Fast virtual deployment of self-expandable stents: Method and \textit{in vitro} evaluation for intracranial aneurysmal stenting

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\textbf{Abstract}

\textit{Introduction:} Minimally invasive treatment approaches, like the implantation of percutaneous stents, are becoming more popular every day for the treatment of intracranial aneurysms. The outcome of such treatments is related to factors like vessel and aneurysm geometry, hemodynamic conditions and device design. For this reason, a tool for assessing stenting alternatives beforehand is crucial.

\textit{Methodology:} The Fast Virtual Stenting (FVS) method, which provides an estimation of the configuration of intracranial stents when released in realistic geometries, is proposed in this paper. This method is based on constrained simplex deformable models. The constraints are used to account for the stent design. An algorithm for its computational implementation is also proposed. The performance of the proposed methodology was contrasted with real stents released in a silicone phantom.

\textit{Results:} \textit{In vitro} experiments were performed on the phantom where a contrast injection was performed. Subsequently, corresponding Computational Fluid Dynamics (CFD) analyzes were carried out on a digital replica of the phantom with the virtually released stent. Virtual angiographies are used to compare \textit{in vitro} experiments and CFD analysis. Contrast time–density curves for \textit{in vitro} and CFD data were generated and used to compare them.

\textit{Conclusions:} Results of both experiments resemble very well, especially when comparing the contrast density curves. The use of FVS methodology in the clinical environment could provide additional information to clinicians before the treatment to choose the therapy that best fits the patient.

\section*{1. Introduction}

Intracranial aneurysms are pathological dilatations of cerebral vessels whose rupture leads to catastrophic complications such as diffuse bleeding in the brain cavities. The main treatment options for intracranial aneurysms include clipping of the aneurysm or endovascular treatment. Clipping consists of the placement of a metal clip around the neck of the aneurysm so as to isolate the aneurysm from the parent vasculature and to re-establish physiological blood flow. This technique involves craniotomy and a long post-interventional hospitalization demanding expensive health care costs. Regarding endovascular treatments, we can mention coiling and stent implantation. Aneurysm coiling involves the filling of the aneurysm with coils, reducing blood flow into the aneurysmal sac, promoting clotting and isolating the aneurysm from the main blood stream. Intracranial stenting involves the placement of a flexible self-expanding porous tubular mesh made of nitinol or other alloys on the parent vessel across the aneurysm neck. Coiling and stenting are frequently combined to support coil packing inside the aneurysm (Wanke et al., 2003; Vanninen et al., 2003; Felber et al., 2004; Alfke et al., 2004). In the case of aneurysms, stents are widely used to alter the flow reaching the aneurysm redirecting it from the aneurysm neck or dome towards the original blood stream. Additionally, it has been suggested that the use of multiple stents (Crowley et al., 2009) or flow diverters (Wanke and Forsting, 2008; Sadasivan et al., 2009) (a different family of stents consisting of a finer mesh) might be enough to provide sufficient hemodynamic resistance to blood flow through the aneurysmal sac.
aneurysm neck, restituting the original vessel shape and the physiological blood flow along the parent vessel. Bearing this in mind, a tool providing an estimation of the stent configuration after release in a patient specific anatomy would provide useful information to the clinician. The need and practicality of such a tool has been previously outlined by Karmonik et al. (2005). The authors state that the ability to visualize a virtually released stent within the parent artery provides information that is not visible in medical images (e.g., stent attachment to vessel wall and covering of the aneurysm neck). Such tool would allow evaluating in silico different alternatives for the intervention assessing its potential outcome prior to treatment. Supplementary to the purely qualitative aspects of knowing the stent configuration after its release, obtaining the released stent geometry beforehand is a first step towards introducing more complex tools for early therapy assessment. In that way, the combination of anatomically accurate vascular geometries with Computational Fluid Dynamics (CFD) has been extensively used to simulate flow in cerebral vasculature (Cebral and Löhner, 2005; Steinman et al., 2003). Furthermore, CFD has been satisfactorily used to assess the possible outcome of different stenting procedures in patient-specific geometries (Cebral and Löhner, 2005; Appanaboyina et al., 2007; Radaelli et al., 2008). On this regard, Sadasivan et al. (2002, 2009) proposed a methodology for the evaluation of stenting therapy by the evaluation of contrast residence time in the aneurysm which, in combination with CFD, could well be used for early therapy assessment (Ford et al., 2005; Cebral and Löhner, 2005).

