C^{ij}\) are set to zero which means that for a uniform state \(u=\)constant the balance of fluxes
for interior points with cut edges will not vanish. This is corrected by defining a new bound-
ary point to impose total/normal velocities, as well as adding a 'boundary contribution',
resulting in
\[ M^i \Delta u^i = \Delta t \left[ \sum_{i \in \Omega} C^{ij} (F_i + F_j) + C_i^j F_j \right] . \] (2.2)

\( \Gamma \) denotes the location of the embedded surface

From the condition that \( \Delta u = 0 \) for \( u = \text{constant} \) the point coefficients \( C^i \) are obtained. Given that gradients \( g \) (e.g. for limiting) are also constructed using a loop of the form given by equation (1) as

\[ M^i g^i = \sum_{i \in \Omega} C^{ij} (u_i + u_j) , \] (2.3)

it would be desirable to build \( C^i \) in such a way that the constant gradient of a linear function \( u \) can be obtained exactly. But since the number of coefficients is small this seems impossible. Therefore the gradients at the boundary are either set to zero or extrapolated from the interior of the domain.

The mass matrix \( M^i \) of points surrounded by cut edges must be modified to reflect the reduced volume due to cut elements. The simplest possible modification of \( M^i \) is given by the so called 'cut edge fraction' method [31]. In a pass over the edges, the smallest 'cut edge fraction' \( \xi \) for all the edges surrounding a point is found as shown in figure 2.4. The modified matrix is then given by

\[ M^i_s = \frac{1 + \xi_{\text{min}}}{2} M^i . \] (2.4)

Note that the value of the modified mass-matrix can never fall below half its original value, implying that the timestep sizes will always be acceptable.
Boundary conditions

The points which belong to the cut edges that are eliminated are now part of a boundary. Hence the usual Navier-Stokes boundary conditions are applied i.e. the velocity is set to zero if the embedded surface is at rest. The setting up of this boundary condition may be improved by extrapolating the velocity from the surface with field information. The location where the flow velocity is equal to the surface velocity is the surface itself, and not the closest boundary point. As shown in figure 2.5 for each boundary point the closest point on the CSD face is found. Followed by which three neighboring non-boundary points are found and a tetrahedral element that contains the boundary point is formed. The velocity imposed at the field point is then found by interpolation. In this way, the boundary velocity 'lags' the field velocity by one timestep.
2.2.2 Immersed Unstructured Grid Method

In this method the embedded object is assumed to be given by a tetrahedral mesh rather than the CSD faces. The motivation for this technique comes from the fact that when fluid flows over a body it exerts some pressure force on the surface and if the surface is no-slip then the fluid also exert some shear force [55]. Conversely, the surface also exerts some force of the opposite sign in the fluid and if its a no-slip case this force is what brings the fluid to rest on the surface. Hence, without the surface if the correct set of forces are applied to the fluid then the fluid would flow as though it were passing over a solid object. In this method this is achieved by adding suitable force functions to the flowfield. Considering a rigid body as shown in Figure 2.6, setting the fluid velocity same as the body velocity within the fluid can be accomplished by applying a force term of the form

\[ f = -c_0 (v_b - v) , \]  

for the points inside the body. This is known as the penalty force technique and the approach used here follows that of Mohd-Yusof [56]. The right-hand side for the flow equations at immersed points is evaluated first. Then a force is added such that the velocity at the next timestep satisfies the kinematic boundary conditions. Below given is the spatially discretized form of the momentum equations at each point \( i \)

\[ M \frac{\Delta v_i}{\Delta t} = r_i + f_i , \]  

\( f_i \) is obtained as

\[ f_i = M \frac{w_i^{n+1} - v_i^n}{\Delta t} - r_i , \]

where \( w_i \) denotes the velocity of the immersed body at location of point \( i \) and \( n \) is the timestep. For explicit time-stepping schemes, this force function in effect imposes the
velocity of the immersed body at the new timestep. Therefore this is denoted as a 'kinetic-kinematic' approach.

### 2.2.3 Treatment of spheres

Two basic options can be used to represent the geometry of the stents: (a) generate a surface triangulation (explicit surface definition), or (b) generate a series of overlapping spheres to define the stent geometry (implicit surface definition). In the latter approach, the distance between the spheres can be adjusted to represent the stent to any degree of accuracy. The two approaches are illustrated in Figure 2.7. Since the geometry of the stent is only used to determine the elements of the computational grids that are cut by the stent surface, these two approaches differ only in the way that the intersection between the stent wire surfaces and the computational grid is computed. In the former approach this requires computing the intersections between triangles of the stent surface and edges of the computational tetrahedral mesh. In contrast, the latter approach requires intersections between these edges and the spheres representing the stent, which is simpler to implement.

The information required to represent the stent is the center for each sphere and its radius. The adaptive embedded or the immersed technique can be linked to these spheres in a natural way. As shown in Figure 2.8 the points of the element into which the center of
the sphere falls are marked as 1’s and additional layer of points as 2’s. All edges touching any of the marked points are subsequently marked as crossed. Once the crossed edges are known the embedded/immersed mesh technique can be used. In the case of the embedded the velocity of the sphere is imposed at the endpoints of the crossed edges. And in the case of the immersed, the velocity of the particle is imposed at the interior points.

Figure 2.8: Treatment of embedded spheres.


