Influence of different computational approaches for stent deployment on cerebral aneurysm haemodynamics

Annarita Bernardini1,2, Ignacio Larrabide1,2,*, Hernán G. Morales1,2, Giancarlo Pennati3, Lorenza Petrini3, Salvatore Cito1,2 and Alejandro F. Frangi1,2,4

1 Center for Computational Imaging and Simulation Technologies in Biomedicine (CISTIB), Universitat Pompeu Fabra (UPF), C/Roc Boronat 138, 08018 Barcelona, Spain
2 Networking Center on Biomedical Research (CIBER-BBN), C/Roc Boronat 138, 08018 Barcelona, Spain
3 Laboratory of Biological Structure Mechanics, Politecnico di Milano, Piazza Leonardo da Vinci 32, 20133 Milan, Italy
4 Institució Catalana de Recerca i Estudis Avançats (ICREA), C/Roc Boronat 138, 08018 Barcelona, Spain

Cerebral aneurysms are abnormal focal dilatations of artery walls. The interest in virtual tools to help clinicians to value the effectiveness of different procedures for cerebral aneurysm treatment is constantly growing. This study is focused on the analysis of the influence of different stent deployment approaches on intra-aneurysmal haemodynamics using computational fluid dynamics (CFD). A self-expanding stent was deployed in an idealized aneurysmatic cerebral vessel in two initial positions. Different cases characterized by a progression of simplifications on stent modelling (geometry and material) and vessel material properties were set up, using finite element and fast virtual stenting methods. Then, CFD analysis was performed for untreated and stented vessels. Haemodynamic parameters were analysed qualitatively and quantitatively, comparing the cases and the two initial positions. All the cases predicted a reduction of average wall shear stress and average velocity of almost 50 per cent after stent deployment for both initial positions. Results highlighted that, although some differences in calculated parameters existed across the cases based on the modelling simplifications, all the approaches described the most important effects on intra-aneurysmal haemodynamics. Hence, simpler and faster modelling approaches could be included in clinical workflow and, despite the adopted simplifications, support clinicians in the treatment planning.

Keywords: computational fluid dynamics; virtual stenting; self-expanding stent; cerebral aneurysm

1. INTRODUCTION

Cerebral aneurysms are abnormal focal dilatations of the artery walls, typically present near or at the bifurcations of the circle of Willis. It is estimated that worldwide 2–5% of the population harbours one or more cerebral aneurysms [1]. Statistics indicate that around 2 per cent of cerebral aneurysms will rupture, resulting in subarachnoid haemorrhage, with 50 per cent of these leading to death [2]. Aneurysms are usually asymptomatic, but the greater improvement and availability of non-invasive neuroradiological imaging techniques make their incidental discovery more and more frequent. Once an aneurysm is detected, preventive and elective intervention is usually followed to avoid its rupture [3]. Among treatment options, such as clipping, coiling or stent-supported coiling, clinicians select the most appropriate therapy for the patient to exclude the aneurysm from the blood circulation.

Several computational tools are being developed to help clinicians in the decision-making and treatment planning process [4–8]. These tools should produce accurate and clinically meaningful results, with a high level of usability and automation at low computational costs. Some of these tools allow endovascular devices to be deployed virtually inside patient-specific vascular geometries [9–12]. They can allow computational fluid dynamics (CFD) to be performed simulations before and after treatment to analyse the haemodynamic alterations induced by the implantation of the device.

Normally, the outcome of endovascular treatment (including stenting) is assessed by the use of angiographic
Virtual stenting and haemodynamics

A. Bernardini et al.

On the other hand, Larrabide

room. Furthermore, their application in clinical practice

too much computationally demanding for an operating

methods are not user-friendly for clinicians and are

interaction play an important role. Nevertheless, these

and the vessel and the stress field generated by this

approaches was compared with different equally prob-

relevance of the differences found throughout the

is a loss of accuracy in the released stent configuration

using only FE methods, and down to few seconds

unit (CPU) was used for processing the simulations

geometry of a cerebral vessel with an aneurysm neck

an early stage if the treatment has a positive effect or

not, thus requiring an additional treatment, specific

follow-up or if the patient can be discharged. Recently,

the simulation of an angiography trough the virtual

injection of a contrast agent (dye) has been included in

CFD models to give clinicians and researchers a closer

and additional instrument for treatment evaluation

[11,14,15].