1.1. Previous work

A considerable amount of work has been devoted over the past years to develop computational models of vascular stents, their physical behavior and efficiency for treatment. Most of this effort has been dedicated to coronary stents, which are used in the treatment of coronary artery disease (CAD) to maintain the vessel open after angioplasty. Previous work has focused on structural analysis of stent cells (McGarry et al., 2004; Gu et al., 2005; Theriault et al., 2006; Xia et al., 2007), constitutive modelling of stent materials (Petrini et al., 2004; Migliavacca et al., 2005), structural analysis of the interaction between the stent and the vascular wall (Holzapfel et al., 2002; Migliavacca et al., 2004; Timmins et al., 2007), and modelling of the hemodynamic alteration due to the presence of the stent (Deplano et al., 2004; LaDisa et al., 2004; Seo et al., 2005). In such cases, Finite Element Analysis (FEA) is an appropriate methodology for representing the detailed mechanical behavior of the stent material, its design and effect on the vascular wall. In the case of cerebral aneurysm stenting, more detailed description of the interaction between the stent and the vessel wall has been considered in (Bludszuweit-Philipp et al., 2008), where the stent-vessel interaction is approached and integrated with CFD simulation for the prediction of thrombus formation.

All these approaches are designed to model the stent behavior, its interaction with the vessel wall and the local blood flow from a strictly a mechanical point of view. Still, they are not particularly thought for its application on clinical problems where computationally fast techniques that can be applied as part of the daily clinical practice are required. To the best of our knowledge, so far only two approaches have been proposed to model the self-expandable stent release in patient-specific vessels. A first approach has been introduced by Cebral and Löhner (2005) and further studied by Appanaboyina et al. (2007). The stent geometry is mapped as a texture over a cylinder expanded inside the target vessel and the 3D representation of the stent is recovered. However, this method is purely based on the deformation of a cylindrical mesh and its results may be affected by its non-uniform deformation inside the vessel. Another approach has been proposed by Valencia et al. (2007), where the authors suggest to use a simplex deformable model that is initialized as a cylinder and deformed under the effect of internal and external forces to conform to the vessel centerline. Still, this method is based on deforming an already expanded cylinder, not properly taking into account neither the stent geometry nor its compliance to the vessel morphology.

2. Methodology

Deformable simplex models have been previously used by Delingette (1999) in object reconstruction and by Montagnat and Delingette for free-form (Montagnat et al., 1998) and constrained (Montagnat and Delingette, 2005) deformation. The main idea behind this methodology is the use of a second-order partial differential equation for moving a mesh under the effect of internal and external forces. The evolution equation under consideration has the form

$$\rho \frac{\partial^2 \mathbf{p}(t)}{\partial t^2} + \gamma \frac{\partial \mathbf{p}(t)}{\partial t} = \alpha \mathbf{f}_{\text{int}}(\mathbf{p}(t)) + \beta \mathbf{f}_{\text{ext}}(\mathbf{p}(t)).$$

(1)

where $t$ is time, $\mathbf{p} \in \mathcal{S}$ is a point of the simplex mesh, $\rho$ is the mass of the point $\mathbf{p}$, $\gamma$ is a damping parameter (viscous drag), $\mathbf{f}_{\text{int}}(\mathbf{p}(t))$ is the internal surface force, $\mathbf{f}_{\text{ext}}(\mathbf{p}(t))$ is the external surface force and $\alpha$ and $\beta$ are weights controlling the balance between internal/external forces at point $\mathbf{p}$. In order to find a numerical approximation for the solution of Eq. (1) a finite difference discretization is used. The explicit finite difference discretization of Eq. (1) can be written as

$$\mathbf{p}^{i+1} = \mathbf{p}^i + (1 - \gamma)(\mathbf{p}^i - \mathbf{p}^{i-1}) + \alpha \mathbf{f}_{\text{int}}(\mathbf{p}^i) + \beta \mathbf{f}_{\text{ext}}(\mathbf{p}^i).$$