2.3 Results

2.3.1 Flow past a cylinder

The methodology was first tested on a flow around a circular cylinder under steady flow conditions. This computational experiment was performed to compare the results obtained with the embedded and body-fitted methods and experimental data from the literature. The cylinder was 0.01 cm in diameter, which is typical of the diameter of stent wire. This cylinder was placed in a box of dimensions 1.0 cm x 0.5 cm x 0.01 cm. The origin is at the center of the cylinder, with the x-axis from left to right and the y-axis in the vertical direction. A body-conforming mesh was generated with a resolution of 0.02 cm in the far field and 0.00075 cm near the surface of the cylinder. The mesh consisted approximately 365K elements (see Figure 2.9). Steady-state simulations were carried out using a Reynolds number of \( Re = 25 \). This is a typical value for the local Reynolds number of a stent in a cerebral artery. This can be calculated with the assumptions that for blood \( \rho = 1.0 \text{g/cm}^3 \), \( \mu = 0.04P \), and in cerebral arteries blood velocity \( v = 100 \text{cm/s} \) and the stent wire has a diameter of \( d = 0.01 \text{cm} \). A uniform velocity profile was prescribed at the inflow, traction-free boundary conditions at the outlet and no-penetration boundary conditions \( (\dot{r}n = 0) \) were used in the walls of the box and no slip boundary condition \( (v = 0) \) were applied at the surface of the cylinder. The flow direction in Figure 2.10 is from left to right. For the embedded approach, an initial uniform mesh without the cylinder was created with a resolution of 0.01 cm. This grid contained approximately 36K elements. Then a cylindrical surface was embedded in this grid and adaptively refined around it. The mesh was refined with four, five, six and seven levels of refinement obtaining approximately 92K, 242K, 822K and 3M elements, respectively. Computational fluid dynamics (CFD) simulations were performed for these grids using the same conditions prescribed in body-fitted run. Figure 2.9 shows the details of the finite element grids. Since stents are represented as a sequence of overlapping spheres, similar set of runs were performed by approximating the cylinder with 50 overlapping spheres and the results are shown in Table 2.1. Figure 2.9 shows
Table 2.1: Drag coefficient for $Re = 25$

<table>
<thead>
<tr>
<th></th>
<th>Experimental</th>
<th>Body-fitted</th>
<th>Level 4</th>
<th>Level 5</th>
<th>Level 6</th>
<th>Level 7</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>1.8597</td>
<td>1.8488</td>
<td>-0.6%</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Body-fitted</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Embedded surface triangulation</td>
<td>1.6851</td>
<td>1.7439</td>
<td>1.809</td>
<td>1.847</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(-9.4%)</td>
<td>(-6.2%)</td>
<td>(-2.7%)</td>
<td>(-0.7%)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Embedded overlapping spheres</td>
<td>1.607</td>
<td>1.7247</td>
<td>1.8147</td>
<td>1.8773</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(-13.6%)</td>
<td>(-7.3%)</td>
<td>(-2.4%)</td>
<td>(1.0%)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

The contours of velocity magnitude obtained with the body-fitted grid (top) and with the embedding approach after four (center) and six (bottom) levels of refinement. The results obtained for the embedded technique show good agreement with the body-fitted results.

Table 2.1 shows the value of the drag coefficient and its percentage error in brackets for the body-fitted and embedded simulations, in comparison with the experimental value [57]. The results show that the error decreases with the increase in levels of refinement. These results show that for the low Reynolds numbers usually encountered in blood flows with stents, the embedded technique produces good results. This example also shows that as the level of refinement is increased, the solution becomes more accurate and the drag coefficient converges to the experimental value. Since the mesh refinement is entirely automatic, no extra modeling work is needed to achieve higher levels of accuracy. This is what makes the embedded approach very attractive especially for devices with complex geometries which can have region of self contact or have small gaps with the vascular walls.
Figure 2.9: Grids used for the flow simulation. From top to bottom: body-fitted grid, uniform mesh with no refinement, mesh with 2 levels of refinement and mesh after 4 levels of refinement.
Figure 2.10: Velocity contours for the body fitted grid (top), the embedded grid with four levels of refinement (center) and the embedded grid with six levels of refinement (bottom).
2.3.2 Idealized stented aneurysm

In this example, an idealized aneurysm model was constructed by merging the triangulations of a cylinder and a sphere. The parent vessel was modeled as a straight cylinder of 0.35 cm diameter. The spherical aneurysm was modeled as a sphere of radius 0.466 cm displaced 0.4 cm from the axis of the parent vessel. A stent was modeled with 12 intersecting helices of 0.01 cm in diameter and 0.5 cm long. The helices were distributed along the circumference of the parent vessel with alternating directions of rotation, and one turn from one end to the other. The body-fitted and the embedded techniques were used to grid this idealized stented aneurysm. The body-fitted grid was constructed as follows. First, a vascular model was created by merging the surface triangulations of the parent vessel (cylinder) and the aneurysm (sphere) producing a smooth aneurysm neck. Second, a geometrical model of the stent was created by generating independent triangulations along each helical wire and then fusing them with the same surface-merging algorithm. Then, the final geometrical model was created by fusing (subtracting) the triangulations of the vascular model and the stent. This geometrical model was used to generate a finite element grid, that contained approximately 10M elements. For the embedded approach the vascular model (without the stent) was meshed, resulting in an initial grid of 763K elements. This grid was then refined one, two and three levels, resulting in a total of 1.2M, 2.85M and 8.5M elements, respectively. Figure 2.11 shows the geometry of the model and the grids.

The flow patterns before and after stenting were computed under steady flow conditions. The stented aneurysm was modeled with the body-fitted and embedded approaches. A parabolic velocity profile corresponding to a total flow rate of $Q = 4.78 \text{ ml/s}$ was prescribed at the model entrance and traction-free boundary conditions at the outlet. Figure 2.10 shows the velocity contours (left column) and streamlines (right column) before stenting (top row) and after stenting using the body-fitted (center row) and embedded (bottom row) techniques. It can be seen that the results obtained with bodyfitted and embedded grid approaches are similar. These visualizations also show the alterations in the aneurysmal flow pattern caused by the stent. Before stenting the inflow zone is located at the distal part of
the neck and the flow rotates in the clockwise direction inside the aneurysm. After stenting, the inflow zone is located at the proximal end of the neck and the intra-aneurysmal flow rotates in the counter-clockwise direction. This observation is consistent with observations made by Lieber et al. using in vitro models and particle image velocimetry techniques [58].

In order to further compare the solutions obtained with the body-fitted and embedded grids the velocity magnitude was plotted along three lines placed inside the aneurysm parallel to the neck and at different depths. Figure 2.12 shows the plots of the velocity magnitude along these lines. The location of these lines inside the aneurysms is also shown in Figure 2.12. These results also show a good agreement between the two approaches and that the embedded grids yield results that converge to the body-fitted result as the levels of refinement are increased. The reduction in the maximum velocity along each of these lines obtained with the different methods are listed in Table 2.2. The left column of this table lists the maximum values of the velocity along each line. The other columns list the factor by which this maximum velocity is reduced, as computed by the different methods. The percent values in parentheses are the relative errors between the velocity reduction factors with respect to the one computed with the body-fitted grid. This table indicates that with only two levels of refinement, the velocity reduction after stenting was predicted within 5%
Figure 2.12: Velocity contours (left column) and streamlines (right column), before stenting (top row), body-fitted (middle row) and embedded (bottom row). The flow is from right to left.
of the prediction made by the body-fitted method.