Regarding the treatment of cerebral aneurysms with

stents, these are frequently used to provide support for

the coil in wide-necked aneurysms. In fact, it was

proposed that stents could prevent aneurysm

re-canalization after coiling by reducing coil migration

and residual ostium [16–19]. Moreover, it has

been found that the position of struts and links

above the ostium (configuration of the released stent)

has a considerable impact on the resulting blood

flow [19–25].

The mechanics of the stent release can be mathem-

atically described by mean of partial differential

equations that can be solved using finite-element (FE)

methods. These methods are used extensively in simu-
lating the deployment of stents in stenotic arteries

[26–31], where the forces exchanged between the stent

and the vessel and the stress field generated by this

interaction play an important role. Nevertheless, these

methods are not user-friendly for clinicians and are

too much computationally demanding for an operating

room. Furthermore, their application in clinical practice

on patient-specific anatomies poses major challenges.

On the other hand, Larrabide et al. [11] recently pro-
posed a method, called fast virtual stenting (FVS),
capable of simulating the expansion of stents into

patient-specific aneurysmatic vessels. This method is

based on deformable meshes [32] and simple geometri-
cal constraints, which neglect material properties. It is

semi-automatic, easy to implement and drastically

reduces the computational cost of virtual stenting.

Bernardini et al. [33] investigated the effect of simpli-
fications in computational approaches for simulating

the stenting on computational time and accuracy of

results, using both FE and FVS methods. An open-
cell self-expanding stent was released in an idealized

geometry of a cerebral vessel with an aneurysm neck

by progressively simplified stent deployment

approaches. The time for which a central processing

unit (CPU) was used for processing the simulations

(CPU time) decreased from tens of thousands to hun-
dreds of seconds by simplifying the stent morphology

using only FE methods, and down to few seconds

using the FVS method. They also identified that there

is a loss of accuracy in the released stent configuration

by neglecting material properties. Nevertheless, the

relevance of the differences found throughout the

approaches was compared with different equally prob-

able initial positions of the stent but not evaluated

from the haemodynamic point of view.

Alternatively to this previous work, in the current

study, FE and FVS methods for stent deployment

were used to obtain the deployed stent representation

and the main focus was on CFD simulations performed

for each approach. The purpose of this work is to under-

stand the influence of the stent modelling approach on

intra-aneurysmal haemodynamics.

2. MATERIAL AND METHODS

Models of a self-expanding stent and of an idealized

aneurysmatic cerebral vessel were considered. First,

the stent was virtually placed using five progressively

simplified deploying approaches, four using the FE

method and one using the FVS method. Then, CFD

simulations were performed in stented vessels and in

the untreated vessels (UVs).

2.1. Geometrical models

The geometrical models were generated using the com-

mercial computer-aided design (CAD) software

RHINOCEROS v. 4.0 (McNeel & Associates, Indianapolis,

IN, USA). A geometrical model of a self-expanding

Neuroform stent (Boston Scientific, Natick, MA,

USA) obtained from a micro-CT scan was created.

This device is a high-porosity non-symmetric open-cell

stent, and it has been selected because these features

better highlight the differences among different released

positions when compared with symmetric closed-cell

stents. The stent is composed of four rings of 14

struts each, connected by 12 links, three between

pairs of rings. Its total length is 10 mm and the nominal

diameter is 4.5 mm. Struts and links have a length of

2.43 and 0.26 mm, respectively, with a rectangular

cross section of 0.1 × 0.08 mm², as presented in

figure 1a. This model was represented in two ways:

keeping the morphology (cross section and shape of

strut endings) of the three-dimensional components

(three-dimensional-strut stent model, figure 1a), and

giving a uni-dimensional description of the stent

struts (one-dimensional-strut stent model, figure 1c).

One-dimensional-strut stent considers only the centre-

lines of struts and links, and approximates them as

straight lines (figure 1b).

The vascular geometrical model corresponds to an

idealized aneurysmatic vessel previously used by

Bernardini et al. [33]. The vessel geometry was com-

pleted with an extrusion of its endings, keeping the

vessel diameter of 3.6 mm and its curvature radius of

30 mm. A side-wall aneurysm was also added. The

aneurysm had an aspect ratio (AR), dome-depth to

neck ratio, of 1.8 (figure 1d); treatment is highly

recommended for this type of aneurysm [34].