(2)

At the mesh boundary a free boundary condition is used. These models are usually discretized using simplex meshes. In this work we will use 2-simplex meshes. In $\mathbb{R}^3$, 2-simplex meshes

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are surfaces representations which are closely related to triangular meshes. In particular, the underlying graph that define them is dual (Fig. 1a). Complementary definitions and additional information on simplex meshes can be found in the work of Delingette (1999).

On the other hand, simplex meshes are not appropriate for representing stents as these usually do not comply with the definition of 2-simplex mesh. To overcome this limitation, geometrical information of the stent is also taken into account. This can be easily obtained from a micro-CT scan of the stent or directly from the stent manufacturer when possible. Geometrical characteristics of the stent in the “free” state (i.e., expanded outside the vessel) are used to guide the deformation of the mesh. The geometrical characteristics recorded from the free state are set as the reference configuration for the geometrical constraints. Four geometrical constraints are considered:

- Stent design (strut pattern): As most stents have a repeating cell design, the stent is modeled as a set of cells (Fig. 1b and c). By the repetition of stent cells the design of the whole stent is built. This approach allows to map any stent design on the simplex mesh by a simple repetition process.
- Strut length: Total length of a strut between the two ends where it is attached to the stent mesh. This length is measured in the free state configuration.
- Angle between struts: Angle between pairs of struts. This angle is measured in the free state configuration.
- Deployed stent radius: Corresponds to the stent radius in the free state configuration.

The reasons for using these constraints are twofold. First, these parameters are enough to describe the global stent geometry if we are not interested in detailed structural behavior of the stent (where additional information like strut cross-sectional shape, as well as distal and proximal stent designs would be required). Second, this information is relatively easy to obtain for different stents. Such information is stored in a subset of points of the simplex mesh which we recall as stent points.

The proposed method does not ensure nor force that the final configuration of the stent fully conforms to the stent free state configuration, which are imposed as “soft constraints”. In this way, the deformation is stopped when internal and external forces are balanced. The reason for using this simplification is that the stent constraints are measured for the free state when released outside the vessel. When released inside a vessel, the stent does not recover the free state configuration. The use of soft constraints, where a balance between internal and external forces is required, seems more appropriate.

The proposed methodology is validated using two different commercial stents which were released in a silicon phantom that represents a basilar artery bifurcation with a terminal aneurysm. Also, in vitro experiments with these two stents were performed and compared with corresponding CFD models. Using the results from these experiments, the FVS method was evaluated in two ways. Using the FVS method, the same stents were virtually released in the phantom digital model. For the digital model of the phantom with the virtually released stents, CFD simulations and virtual angiograms were produced. Finally, the resulting angiograms (real and virtual) and the contrast residence time in the aneurysm were compared.

2.1. Stent geometrical representation and constraints

As mentioned before, intracranial stents are cylindrical in shape and can be represented using a cylindrical mesh. With this in mind, we consider the mesh as a set of rings, and we define for each ring its center of mass \( c \), which is updated as the mesh moves. We also define the stent mesh using a subset of points \( p_b \in \varphi \), and we consider \( \varphi \) as the set of struts of the stent.

For the deformation of the simplex mesh a new set of shape-constraining forces is proposed. These forces act on the stent points \( p_b \in \varphi \), leading the background mesh to comply with the geometrical constraints. Consequently, we add shape-constraining force terms to Eq. (2) which is rewritten as

\[
\begin{align*}
\mathbf{p}_t^{n+1} = \mathbf{p}_t^n + (1 - \gamma)(\mathbf{p}_t^n - \mathbf{p}_t^{n-1}) \\
&+ \left( \chi f_s(\mathbf{p}_t^n) + \psi f_a(\mathbf{p}_t^n) + f_a(\mathbf{p}_t^n) + f_a(\mathbf{p}_t^n) + f_a(\mathbf{p}_t^n) + f_a(\mathbf{p}_t^n) \right).
\end{align*}
\]