Table 2.2: Comparison of velocity reduction between body-fitted and embedded.

<table>
<thead>
<tr>
<th>Line</th>
<th>Pre-stenting</th>
<th>Body-fitted</th>
<th>Embedded level 1</th>
<th>Embedded level 2</th>
<th>Embedded level 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>3.26 cm/s</td>
<td>0.95</td>
<td>0.85 (11%)</td>
<td>0.9 (5%)</td>
<td>0.92 (3%)</td>
</tr>
<tr>
<td>2</td>
<td>6.54 cm/s</td>
<td>0.9</td>
<td>0.83 (8%)</td>
<td>0.86 (4%)</td>
<td>0.88</td>
</tr>
<tr>
<td>3</td>
<td>9.1 cm/s</td>
<td>0.72</td>
<td>0.65 (10%)</td>
<td>0.69 (4%)</td>
<td>0.71 (1%)</td>
</tr>
</tbody>
</table>

2.4 Summary

A methodology to simulate blood flows in patient-specific models with stents that combines body-fitted and embedded unstructured grids has been developed and evaluated. This
approach to a very large extent reduces the hardship associated with just using the body-conforming technique to mesh the parent vessel and around the endovascular devices. This is so because the mesh refinement and embedding steps are done in a fully automated way in a few minutes, while the generation of body-conforming grids can be a tedious manual process that can take from hours to weeks, depending on the geometric complexity of the problem. The methodology was evaluated with the flow past a circular cylinder and an idealized stented aneurysm. The results obtained were in close agreement with body-fitted computations and experimental results. These results also show that quantities such as the reduction in intra-aneurysmal flow velocity after stenting can be accurately calculated with the embedding approach with a relatively small number of mesh refinement levels.
Chapter 3: Patient-Specific Stenting

The previous chapter dealt with the embedded/immersed method and its comparison with the traditional body-fitted method of running flow simulations. In this chapter the practical application and usefulness of this embedded/immersed method is demonstrated by showing how it could be used in hemodynamic simulations of patient-specific aneurysms with stents. The first section describes the algorithm used to deploy stents in models, then the simulations are compared to in-vivo observations to prove the validity of the approach. Finally, a sensitivity study done to predict the change in intra-aneurysmal flow pattern when a particular stent is deployed in two different configurations is presented.

3.1 Stent deployment

Irrespective of the method chosen to represent the stent geometry, a method to deploy a given stent into a patient-specific vascular model is necessary and this is done in four steps [59].

1. centerline extraction of the parent vessel,

2. initial cylindrical host surface generation,

3. vessel wall’s adaptation to the host surface, and

4. mapping of stent design.

A detailed description of each stage is given below and illustrated in Figure 3.1.

In order to extract the skeleton of the parent vessel a minimal cost path construction algorithm is used [60]. The idea here is to first compute the distance to the wall map within the anatomical model. Then two manually selected endpoints of the skeleton are connected
with the minimum arclength path that travels along the local maxima of the distance map. The final output is a collection of single connected points that represents the centerline of the parent vessel.

Once the centerline is computed the next step is the generation of a cylindrical surface along this line. This is done by first computing an arclength parametrized cubic spline interpolation of the centerline points. Two arclength parameters $s_0$ and $s_1$ are selected in such a way that the interpolated centerline between them satisfies the criteria that it has the arclength of the targeted stent and that it is in the required deployment position. The selection of these parameters is done manually by trial and error. Then a triangulated cylindrical surface is generated along the interpolated centerline between these values. The radius of the cylinder is set to the distance to the wall at each step. The outcome from this step is a triangulated cylindrical surface that lies completely inside the anatomical model matching roughly the vessel geometry and is already in the deployment position.

The third step is aimed at improving the fit of the cylinder with the parent vessel. This is done using the external forces and internal smoothing forces applied to the triangulated surface. The external force is the inflating force or the radial force which is computed as the distance vector between the points in the triangulated surface and the centerline. This force has the effect of inflating the cylindrical surface while maintaining the cylindrical shape. The internal smoothing forces are based on the classical smoothing laplacian operator. The effect of this force is to keep the cylindrical surface smooth while it is being deformed. It is basically an attractive force between each point in the triangulated surface and the points connected to it.

The internal and external forces are applied by updating the vertices of the cylinder by using the classical Newton’s law of motion:

$$m_i \frac{\partial^2 \mathbf{P}_i}{\partial t^2} + \gamma \frac{\partial \mathbf{P}_i}{\partial t} - \tilde{\alpha}_i f_{\text{int}}(\mathbf{P}_i) = \tilde{\beta}_i f_{\text{ext}}(\mathbf{P}_i)$$  \hspace{1cm} (3.1)
where $m_i$ is the $i$th vertex mass and $\gamma$ is a damping parameter. The parameters $\tilde{\alpha}_i$ and $\tilde{\beta}_i$ controls the influence of the internal and external forces respectively. The discretization of Equation 3.1 using a fully explicit discretization scheme leads to

$$
p_i^{n+1} = p_i^n + (1 - \gamma)(p_i^n - p_i^{n-1}) + \alpha_i f_{int}(p_i^n) + \beta_i f_{ext}(p_i^n)$$

(3.2)

where $\alpha_i$ and $\beta_i$ are force weights including the point mass and the timestep. Stability of this scheme depends on $\alpha_i$, $\beta_i$ and $\gamma$ lying inside $[0, 1/2], [0, 1]$ and $[0, 1]$ respectively. All the cases in this text were performed with the parameters $\gamma = 1$, $\alpha_i = 0.01$ and $\beta_i = 0.0005$. These values were selected on a trial and error basis using several test cases.