2.2. Stent deployment

Five modelling approaches were adopted to deploy the

stent in the vessel. A progression of simplifications on

the morphology of the stent and on material properties

was considered. Table 1 summarizes the main features

of the five cases. Two relative positions of the stent

struts with respect to the ostium were considered: asym-

metrical (posA) and symmetrical (posB) (figure 1c),

both of them obtained by translating and rotating the

strut endings) of the three-dimensional components

(three-dimensional-strut stent model, figure 1a), and

giving a uni-dimensional description of the stent

struts (one-dimensional-strut stent model, figure 1c).

One-dimensional-strut stent considers only the centre-

lines of struts and links, and approximates them as

straight lines (figure 1b).

The vascular geometrical model corresponds to an

idealized aneurysmatic vessel previously used by

Bernardini et al. [33]. The vessel geometry was com-

pleted with an extrusion of its endings, keeping the

vessel diameter of 3.6 mm and its curvature radius of

30 mm. A side-wall aneurysm was also added. The

aneurysm had an aspect ratio (AR), dome-depth to

neck ratio, of 1.8 (figure 1d); treatment is highly

recommended for this type of aneurysm [34].

2.2. Stent deployment

Five modelling approaches were adopted to deploy the

stent in the vessel. A progression of simplifications on

the morphology of the stent and on material properties

was considered. Table 1 summarizes the main features

of the five cases. Two relative positions of the stent

struts with respect to the ostium were considered: asym-

metrical (posA) and symmetrical (posB) (figure 1c),

both of them obtained by translating and rotating the
stent at the un-deployed position. These two positions were chosen since they showed the greatest difference in the released stent configuration [33].

Four cases were deployed using the FE method. Three-dimensional- and one-dimensional-strut stents were deployed in deformable (FE3H, FE1H) and rigid (FE3R, FE1R) vascular walls. The self-expanding Neuroform stent is made of Nitinol, a nickel–titanium shape memory alloy. The pseudo-elastic material property of the Nitinol was described using average values taken from literature [29,31] by introducing a previously developed user subroutine [35] (figure 2a). The vessel was modelled as either a deformable or rigid body. The deformable wall was described as a single homogeneous isotropic hyperelastic layer of 0.3 mm of thickness. The material parameters were derived from experimental data on cerebral vessels [36] (figure 2b). The simplification of characterizing the vessel wall by its whole stiffness is reasonable as it was clinically observed that stent insertion does not modify the shape and dimension of the vessel [16], suggesting a modest influence of the stent on the vascular wall deformation.

The commercial code ABAQUS/Standard 6.8EF (Simulia Corp, Providence, RI, USA) was used to perform a large deformation analysis for the FE-based simulations. The classical displacement formulation solved by Newton’s method was adopted. The stent deployment was simulated in two steps: crimping and release. A uniform inward radial displacement was applied to the nodes of the internal surface of the stent to fit it completely inside the vessel (crimping step). The stent was constrained at nodes in the endings of each strut to avoid rotation and at nodes in the middle of the three links of the central section to avoid translation. Then, by removing the constraints in the radial direction, the stent was allowed to self-expand until it contacted the vessel wall (release step). A stent–vessel wall contact

<table>
<thead>
<tr>
<th>case abbreviation</th>
<th>FE3H</th>
<th>FE3R</th>
<th>FE1H</th>
<th>FE1R</th>
<th>FVS</th>
</tr>
</thead>
<tbody>
<tr>
<td>method</td>
<td>FEM</td>
<td>FEM</td>
<td>FEM</td>
<td>FEM</td>
<td>FVS</td>
</tr>
<tr>
<td>strut and link description</td>
<td>3D</td>
<td>3D</td>
<td>1D</td>
<td>1D</td>
<td></td>
</tr>
<tr>
<td>stent material</td>
<td>Nitinol</td>
<td>Nitinol</td>
<td>Nitinol</td>
<td>Nitinol</td>
<td>—</td>
</tr>
<tr>
<td>vessel material</td>
<td>deformable, hyperelastic</td>
<td>rigid body</td>
<td>deformable, hyperelastic</td>
<td>rigid body</td>
<td>rigid body</td>
</tr>
</tbody>
</table>

Figure 1. (a) Three-dimensional-strut stent (rectangular cross section: \(a = 0.1\) mm and \(b = 0.08\) mm for the struts; \(a = 0.34\) mm and \(b = 0.08\) mm for the links). (b) Details of the approximation of one-dimensional-strut stent. (c) One-dimensional-strut stent (circular cross section: \(d = 0.1\) mm for struts and links). (d) Geometrical model of the idealized aneurysmatic cerebral vessel (\(D_v = 3.6\) mm; \(R = 30\) mm; \(H = 8.3\) mm; \(D_n = 4.5\) mm). (e) Relative positions of the stent with respect to the aneurysm ostium: asymmetric (posA) and symmetric (posB) link.