(3)

In this model, parameters \( \chi \) and \( \psi \) are considered equal to 1. Variables \( \chi \) and \( \psi \) are used to balance the influence of \( f_s \) and \( f_a \), respectively, and are parameters which need to be specified for each stent. The internal forces are represented by the terms \( f_s, f_a, f_s, f_a \), and \( f_s, f_a, f_s, f_a \). The force terms \( f_s, f_a, f_s, f_a \), and \( f_s, f_a, f_s, f_a \) are considered to act on the surface of the simplex mesh and only the surface tangential component is taken into account (i.e., they are projected over the surface). In particular, both forces \( f_s, f_a \) act over the points \( p_b \in \varphi \). To simplify the notation, the superscript \( t \) (representing the iteration number) will not be used in the remainder of the text. These terms are described below.

**Strut length force** \( f_s \): The strut length force acts over the stent mesh points shifting them to comply with the corresponding strut length constraint (Fig. 2a). Then, we define

\[
f_s(p_b) = (p_b - p_a)
\]

![Fig. 2](image.png)  
**Fig. 2.** Detail for \( f_s \) and \( f_a \). For \( f_s \), the notation \( a = ||p_{b1} - p_{a1}|| \cos(h) \) and \( b = ||p_{b1} - p_{a1}|| \cos(h) \) was considered for clarity.

being $\mathbf{p}_b = \mathbf{p}_b + \sum_{k=1}^{n_b} \Delta l_b (\mathbf{p}_b - \mathbf{p}_s) / \| \mathbf{p}_b - \mathbf{p}_s \|$

being $A$ the set of neighbors of point $\mathbf{p}_b$ in the background mesh, $| \cdot |$ is the cardinality of the set, and

$\Delta l_b = l_b' - l_b$.

In the former, $l_b'$ is the reference length of the strut bounded by points $\mathbf{p}_b$ and $\mathbf{p}_s$, and $l_b$ is its current length. This force is intended to preserve the strut length observed in the reference configuration at the end of the stent release (Fig. 2a).

**Angle force ($f_a$):** Similarly to the length force, the angle force acts over the stent mesh points shifting them to comply with the corresponding constraint (Fig. 2b). Therefore, we define

$f_a(\mathbf{p}_s) = (\mathbf{p}_s' - \mathbf{p}_s)$.

with

$\mathbf{p}_s' = \mathbf{p}_s + \sum_{k \in A_b} \Delta \theta_b (\mathbf{p}_s - \mathbf{p}_s') / \| \mathbf{p}_s - \mathbf{p}_s' \|$.

being $A_b$ the set of angles $\theta_b$ formed between $\mathbf{p}_s$ and the two neighbors in the stent mesh,

$\mathbf{p}_s' = \frac{\mathbf{p}_s + \mathbf{p}_s''}{2},$

and

$\Delta \theta_b = |b| - |a|,$

where $a = \| \mathbf{p}_s - \mathbf{p}_s' \| \cos(\theta_b)$ and $b = \| \mathbf{p}_s - \mathbf{p}_s'' \| \cos(\theta_b)$. $\theta_b = \theta_b^s p \mathbf{p}_s^s p_0^s$ is the reference angle bounded by points $\mathbf{p}_s$, $\mathbf{p}_s'$ and $\mathbf{p}_s''$ and $\theta_b$ is the current angle. This force is intended to preserve the angles between the struts observed in the reference configuration at the end of the stent release (Fig. 2b).

**Smoothing forces ($f_s$):** The tangential component of the smoothing force (as described by Delingette (1999)).

$f_s(\mathbf{p}_s) = (\mathbf{p}_s' - \mathbf{p}_s)$.

where $\mathbf{p}_s'$ corresponds to the position of $\mathbf{p}_s$ ensuring a smooth distribution of simplex points (Montagnat and Delingette, 2005). The simplex metric parameters $\epsilon_i'$ were set to 1/3 for all simplex points.