Boundary conditions are applied for all the points lying on the top and bottom ends of the cylinder. For these points only the components of the total force that are in the planes defined by the top and bottom ends of the cylinder are kept. This is done to maintain the original stent length and deployment position along the vessel.

During the deformation process when most of the points of the cylindrical surface are on the vessel walls the process is stopped. This is done interactively in order to prevent the cylindrical surface from entering into the aneurysm. If a point crosses the vessel wall it is projected back onto the wall and kept fixed in that position for the rest of the simulation.

The final step consist of mapping the different stent designs on the cylindrical surface. The stent designs are drawn as a collection of connected lines with appropriate thickness. The number and distribution of the triangular elements on the rectangle matches the host surface. Using a simple coordinate transformation, the stent designs are mapped onto the deformed cylinder in order to obtain the final stent models in the deployed state. If the deployed stent needs to be rotated inside the vessel a constant phase angle is used.

Once the stent is deployed using the embedded/immersed method the edges of the computational grid cut by the surface of the stent, which is either represented as a triangulation or as a set of overlapping spheres, are identified. At the intersection points a no-slip boundary condition is automatically introduced and the cut edges are excluded from the flow.
Figure 3.1: Steps followed in the vascular deployment of stents: the 3D rotational angiography image (top row, left column), reconstructed anatomical model (top row, right column), construction of the skeleton of the vascular model (second row, left column), initial cylindrical surface (second row, right column), cylindrical surface in the vessel skeleton (third row, left column), final cylindrical surface (third row, right column), two stent designs (fourth row) and the stents after deployment (fifth row).
calculation. In addition, regions of connected edges that lie outside the computational domain, e.g. completely inside the stent wires, are identified and deactivated. An attractive feature of this embedded approach is that it only requires minimal modification of existing incompressible flow solvers.

### 3.2 In-vivo evaluation

In order to demonstrate the validity of the flow simulations using stents a qualitative comparison was done by generating virtual angiograms and comparing them with conventional angiograms acquired on a patient that was treated with a braided stent only (no coils). The conventional angiograms were acquired before and immediately after stenting and during a follow up examination a week later. These angiography images were acquired on a Philips flat panel system (Philips Medical Systems, Best, The Netherlands) and reconstructed into a 3D dataset of 256x256x256 isotropic voxels of 0.2847 mm.

After computing the flow field the virtual angiograms were constructed by numerically solving the transport equation:

$$\frac{\partial \phi}{\partial t} + \vec{v} \cdot \nabla \phi = 0,$$

where $\phi$ represents the dye concentration field, and $\vec{v}$ is the velocity field computed by solving the Navier-Stokes equations. This equation was numerically solved using a finite element scheme [61]. These angiograms are visualizations of the passage of a simulated bolus of contrast transported by the blood flow. A time-dependent concentration with the shape of a Poisson distribution was imposed at the model inlet simulating a four seconds injection of contrast. The concentration field was then advanced for 10 cardiac cycles assuming periodicity of the flow velocity field. Virtual angiograms were then created by volume rendering the mesh points with opacity and intensity modulated by the concentration field at each instant. Cine loop animations were generated with both the angiograms to observe qualitative differences.
Figure 3.2: Flow pattern (left column) and wall shear stress distribution (right column) at peak systole before (top row) and after (bottom row) stenting.
Figure 3.3 shows the frames from conventional and virtual angiograms. The left panel shows four frames from the conventional (left) and virtual (right) angiograms before aneurysm stenting. The first three frames (from top to bottom) correspond to the filling phase, while the last frame corresponds to the washout phase. From these images it can be clearly seen that the main flow characteristics observed in the conventional angiogram are reasonably reproduced by the computational model. The location and size of the inflow jet and flow impaction zone, and the major vortical structures observed inside the aneurysm for the virtual angiogram were similar to the conventional angiogram. The center panel of Figure 3.3 shows four selected frames of the conventional (left) and virtual angiograms right after stenting. Again the flow pattern observed in the virtual angiograms is similar to the conventional angiogram. Similar to the results shown in Figure 3.2 these visualizations confirm the predicted alteration of the flow pattern from a complex to a simple flow type and slower flow velocities (i.e. increased residence times). The angiograms also show that little contrast reaches the distal part of the dome, while most of the dye concentrates near the inflow region. The rightmost column in Figure 3.3 shows selected frames from a conventional angiogram obtained during a follow up exam a week after treatment. From the images one can observe that the blood is still entering the aneurysm at a lower speed and almost no dye reaches the distal part of the aneurysm dome probably because this part of the aneurysm might be thrombosed. A significant pooling of the contrast agent and increased residence times are also observed.
Figure 3.3: Conventional (gray background) and virtual (white background) prior to stenting (left panel), right after stenting (center panel) and at a one week follow up exam (right panel).
### 3.3 Effects of stent positioning

When a stent is deployed to treat an aneurysm, the interventional neurologist has little control on the exact positioning of the stent. However, depending on the design of the stent, positioning can potentially have a significant effect on the hemodynamics of the stented aneurysm. Therefore, it is important to study the sensitivity of the flow diversion characteristics of different stent designs to their positioning. This section presents results of different positioning of the Neuroform stent in four patient-specific geometries. One additional simulation was performed for each model by rotating the stent half a cell, i.e. the maximum angular displacement with respect to the initial positioning. The flow patterns and wall shear stress distributions at peak systole are shown in Figure 3.4. These results show that the flow modifications obtained with the two positioning of the stent are in close agreement. Only for patient 1 (top row of Figure 3.4) there was a slight difference in the deviation of the inflow jet and the WSS near the inflow zone of the aneurysm neck. Additionally, in section 4.2 of the next chapter, a substantial difference in the flow patterns obtained for the ACoA aneurysm after repositioning the stent across the aneurysm neck will be observed. Hence the difference is expected to be larger in stents with large cells because they can miss the inflow jet, or even squeeze the inflow jet producing higher velocities. In contrast, smaller closed cells are more likely to disrupt the inflow jet in a more consistent manner irrespectively of the stent positioning.
Figure 3.4: Streamlines and WSS results for the original stent configuration (first and third row) and the rotated configuration (second and fourth row).
3.4 Partial stent modeling