Figure 2. (a) Pseudo-elastic model for Nitinol. (b) Hyperelastic constitutive law for the vessel, based on a second-order reduced polynomial strain energy density function (parameters: \(v = 0.49; \ D_1 = D_2 = 0; \ C_{10} = 3.3096; \ C_{20} = 0.6353, \ C_{01} = -3.0197, \ C_{11} = 0.5814, \ C_{02} = -0.2745\)).
was managed using a soft exponential frictionless interaction [37]. To prevent rigid motions, the vessel was fixed at its endings during the whole deployment. The three-dimensional- and one-dimensional-strut stent models were discretized in $8 \times 10^4$ 8-node brick elements and $4 \times 10^2$ 2-node linear beam elements, respectively. The beam elements of the one-dimensional-strut stents were modelled with circular cross section of 0.1 mm of diameter (figure 1c). The vessel wall was discretized in $10^4$ 3-node shell elements.

The fifth stent case used the FVS method of deployment (figure 3). The struts and their connectivity were defined over a subset of the points of a 2-simplex mesh with a size of $32 \times 20$. No material properties were defined for the stent, and the vessel was described as a rigid body. The simplex mesh was arranged as a cylinder of 2.6 mm of diameter and aligned along the curved vessel centreline. The expansion of the stent is governed by a system of second-order differential equations (2.1) discretized with finite differences:

$$
\rho \frac{\partial^2 p_i(t)}{\partial t^2} + \gamma \frac{\partial p_i(t)}{\partial t} - \alpha f_{int}(p_i(t)) + \chi f_{length}(p_i(t)) = (1 - \alpha)f_{ext}(p_i(t)),
$$

where $t$ is an artificial time describing time evolution and $p_i$ refers to the $i$th point of the simplex mesh, $f_{int}$ represents the internal force accounting for mesh smoothness and the mesh expansion force, $f_{length}$ represents the stent-shape constraining force, and $f_{ext}$ accounts for the interaction between the stent and the vessel wall. The constants $\alpha$ and $\chi$ are weighting parameters for the internal–external forces and constraints, respectively, set as $\alpha = 0.5$ and $\chi = 0.9$ according to the work of Larrabide et al. [11]. The interaction between the stent and the wall is defined by the interruption of the expanding force when the mesh reaches a distance of half the strut radial thickness from the vessel wall.

2.3. Haemodynamic model

Before CFD simulations, three geometrical processes were applied. First, an ostium plane that distinguishes the aneurysm dome from the vessel lumen was defined (figure 4a). This plane was the same for all the CFD cases and did not intersect stent struts. Second, an automatic tubular three-dimensional surface of 0.1 mm of diameter was automatically created around the lines of the deformed one-dimensional-strut stents. Third, the stent was cut and the struts lying entirely on the vessel wall were removed (figure 4b). In literature it has been suggested that using the patch of the stent over the ostium instead of the whole stent reduces the computational cost of the CFD simulations without inducing important intra-aneurysmal haemodynamic variations [20].

The CFD analysis was performed using ANSYS CFX v. 12.0 (ANSYS Inc., Canonsburg, PA, USA) to solve the unsteady three-dimensional Navier–Stokes...
equations, consisting of the continuity equation (2.2) and the momentum equation (2.3):

$$\nabla \cdot \mathbf{v} = 0$$  \hspace{2cm} (2.2)

and

$$\rho \left( \frac{\partial \mathbf{v}}{\partial t} + \mathbf{v} \cdot \nabla \mathbf{v} \right) - \nabla \cdot \mu \nabla \mathbf{v} = - \nabla p.$$  \hspace{2cm} (2.3)

where \( \mathbf{v} \) is the velocity vector, \( \rho \) the fluid density, \( t \) the time, \( \mu \) the fluid dynamic viscosity and \( p \) the pressure. The blood was assumed to be a Newtonian incompressible fluid, with \( \mu = 0.004 \text{ Pa } \cdot \text{s} \) and \( \rho = 1060 \text{ kg} \cdot \text{m}^{-3} \).