**Stent expanding force ($f_{exp}$):** Self-expandable stents expand, under the effect of internal stress, to a preset radius $r_0$ which corresponds to the reference configuration. In order to reproduce this behavior, springs acting along the radial direction of the stent are used. The stent radius $r_0$ is considered as the spring rest position. Then, the magnitude of the expanding force can be expressed as:

$\| f_{exp}(\mathbf{p}_s) \| = k (r_0 - \| \mathbf{p}_s - \mathbf{c}_s \|),$ (5)

where $k$ is the stiffness of the spring. As this force expands the stent, we define its direction along the normal to the simplex mesh at each point ($\mathbf{n}_s$). The stiffness of the spring model $k$ is computed as follows:

$k = A(\mathbf{p}_s) E / r_0$. (6)

where $A(\mathbf{p}_s)$ is the area corresponding to each spring (total mesh area$/|\mathbf{p}_s|$) in the reference configuration and $E$ is the Young’s elasticity modulus of the spring ($E \approx 0.5$ N/m$^2$ was considered).

In order to obtain a realistic stent configuration the stent cross-sectional shape must remain approximately circular. To model this behavior a new variable $\eta_i$ was taken into account. In this way, all the points on the same ring $s$ are maintained at similar distance from the corresponding center of mass $\mathbf{c}_s$. Hence, the mean radius of section $r_i$ is given by

$\bar{r}_s = \sum_{k=1}^{n_b} \frac{| \mathbf{p}_k - \mathbf{c}_s |}{n_b},$ (7)

where $n_b$ is the number of points in each ring (constant for all rings within the same stent). Then, $\eta_i$ will be zero or one according to the following:

$\eta_i = \begin{cases} 1 & \| \mathbf{p}_s - \mathbf{c}_s \| < \bar{r}_s - \| \mathbf{p}_s - \mathbf{c}_s \|, \\ 0 & \| \mathbf{p}_s - \mathbf{c}_s \| > \bar{r}_s. \end{cases}$ (8)

This parameter is entailed to filter the points that are not within a certain threshold (given by parameter $p$) of the average radius $s$. The value $p$ is the percentage threshold limit for $\eta_i$, ranging between 0 and 1 and is a parameter of the model. The stiffer the stent is in the radial direction, the closer to 0 $p$ should be, thereby ensuring a more “circular” shape of the cross-section.

Putting all this together, $f_{exp}$ can be written as:

$f_{exp}(\mathbf{p}_s) = \mathbf{n}_s \eta_i k (r_0 - \| \mathbf{p}_s - \mathbf{c}_s \|)$. (8)

**External forces ($f_{ext}$):** When deployed inside a real vessel, the stent deformation is stopped if the stent reaches the vessel wall, in which case, the vessel is exerting a force on the stent. For this, the point to surface distance is computed from each node to the closest location on the surface. This force is modeled as being equal in magnitude and with opposite direction to $f_{exp}(\mathbf{p}_s)$ at points $\mathbf{p}_s$ which are closer than half the stent radial thickness to the vessel surface. In this way we ensure that the simplex mesh remains inside the vessel when the algorithm stops.

### 2.2. Fast Virtual Stenting algorithm

To virtually release the stent, it is first required to initialize the mesh inside the vessel geometry. For this, the mesh points are created in the form of a cylinder around the vessel centerline. By incorporating the connectivity between these points the simplex mesh is obtained. This initial mesh is used as an approximation of the initial position of the stent inside the catheter before its release. The mesh is initialized in three steps:

1. creation of the simplex points around the vessel centerline,
2. creation of the simplex mesh connectivity using the points from step 1, and
3. creation of the stent mesh.

After the stent is properly located inside the vessel geometry, the iterative algorithm can be started. We will refer to the vascular geometry where the stent is being released as $S_s$ and to the corresponding vessel centerline as $L_s$. The geometrical description of the stent being released will be recalled as $D_s$ and the final stent configuration as $S_s$. In order to quantify the mesh displacement in each iteration of the algorithm, we define:

$\Delta \mathbf{p} = \sum_{t=1}^{N-1} \frac{\| \mathbf{p}_t - \mathbf{p}_{t-1} \|}{N}$. (9)

where $N = | \mathbf{p} |$. In this case, the simplex and stent mesh forces are computed sequentially, and then the mesh position is updated based on both forces (Algorithm 1). The variable $e$, which determines the convergence of the algorithm, was considered as 0.001. The computational implementation of this method was done in ANSI C++. All the stent release experiments mentioned in this work were performed on an Intel® Core™ Duo CPU T7300 2.00 GHz with 2 Gb of memory and no parallelization (e.g., multi-threading) was used.