When comparing the relative performance of different stents as flow diverters one is often concerned with the alterations of the flow pattern and WSS in the aneurysm and not so much in the detailed flow fields near the walls of the parent vessel. In such cases, the computational expense of the simulations can be significantly reduced if only the portion of the stent that crosses the aneurysm neck is modeled. This would result in much smaller computational meshes. Therefore, in order to assess whether this approach yields reasonable results, comparisons of the flow fields obtained by modeling the entire stent and only the portion that covers the neck were carried out. For this purpose, the same four models of section 3.1 and the Neuroform stent were used. The original stent models were interactively cut and the portions that lied entirely on the vessel walls were removed. The original and truncated or partial stent models in their deployed state for each patient are shown in Figure 3.5. The results of the flow simulations with both stent models for each patient are presented in Figure 3.6. These results show that the flow patterns and wall shear stress distributions obtained with the truncated stent models closely match those obtained with the full stent models. Additionally, Table 3.1 lists the mesh size and the computation times taken for the full and truncated stent simulations.

Table 3.1: Mesh size and computation time taken to simulate flow with the full and partial stents.

<table>
<thead>
<tr>
<th></th>
<th>Full</th>
<th>Partial</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No. of elements</td>
<td>Time taken</td>
</tr>
<tr>
<td></td>
<td>(in millions)</td>
<td>(in hrs)</td>
</tr>
<tr>
<td>Model 1</td>
<td>4.3</td>
<td>20:23</td>
</tr>
<tr>
<td>Model 2</td>
<td>7.1</td>
<td>16:30</td>
</tr>
<tr>
<td>Model 3</td>
<td>5.3</td>
<td>11:43</td>
</tr>
<tr>
<td>Model 4</td>
<td>3.9</td>
<td>10</td>
</tr>
</tbody>
</table>
Figure 3.5: The full stent and the partial stent for the different aneurysms.
Figure 3.6: Streamlines and WSS results for the different aneurysms with full and partial stents.
3.5 Summary

A stent deployment algorithm was presented which can be used to perform virtual stenting simulations with patient-specific anatomical models with a given stent design. This method represents a first attempt at modeling the stent geometry after deployment, and it can be used as the initial configuration for more complex models accounting for the material properties of the stent and the vessel wall. It was shown that hemodynamic simulations performed using patient-specific computational models constructed from medical images are capable of realistically representing the in vivo intra-aneurysmal hemodynamic characteristics observed during conventional angiography examinations before and after stenting. This confirms that such pre and post-stenting patient-specific simulations can be used to better understand the effects of different stent designs and to predict the alteration in the intra-aneurysmal hemodynamic patterns of an aneurysm produced by a given stent. This is important for designing better flow diverting devices such as stents and improving patient treatment plans. A sensitivity study was done to study the effects of stent positioning on flow patterns inside the aneurysm. Although the positioning can potentially have a significant effect in many situations these effects are actually quite small. The performance of the stent is related to the number of wires that cross the inflow jet blocking the flow into the aneurysm. This depends not only on the stent design but also on the fluid dynamics characteristics of the aneurysm. It is to expect that the flow diverting characteristics of highly porous stents with asymmetric cell designs will be more dependent on the positioning, especially for aneurysms with thin or concentrated inflow jets. Stents with smaller and symmetrical cells may have more chances of disrupting the inflow jet independently of their positioning.
Chapter 4: Flow Alteration Analysis

In the previous chapters the methodologies required to deploy a stent and run patient-specific flow simulations were described and demonstrated. But in order to evaluate the effectiveness of a stent qualitative or quantitative assessment of the intra-aneurysmal flow is necessary. The first part of the chapter deals with the effect of stent designs on change in intra-aneurysmal flow patterns. Three stent designs were used with four patient-specific models to conduct hemodynamic simulations. The second part of the chapter is concerning the different options physicians have to treat certain aneurysms with regards to stent placement. And in the final part of this chapter an attempt to quantify the change in intra-aneurysmal flow before and after stenting by calculating various hemodynamic quantities is explained.

4.1 Effects of Stent Design

In this section the methodology is demonstrated by simulating the effects of three different stent designs on four patient-specific models of internal carotid artery (ICA) aneurysms (a total of 12 simulations). This illustrates how the methodology can be used to select the best available flow diverting device for a given aneurysm. The vascular models were created from 3DRA images as described in the methods section. Figure 4.1 shows the 3DRA images and the corresponding vascular models with the various stent designs. The aneurysms were located in the left ICA for patients 1 and 4, and in the right ICA for patients 2 and 3. The first stent design (third row) made up of rhomboidal cells corresponds to the Neuroform stent from Boston Scientific, Inc., and the other two stents are helical stents with the stent wire pointing towards the left (fourth row) and right directions (fifth row). Pulsatile flow calculations were performed without and with the stents inside the vascular models. Figures
4.2 and 4.3 show the streamlines and wall shear stress (WSS) distributions at peak systole, before and after stenting, respectively.

In the patient 1, the inflow jet is located towards the left and distal part of the neck, impacting on the distal part of the body of the aneurysm. All the stents are seen to diffuse the inflow jet and re-direct it towards the dome of the aneurysm, and reduce the WSS in the aneurysm. The stent that diffuses the inflow jet and reduces the WSS the most for this patient seems to be the left helical stent.

In patient 2, the inflow is located at the distal part of the neck and the intraaneurysmal flow circulates in counter-clockwise direction. The Neuroform stent produces very little modification to the flow pattern but an increase in the WSS at the distal part of the neck can be observed. This is because blood is being squeezed between the stent wires and the vessel walls. In contrast, the two helical stents produce more significant flow modifications. In particular, they shift the inflow to the proximal part of the neck and change the direction of the intra-aneurysmal flow circulation to clockwise. As in the previous case, the left helical stent seems to diffuse the inflow jet and reduce the WSS the most.

In patient 3, the inflow is located in the distal part of the neck and the flow recirculates in the clockwise direction. All stents diffuse and deviate the inflow jet towards the left part of the neck. However, they do not change the direction of flow circulation in this aneurysm. As in the previous case, a more pronounced increase in the WSS at the distal neck can be observed for the Neuroform stent. The helical stents seem to yield smoother flow patterns, i.e. less vortical structures, than the Neuroform. Again, the stent that seems to cause a larger diffusion of the inflow jet and reduction in the WSS is the left helical stent.