To simulate an angiography, the injection of the dye was performed by solving a transport (convection-diffusion) equation (2.4):

$$\frac{\partial C}{\partial t} + \mathbf{v} \cdot (\nabla C) - k \nabla^2 C = 0,$$  \hspace{2cm} (2.4)

where \( C \) is the scalar field of dye concentration and \( k \) is the diffusion coefficient. The dye was modelled as a massless scalar passively transported by the fluid using a negligible diffusion coefficient \( (k = 10^{-15} \text{ m}^2 \cdot \text{s}^{-1}) \). No flow rate change owing to the injection of the dye was considered. The temporal terms in the equations were discretized by using an Euler implicit scheme, and the spatial terms were discretized with second-order accuracy.

The assumption of a rigid wall with no-slip boundary condition was imposed on the vessel wall and stent. A constant pressure was applied to the outlet. At the inlet, a physiological curve of mean cross-sectional velocity was imposed \cite{15} (figure 5). This velocity was characterized by a parabolic profile at the inlet and a time-averaged Reynolds number of 250. Four cardiac cycles were computed in order to simulate the angiography. For the quantitative analysis, only the results of the last cardiac cycle were taken into account.

A time independence analysis has been performed. Four time steps \((0.01 \text{ s}, 0.005 \text{ s}, 0.001 \text{ s} \text{ and } 0.0005 \text{ s}) \) were tested. A difference of less than 1 per cent of (2.5) and (2.9) between the three smallest time steps has been observed, and the value of 0.005 s of time step has been used.

A root mean square (r.m.s.) of the residuals equal to 5 has been imposed as a criterion of convergence of the numerical solution.

The CFD cases were discretized with unstructured grids composed of tetrahedral elements, generated by ANSYS ICEM CFD v. 12.0 (ANSYS Inc). Three prism layers were applied at the aneurysm wall. A mesh independency analysis was performed, testing six different mesh sizes. To choose the mesh size, the criteria of reaching the accuracy of more than 99.5 per cent on (2.5) and (2.9) was used. From this, a global element size of 0.3 mm was selected, except at the vessel and aneurysm wall \((0.15 \text{ mm}) \) and around the stent struts \((0.02 \text{ mm}) \) \cite{38} (figure 4c). Finally, UV and stented cases resulted in around \( 4 \times 10^6 \) and \( 7.9 \times 10^6 \) mesh elements, respectively.

To study the flow in the aneurysm, for each case, the average WSS \( \text{WSS}(t), (2.5) \), the average velocity \( \bar{v}(t), (2.7) \) and the mass inflow rate \( \bar{m}_{\text{IR}}(t), (2.9) \) over the inflow region were considered. The inflow region was defined as the area on the ostium plane where the flow enters the aneurysm \cite{22}. The WSS\% \((2.6), \nu\% (2.8) \) and \( \bar{m}_{\text{IR}}\% (2.10) \) are the percentages, averaged over time, of (2.5), (2.7) and (2.9) in the stented cases (S) with respect to the corresponding in UV. These parameters are widely used in literature to characterize the intra-aneurysmal haemodynamic conditions regarding aneurysm genesis, growth and rupture \cite{19,39}.

$$\text{WSS}(t) = \frac{\int_{A_w} \text{WSS}(t) \, dA_w}{A_w},$$ \hspace{2cm} (2.5)

$$\text{WSS\%} = \frac{\int_{0}^{t} (\text{WSS}(t)/\text{WSS}(t)_{\text{UV}}) \, dt}{v},$$ \hspace{2cm} (2.6)

$$\bar{v}(t) = \frac{\int_{V_D} v(t) \, dV_D}{V_D},$$ \hspace{2cm} (2.7)

$$\nu\% = \frac{\int_{0}^{t} (\bar{v}(t)/\bar{v}_{\text{UV}}) \, dt}{c},$$ \hspace{2cm} (2.8)

$$\bar{m}_{\text{IR}}(t) = \rho \cdot \int_{A_m} \bar{v}(t) \, dA_m,$$ \hspace{2cm} (2.9)

$$\bar{m}_{\text{IR}}\% = \frac{\int_{0}^{t} (\bar{m}_{\text{IR}}(t)/\bar{m}_{\text{IR}}(t)_{\text{UV}}) \, dt}{c},$$ \hspace{2cm} (2.10)

(W, aneurysm wall; D, aneurysm dome; IR, inflow region; cc, cardiac cycle).