In patient 4, the inflow is located at the proximal part of the neck and flow mainly circulates in the counter-clockwise direction. In this case, the Neuroform stent seems to concentrate the inflow jet, while the helical stents produce a small diffusion of the inflow jet and a smoother intra-aneurysmal flow pattern. All stents produced an increase of the WSS at the distal neck, without changing the circulation direction. The stent that seems to cause a larger diffusion of the inflow jet and larger reduction of the WSS is the right helical
Figure 4.1: Original 3DRA images, computational models and the three stents deployed into the vascular models.
Finally, the same stent may perform differently on different aneurysms. The selection of the best available flow diverter depends also on the hemodynamic pattern of the aneurysm and is therefore patient specific.

Figure 4.2: Streamlines for the four aneurysm models, pre-stented case (top row), with Neuroform stent (second row), left helical stent (third row) and right helical stent (bottom row).
Figure 4.3: Wall shear stress for the four aneurysm models, pre-stented case (top row), with Neuroform stent (second row), left helical stent (third row) and right helical stent (bottom row).
4.2 Effects of Treatment Options

This section illustrates how the methodology can be used to select the best therapeutic option for a given patient. Three patients with intracranial aneurysms are considered. One patient had two aneurysms in the left middle cerebral artery (MCA) and one in the anterior communicating artery (ACoA). The other two patients had wide neck aneurysms at the tip of the basilar artery (BA). Patient specific models were constructed from 3DRA images. For patient 1, bilateral 3DRA images were used to create the vascular model including the left and right ICAs and MCAs. Figure 4.4 shows the anatomical models and Figure 4.5 shows the various endovascular stenting options possible for treating the aneurysm.

For each aneurysm, different treatment options were considered. Each option consisted in deploying one or two Neuroform stents in different ways. For the ACoA aneurysm of patient 1, the first option was to place a stent from the A1 segment of left anterior cerebral artery (ACA) to the A2 segment of the left ACA. The second option consisted in placing the stent from the left A1 to the right A1. For the MCA aneurysm and for the BA aneurysms of patients 2 and 3 the first option was to place a stent from the parent artery to one of the daughter branches. The second option was to place the stent from the parent artery to the other daughter branch, and the third option was to place two stents from the parent artery to each of the daughter branches “Y” (stent in stent technique). A total of 6 simulations were carried out for this analysis. The flow patterns at peak systole for each aneurysm before stenting and after treatment with each option are presented in Figure 4.6. The corresponding distributions of WSS are presented in Figure 4.7.

The ACoA aneurysm of patient 1 is fed from the left ACA with a fairly wide inflow jet that impacts on the back wall of the aneurysm producing a relatively high WSS in the body and dome of the aneurysm. When the stent is placed from the left A1 to the left A2, the inflow jet is disrupted causing a significant alteration in the intraaneurysmal flow pattern and reduction of the WSS. However, there is still a region of elevated WSS at the same impaction zone. When the stent is placed across the neck of the aneurysm (from the left A1
to the right A2) the inflow jet is slightly deviated towards the dome of the aneurysm and the major flow structures within the aneurysm are not significantly modified. The region of elevated WSS is found at the dome of the aneurysm. A second simulation with this option was carried out after rotating the stent half a cell. The results are shown in the top-right corner panel of Figures 4.6 and 4.7. In this case, there is more obstruction of the inflow jet,
a less complex velocity pattern, and a larger reduction of the WSS in the aneurysm.

In the MCA aneurysm of patient 1, the inflow jet impacts on the dome of the aneurysm creating a region of elevated WSS along the body and dome of the aneurysm. When
deploying the stent from the parent vessel to the left branch, the inflow jet is slightly
diffused, the intra-aneurysmal vortical structures are modified, and the WSS is reduced at
the dome of the aneurysm. At the same time the inflow into the smaller MCA aneurysm
is also reduced. When the stent is deployed to the other daughter (right) branch, there
is a larger obstruction of the inflow jet and a more complex intraaneurysmal flow pattern
with an overall lower WSS in the aneurysm. However, the inflow into the smaller MCA
aneurysm is increased. When deploying one stent to each of the daughter branches, the
intra-aneurysmal flow pattern displays new vortical structures, i.e. more complex flow
pattern, and lower WSS in the aneurysm. At the same time the inflow into the smaller
MCA aneurysm is also reduced. However, in this case there is an increase of WSS in the
outflow part of the neck of the aneurysm.

In the BA tip aneurysm of patient 2, the inflow jet impacts on the body of the aneurysm
producing a region of elevated WSS on the right side of the aneurysm and then splits and
exits towards the left and right posterior cerebral arteries (PCA). All stenting options
produced significant disruption of the inflow jet, reduction of the intra-aneurysmal velocity,
alteration of the flow structure, and reduction of the WSS in the aneurysm. However, when
deploying a single stent from the BA to the left PCA (column 3 of Figures 4.6 and 4.7)
there is only a slight increase of the WSS at the outflow part of the neck compared to the
other two cases.

In the BA tip aneurysm of patient 3, a concentrated inflow jet impacts on the dome of
the aneurysm producing a region of high WSS at the dome and a complex intra-aneurysmal
flow pattern. When stenting from BA artery to the right PCA, the inflow jet is slightly
deviated to the right but there is still a region of high WSS at the dome. When stenting
from the BA to the left PCA, the inflow jet appears to be more concentrated and impacting
on a smaller region of the dome. When two stents are deployed, one into each PCA, the
inflow jets is diffused, the intra-aneurysmal flow structures are modified, and the WSS in
the dome is significantly reduced. However, in all cases an increase of WSS is observed in
the outflow regions of the stented aneurysm neck.
In conclusion, these computations demonstrate that with the techniques described earlier it is possible to qualitatively assess the effects of different treatment plans and that these effects are not always intuitive.
Figure 4.7: Wall shear stress distributions before and after the different stenting options.
4.3 Quantitative Methods

A quantitative assessment of the change in intra-aneurysmal flow dynamics after stenting can be conducted by calculating and comparing quantities such as velocity, kinetic energy, vorticity, wall shear stress and mass flux into the aneurysm to the pre-stent configuration. This is possible only if the velocity values inside the aneurysm can be segregated from the rest of the domain. Algorithms were developed for each stage of the segregation process as shown in Figure 4.8. Briefly explaining the procedure involved, first the elements and their points surrounding the neck of the aneurysm are identified and deleted. Next the aneurysm was deleted leaving the patient-specific model with a big hole. Then the hole was closed with a smooth surface that forms the neck dividing the aneurysm from the main vessel. Finally, a code was developed to use this neck to mask all the points inside the aneurysm and calculate the above mentioned quantities.

The procedure described above was demonstrated on three patient-specific aneurysm models. Simulations were performed with and without the stent to calculate the change in intra-aneurysmal flow dynamics. The Pipeline stent was used for the stented cases. Figure 4.9 shows the streamline and WSS results and Table 4.1 gives the various hemodynamic quantities calculated from these simulations. The values in the brackets show the percentage reduction of that particular quantity compared to the pre-stent case. A qualitative comparison of the streamlines from the pre and post-stented simulations for the three models show how the stent alters the chaotic intra-aneurysmal flow to a smooth steady low-velocity flow. Also a reduction in WSS can be seen for the stented cases in comparison to the pre-stented cases. The values in Table 4.1 confirm these qualitative results quantitatively.
Table 4.1: Quantitative values for the 3 models. Model 1: figure 4.8, left column; Model 2: figure 4.8, middle column; Model 3: figure 4.8, right column.

<table>
<thead>
<tr>
<th></th>
<th>Model 1</th>
<th>Model 2</th>
<th>Model 3</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>pre-stent</td>
<td>post-stent</td>
<td>pre-stent</td>
</tr>
<tr>
<td>max. velocity (cm/s)</td>
<td>26.5</td>
<td>6.1</td>
<td>39.2</td>
</tr>
<tr>
<td></td>
<td>(77%)</td>
<td>(84%)</td>
<td>(65%)</td>
</tr>
<tr>
<td>mean velocity (cm/s)</td>
<td>2.7</td>
<td>0.57</td>
<td>16.1</td>
</tr>
<tr>
<td></td>
<td>(79%)</td>
<td>(91%)</td>
<td>(94%)</td>
</tr>
<tr>
<td>total kinetic energy (erg)</td>
<td>31.2</td>
<td>1.6</td>
<td>1.7</td>
</tr>
<tr>
<td></td>
<td>(95%)</td>
<td>(99%)</td>
<td>(99%)</td>
</tr>
<tr>
<td>mean vorticity (s⁻¹)</td>
<td>21.2</td>
<td>3.6</td>
<td>523.0</td>
</tr>
<tr>
<td></td>
<td>(83%)</td>
<td>(93%)</td>
<td>(94%)</td>
</tr>
<tr>
<td>max. wall shear stress (dyne/cm²)</td>
<td>30.2</td>
<td>9.1</td>
<td>77.6</td>
</tr>
<tr>
<td></td>
<td>(70%)</td>
<td>(83%)</td>
<td>(83%)</td>
</tr>
<tr>
<td>mean wall shear stress (dyne/cm²)</td>
<td>2.0</td>
<td>0.25</td>
<td>36.1</td>
</tr>
<tr>
<td></td>
<td>(88%)</td>
<td>(94%)</td>
<td>(94%)</td>
</tr>
<tr>
<td>mass flux into the aneurysm (ml/s)</td>
<td>1.77</td>
<td>0.8</td>
<td>4.7</td>
</tr>
<tr>
<td></td>
<td>(55%)</td>
<td>(99%)</td>
<td>(99%)</td>
</tr>
</tbody>
</table>
Figure 4.8: Steps in the aneurysm segregation process.
Figure 4.9: Models, streamlines and WSS results for the pre and post cases.
4.4 Summary

A hybrid body-fitted and embedded gridding technique and the stent deployment algorithm explained in the previous chapters were used to simulate subject-specific cases with various stent designs. The results indicate that the same stent can produce different alterations of the hemodynamics in different aneurysms. The flow modifications largely depend on the inflow characteristics of the aneurysm, which in turn depends on the location of the aneurysm and the geometry of the parent vessel. Secondary flows, induced by the turns and tortuosity of the parent artery, play an important role in determining the way in which the flow enters the aneurysm. The study on the different treatment options and methods to quantitatively assess the hemodynamics inside the aneurysm provide additional tools to use such simulations in planning treatment options, designing better flow diverters and understanding the outcome of endovascular interventions.
Chapter 5: Flow Alteration in Side Branches

The main concern about using stents in cerebral arteries is the possibility of occluding a small branch or perforator artery and causing an ischemic stroke. Therefore calculating the flow reduction through these branches is critical in understanding the effect of the deployed stent on cerebral blood circulation. This chapter describes a methodology developed to calculate flow alterations in perforators and side branches after stenting and reports preliminary results obtained with patient-specific results.

5.1 Boundary Condition Methods

The prediction of flow reduction in small arteries branching off from the parent artery when a stent is deployed requires appropriate outflow boundary conditions. Clearly, prescribing the flow rate at the outflow boundary would not be realistic as this would assume no change in the flow rate from the pre-stented case which is what we are interested in estimating. Imposing traction-free boundary conditions assumes that the distal vascular beds have similar flow impedances which is not a reasonable approximation for a large vessel such as the carotid artery and a perforator. Boundary conditions based on flow impedances or resistances of the distal vascular beds seem to be a more appropriate choice. In order to prescribe resistance boundary condition we created cylindrical extrusions at each outflow boundary and changed the viscosity of these blocks to obtain the desired flow resistance. The viscosity is computed based on Poiseuille flow. This method produces the desired flow divisions without destabilizing the incompressible flow solution.

\[ \mu = \frac{\pi r^4 R}{8L} \]  \hspace{1cm} (5.1)
where $\mu$ is viscosity, $R$ is resistance, $r$ is radius of the vessel and $L$ is the length of the extrusion.

$$R = \frac{\delta P}{Q} \quad (5.2)$$

where $\delta P$ is pressure difference and $Q$ is the flow rate.