3. RESULTS

3.1. Geometrical findings

Before CFD analysis, a qualitative geometrical comparison across the five stented cases (inter-case) and between the two stent positions (inter-position) was performed. First, the hyperelastic properties of the vessel produced a change of the aneurysm ostium to a more circular shape, and a longitudinal straightening of the vessel in the stented part. Second, FVS produced a different configuration of the stent in both positions compared with the other stented cases, with a greater ring opening and enlarged links (the link above the ostium is four times longer than the original length).
Third, the stent position produced a different stent configuration above the ostium. For example, the asymmetry of the link in posA (figure 6) allowed the struts to protrude towards the aneurysm, as it is represented for FE3H, FE3R and FE1H cases, but not for FE1R and FVS cases. On the other hand, the symmetric link in posB prevented the struts from protruding and none of the FE cases showed significant differences (see [33]).

A quantitative comparison of the stent screen effect was done after the three-dimensional reconstruction of the one-dimensional-strut stents (§2.3). To calculate the screen effect, the pixels of the image of the projection of the struts on the ostium plane were counted using the open source software Image J v. 1.43. The screen effect was computed as the percentage of the ratio of the area covered by the stent over the total area. The cases with three-dimensional-strut stent had the largest screen effect in both posA (20%) and posB (around 17%). The screen effect of one-dimensional-strut FE stents was approximately 6 per cent lower than the three-dimensional-strut one. FVS cases showed the lowest values (12% for posA and 10% for posB), because the three-dimensional reconstruction neglects morphological details (such as the rectangular cross-sectional area of the strut) and the released stent configuration is different. A difference of approximately 6 per cent between FE3H and FE1H and between FE3R and FE1R was observed. PosB produced a mean screen effect of 2.8 per cent lower than posA.

### 3.2. Analysis of computational fluid dynamics results

Figure 7 shows the reduction of intra-aneurysmal haemodynamic parameters owing to the presence of the stents. In posA, the mean values of WSS%, v% and \(\bar{\mu}_R\) of the five stented cases were 43 per cent, 40 per cent and 51 per cent, respectively. The inter-case variability of these parameters ranged between 10 and 15 per cent. In posB, while WSS% and v% were comparable with those of posA, \(\bar{\mu}_R\) was almost 66 per cent. The inter-case variability of these parameters ranged from 12 (WSS%) up to 25 per cent (\(\bar{\mu}_R\)).

All the cases highlighted an increase of \(\bar{\mu}_R\) for posB with respect to posA, while not all the cases predicted a unique relationship between the two positions concerning WSS% and v%. In fact, FE3H, FE3R and FE1H indicated a decrease of WSS% and v% in posB, while an opposite trend is shown by FE1R and FVS.

The inter-position variability varied both across the cases considering each parameter (e.g. v% shows a difference between posA and posB from 0.9% in FE3H case to 18.7% in FVS case) and across the parameters considering each case (e.g. FE3H case shows a difference between posA and posB of 2.5% for WSS%, 0.9% for v% and 19.8% for \(\bar{\mu}_R\)). The inter-case and inter-position variabilities results are comparable (around 15%).

The effect of using a patch of the stent over the ostium on intra-aneurysmal haemodynamic parameters was investigated in the most complex case, FE3H, performing a simulation that retained the whole stent. Results showed that using the patch, the intra-aneurysmal haemodynamic parameters are underestimated by around 1 per cent. Furthermore, a percentile analysis at peak systole (2.4 s) showed that the presence of the stent (either the entire stent—additional case FE3Hc—or the patch—FE3H, FE3R, FE1H, FE1R and FVS) decreases the WSS on the aneurysm wall with respect to UV (figure 8).
3.3. Impact zone and dye concentration

Figure 9 represents the area of vessel and aneurysm wall where WSS is lower than 2 Pa and the area where is higher than 2 Pa at peak systole (2.4 s). The value of the threshold used for distinguishing the two areas was selected to compare the cases for its physiological meaning. In fact, Hoi et al. [22] defined this threshold to identify the area of impact of blood (impact zone) on the aneurysm wall. With respect to UV, all stented cases showed a reduction of the impact zone. Depending on stent position, the shape of the impact zone could be different. PosB produced a wider and flatter high WSS area when compared with posA. For both positions, cases FE1R and FVS showed more evident differences with respect to the other stented cases.

The concentration of the dye in the vessel was measured as the integral of the transported scalar in the volume. Figure 10 shows the dye concentration distribution (a colour map of the concentration of the dye) of a longitudinal and a transversal middle plane of the aneurysmatic vessel at three time steps. As can be seen, aneurysm filling (1.125 s) started from the distal part of the ostium and produced a counter-clockwise vortex. Then, during the wash-out phase (1.7 s), the remaining dye inside the dome was diluted and more layers of the vortex appeared towards the aneurysm tip. After the end of the injection (3 s) in two cardiac cycles, the dye was completely washed out. However, an average of 0.1 per cent of dye concentration still remained in the aneurysm dome. The presence of the stent screened the flow entering the aneurysm, decreasing the concentration of dye in the aneurysm. This phenomenon was well represented by all the cases. At 1.7 s, FE1R and FVS of posA had some differences from the other cases in terms of amount and distribution of the dye. In general, posB showed a higher amount of dye inside the aneurysm compared with posA at the same time (1.7 s). In posB, the dye seemed to have the same distribution in the FE cases with the same vessel properties. Minor differences were visible for FVS.

4. DISCUSSION

This study was performed to evaluate the influence of different stent deployment approaches on intra-aneurysmal haemodynamics. Our results indicate that simulating the deployment of a stent in aneurysmatic vessels, besides the computational approach adopted, enables the representation of important effects on intra-aneurysmal haemodynamics in the successful CFD simulations. It was shown that, independently from the stent deployment approach, the insertion of the Neuroform stent model in an idealized aneurysmatic cerebral vessel produced a reduction of average WSS and average velocity inside the aneurysm of almost 50 per cent.

The five computational approaches investigated in this study were similar to those described in Bernardini et al. [33], although in the present study, a more
comprehensive description of the aneurysmatic vessel was adopted. The obtained released stent and vessel configurations confirmed the previous findings [33]. In particular, there are observable differences in released stent and vessel configuration when neglecting the deformability of the vessel. Even larger differences appear increasing the level of simplifications (FE1R and FVS).

Regarding intra-aneurysmal haemodynamics, two factors influenced the results: the released geometries of the stented cases and the morphology of the stent. These two factors can be evaluated separately considering FE3H, FE3R and FE1H cases. The differences between FE3H and FE3R cases are influenced by the vessel model. It has been previously observed that the curvature and neck size of the vessel have an influence on intra-aneurysmal haemodynamics [19,22]. Comparing FE3H and FE1H cases, the morphology of the stent is the reason for their haemodynamic differences. To better understand the role played by the simplified reconstruction of the one-dimensional-strut stents on intra-aneurysmal haemodynamics, an additional simulation was performed with a simplified three-dimensional-strut stent automatically generated on the bases of the centreline of the deployed FE3H stent. Namely, this new deployed stent had the same configuration of FE3H but the same thickness of FE1H (figure 4d), and resulted in a lower screen effect (almost 6% less than FE3H). Compared with FE3H, the automatic stent reconstruction induced an overestimation of the average WSS, velocity and mass inflow rate of 1.35 per cent, 4.13 per cent and 9.3 per cent, respectively. FE1R and FVS are the two cases that combine the simplifications adopted for the stent deployment and the CFD simulations. Results showed that these two cases have more evident differences, both quantitatively and qualitatively, compared with the others.

The effectiveness of the treatment is usually evaluated by clinicians through qualitative analysis of the concentration of the dye in the aneurysm. Simulating an angiography can give clinicians a more familiar instrument to compare pre- and post-treatment. For this reason, it is important that all the computational approaches should highlight the effect of the stenting procedure. The present study showed that the concentration of the dye inside the aneurysm visibly decreased in all the stented cases.

The Neuroform is an open-cell stent with no symmetry in the disposition of the links, so the screen effect to blood flow and the possibility of protrusion of struts towards the aneurysm depends on the orientation and position of the stent in the vessel [40]. Hirabayashi et al. [41] found that the orientation and position of high-porosity stents influence the amount of flow reduction in the aneurysm. In our study, all the cases highlighted inter-position differences in intra-aneurysmal haemodynamics, even though those which combine simplifications (FE1R and FVS) did not show the same trend between posA and posB in all the parameters analysed, as the others did. Furthermore, they resulted in having greater inter-position variability.