The relative values of the resistances used for each outflow boundary were calculated in order to achieve a flow division that did not produce a substantial change in the wall shear stress from the value of the parent artery. This assumes that the arterial network is close to the optimal design point given by Murray’s law based on the principle of minimal work [62]. After adjusting the resistance values in this way for the pre-stenting models, these values were kept constant for the post-stenting calculations. This assumes that the resistance values of the distal vascular beds remain the same after stenting.

Similar to simulations conducted in other chapters pulsatile physiologic flow conditions were prescribed at the inlet boundaries.

### 5.2 Perforators

A model of an aneurysm in a cerebral artery with a perforator was created by adding a perforating branch to model of section 4.3 in chapter 4. The perforator had a diameter of 0.03 cm and a length of 3 vessel diameters. The dimensions of the perforator were taken from typical values reported in the literature [63]. The perforator was attached to the parent vessel in two ways: a) smooth connection, and b) sharp connection (Figure 5.1). Pulsatile flow conditions were imposed at the inflow and resistance boundary conditions were prescribed at the outflow of the perforator and the parent vessel as described above. The resistance values were tuned so that the WSS on the perforator is of similar magnitude compared to the parent vessel. Figure 5.2 shows the WSS on the model and the perforator. The stent wire which has a diameter of 0.01 cm occluded the perforator by about 30%. The diameter of this wire was then increased to cover the inflow area of the perforator approximately by 60%, 70% and 90%. Figure 5.3 shows the flow pattern into the perforator.
for the pre-stent and post-stent cases. Figure 5.4 shows the percentage flow rate reduction with respect to percentage change in the inflow area. These results indicate that more than 90% occlusion is necessary to obtain a change in flow rate of about 7%. This can be explained from the fact that small arteries such as perforators have a much higher distal resistances and thus it is necessary to occlude a large percentage of its origin in order to create a local resistance at the origin comparable to the distal resistance and therefore alter the flow rate significantly. Since in the case of the smooth connection the area of the origin of the perforator is larger, much lower flow reductions were obtained.
Figure 5.1: Perforator attachment to the parent vessel. Smooth connection (top), Sharp connection (bottom).
Figure 5.2: WSS in the model with the perforator.

68
Figure 5.3: Flow through perforator. Left column: funnel type; Right column: sharp edge; Top row: pre-stent; Bottom row: 30%-50% occlusion
Figure 5.4: Flow reduction in perforator.
5.3 Side branches

Since bigger arteries have lower resistance values compared to the perforators and therefore larger changes may be expected, simulations were performed on model 2 and 3 from section 4.3 in chapter 4 to study the effect of occlusion on side branches such as the ophthalmic artery. The inflow to the side branch was approximately covered by 50%, 70% and 90%. Figure 5.5 shows the flow through the ophthalmic artery for the two models and figure 5.6 and 5.7 shows the percentage flow rate reduction with respect to percentage change in the inflow area.
Figure 5.5: Flow through ophthalmic. Left column: Model 2; Right column: Model 3; Top row: model with stent; Center row: pre-stent; Bottom row: 90% occlusion
Figure 5.6: Flow reduction in Model 2.

Figure 5.7: Flow reduction in Model 3.
5.4 Summary

In this chapter a method was developed to simulate flow in jailed vessels. An occlusion of more than 90% changed the flow rate only by about 7%. This may be due to the fact that since perforators have high distal resistances, a 90% occlusion could not create a resistance of similar magnitude to bring a significant change in the flow rate. In the case of the ophthalmic artery a 90% occlusion brought about a change in flow rate of about 0.5% in model 2 and 3.2% in model 3.
Chapter 6: Conclusions

This thesis involved using the embedded/immersed method to simulate blood flow around endovascular stents in subject-specific cerebral aneurysms. In the first part of the thesis this method was evaluated by comparing it with the traditional body-fitted method and experimental data. Results from flow past a circular cylinder and an idealized stented aneurysm were found to be in close agreement with body-fitted computations and experimental results.

Next, a stent deployment algorithm was presented which can be used to deploy stents in a given patient-specific model and perform endovascular stent simulations. ‘Virtual’ angiograms were performed and validated by comparing them to conventional angiograms. From these validations it was clear that patient-specific simulations are capable of realistically representing the intra-aneurysmal flow dynamics before and after stenting and hence these simulations can be used to design better stents as flow diverters and plan endovascular treatment options. A sensitivity study was done to study the effects of stent positioning on flow patterns inside the aneurysm. From this study it was concluded that if the stents are highly porous then the flow diverting characteristics will depend if the stent in the deployed position is able to block the flow into the aneurysm. But if the stent has more wires, i.e. smaller cells then there is more chance of disrupting the inflow jet independently of their positioning.

Subject-specific simulations were performed with various stent designs in order to study the flow alterations caused by the stents in the aneurysms. The results indicate that the same stent can produce different alterations in different aneurysms and that the change in intra-aneurysmal flow depends on the inflow characteristics of the aneurysm which in turn depends on the location of the aneurysm and the parent vessel. The inflow into the aneurysm also depends on the secondary flows which are developed due to the turns and tortuosity of the parent vessel. Additionally the study on the various treatment options showed how
the methodology can be used to select the best therapeutic option in treating an aneurysm. The methods to quantify the flow dynamics inside the aneurysm provide an additional tool to evaluate the effectiveness of a stent.

Finally, the effect of the deployed stent on flow rate reduction in perforator and the ophthalmic artery was also investigated. The results show that in the case of the perforator more than 90% occlusion is required to obtain a flow rate reduction of about 7% and in the case of the ophthalmic artery more than 90% occlusion changed the flow rate by only about 3% for one of the models. The conditions under which such 90% occlusion happens in reality due to stent deployment needs to be further investigated.
Bibliography
Bibliography


Curriculum Vitae

Sunil Appanaboyina was born on July 19th, 1980 in Sydapuram in the state of Andhra Pradesh, India. He finished his undergraduate studies in Mechanical Engineering at University of Madras, India, in 2001. And Masters in Mechanical Engineering from Lamar University, Texas, in 2003. Currently he is finishing his PhD in Computational Science and Informatics from George Mason University, Virginia, and will be graduating in August 2008.