In posB, where the protrusion of struts is prevented, there is a concordance between the haemodynamic parameters and the observed screen effect of the stent. Comparing between cases with the same vessel wall characteristics (FE3H with FE1H, FE3R with FE1R

Figure 10. Dye concentration in the middle cut planes of the frontal and distal side of the vessel.
Virtual stenting and haemodynamics  A. Bernardini et al. 9

and FVS), the lower the screen effect was, the greater values for the different parameters were observed. On the other hand, for posA, the protrusion of the struts made the correlation of the haemodynamics with the screen effect more complex because it was only presented in some of the cases (FE3H, FE3R and FE1H).

Since the orientation and position of the stent are not completely controlled by the clinician, posA and posB are both probable during the intervention. In this regards, the inter-case and inter-position variabilities were comparable in our results, suggesting a sufficient accuracy for all the approaches adopted for stent deployment. Nevertheless, additional investigations with other stents for cerebral aneurysm treatment are mandatory to verify if, despite the observed inter-case variabilities, all the computational approaches allow the performances of various treatments to be distinguished (e.g. different stents).

Frictional forces of the stent on the wall have not been taken into account in the present study and simplified vessel geometry with a regular and smooth wall was used. The authors think that the main findings of the present study, focused on the intra-aneurismal haemodynamics, are not significantly affected by these simplifications. However, future studies will consider a more realistic representation of the vessel and wall–stent interaction. A validation in real cases using medical images would allow a better judgement on the differences between models and reality. Virtual stent deployment together with CFD simulation can provide useful information for cerebral aneurysm interventional planning. It is possible to reduce the computational costs for the stent deployment step by simplifying the approach, keeping acceptable accuracy [33], but still the CFD step is the more time-consuming. Unsteady simulations to capture the pulsatility of the blood flow and obtain the virtual angiography were performed. Their computational time was $10^5$ s, using 15 processors distributed over 8-nodes on a cluster with two Quad-Core Intel Xeon X5355 processors (2.66 GHz) per node sharing 16 GB RAM. This time is compatible with the time available for decision-making in elective interventions, because the treatment is usually scheduled some weeks in advance. To be able to apply these tools just before or during the intervention, the development of more efficient CFD tools or acceptable simplifications in simulations would be mandatory. Currently, some studies are going in this direction, for example, comparing steady state with transient simulations [42,43]. Steady-state simulations might facilitate the inception of haemodynamic simulations into clinical practice by providing the critical information for interventional planning at a lower computational cost.

In summary, five computational approaches for simulating stent deployment have been developed and compared. Below, an outline of the most outstanding features observed with regard to their application in clinical practice is provided:

- **FE3H**: is the most complex case, considering details of the stent shape and material. It also considers mechanical deformation of the vessel on its interaction with the stent, giving a more accurate description of the procedure, but its set-up and simulation time make it impractical for clinical practice today.
- **FE3R**: considers the detailed shape and material of the stent but neglects the mechanical behaviour of the wall, which could be an acceptable simplification in case of aneurysm treatment using Neuroform stents [16]. Its set-up and simulation time are expensive, making it impractical for clinical practice today.
- **FE1H**: simplifies the geometry of the stent but considers its material properties as well as the mechanical behaviour of the vessel wall. It is efficient regarding computational time, but its set-up is time-consuming.
- **FE1R**: stent geometry and vessel wall properties are simplified but the stent material is accurately represented. It is time efficient, but its set-up is time-consuming.
- **FVS**: stent geometry and material properties are simplified. Also, vessel material properties are neglected. On the other hand, it is time efficient and simple to set-up.

Furthermore, the approaches based on FE methods are not always applicable to patient-specific geometries, while the FVS algorithm enables easy stent deployment in anatomically accurate vasculature.

5. CONCLUSION

The present numerical simulation study shows how the choice of a computational approach for deploying the stent influences the CFD analysis, particularly in terms of intra-aneurysmal flow. Variability in intra-aneurysmal haemodynamics existed across the cases, varying according to the parameter analysed and the initial position of the stent considered. The reasons for this variability were discussed and analysed. Average WSS and average velocity were the parameters less affected by the computational approach for deployment and different position of the stent. Nevertheless, in all the analyses carried out, an important reduction of intra-aneurysmal flow was clearly induced in all modelled stent–vessel scenarios.

This work was partially supported within the CENIT programme, as part of CDTEAM and cvREMOD projects funded by the Spanish CDTI and partly within the framework of the @aneurIST Project (IST-2005-027703), which is co-financed by the European Commission within the IST Programme of Sixth Framework Programme.

REFERENCES

Virtual stenting and haemodynamics

A Bernardini et al.

10