3. Experimental setup

The proposed methodology was evaluated with two commercial stents. Each stent was scanned in the free configuration with a micro-CT device (Skyscan, Micro Photonics Inc., Allentown, PA, USA) and a second stent was released inside a silicon phantom (Elastat SRL, Geneva, Switzerland). The phantom with the stents was scanned with an angiographic C-arm Allura Xper FD 20 (Philips Healthcare, Best, the Netherlands) medical imaging device as described below.

In vitro angiograms with the phantoms containing each stent were carried out for both cases. These experiments were compared to CFD simulations on computational models of the phantom with the virtually released stents. Virtual angiograms were generated and subsequently compared with the real angiograms. The experimental setup is detailed below.

3.1. Phantom and stent geometries

The silicone phantom presented in Fig. 3a and b was considered. This geometry represents a vessel bifurcation with a terminal aneurysm similar in size and shape to a typical aneurysm occurring at the basilar artery bifurcation (4 mm diameter) into the right and left posterior cerebral arteries (3.5 mm diameter). In this geometry, two different commercially available stents were released. Details on the stent manufacturer, model and sizes appear in Table 1.

Fig. 4 presents the three-dimensional reconstructions obtained from micro-CT for the two stents used in the experiments together with the corresponding FVS model of each stent. The phantom and stents were scanned using a 9 µm resolution micro-CT. The scans lasted approximately 2 h depending on the sample length. Images were further reconstructed using N-Ricon (Photonics Inc., Allentown, PA, USA). A stack of 1000 TIFF images was finally imported in OsiriX (OsiriX v2.7.5, freeware) where a STL file suitable for numerical simulation software could be exported.

The stent cell design used for the virtual release is presented together with each stent. Stent 1 (Fig. 4a) corresponds to an Enterprise stent (Cordis Neurovascular, Miami Lakes, FL, USA) which is manufactured using laser cut technology. The strut length, which was measured from the micro-CT, was considered the same for all the struts (1.2 mm). Three different angle sizes were considered: 80 (red), 100 (blue) and 180 (green) degrees. The measured strut width was ~0.1 mm. The measured stent full length was 20 mm. In the FVS representation of this stent, the cell repeats four times around the stent and five times on its length resulting in a total of 1280 simplex points. For this stent, in both terminal ends the stent has a different cell shape. For this reason, the final cell on each side was not taken into account. The reason for this is that the terminal end of the stent will not have major influence on aneurysmal hemodynamics and the configuration of the stent across the aneurysm neck is not influenced by the stent shape at the extremities. Stent 2 (Fig. 4b) corresponds to a Silk stent (Balt Extrusion, Paris, France) which is manufactured using braiding technology. This particular stent has two different wire sizes. One out of six wires is 50 µm and the other five are 30 µm width. In the virtually released stents, an average width of 34 µm was used for all the wires to preserve an equivalent stent porosity. The measured stent length was 20 mm. For the FVS representation of this stent the cell is repeated 24 times around the stent and 20

Table 1

<table>
<thead>
<tr>
<th>#</th>
<th>Manufacturer</th>
<th>Model (type)</th>
<th>Length</th>
<th>Closed diameter</th>
<th>Open diameter</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Cordis</td>
<td>Enterprise (laser cut)</td>
<td>20 mm</td>
<td>1.6 mm</td>
<td>3.8 mm</td>
</tr>
<tr>
<td>2</td>
<td>Balt</td>
<td>Silk (braided)</td>
<td>20 mm</td>
<td>2.0 mm</td>
<td>5.0 mm</td>
</tr>
</tbody>
</table>

Fig. 3. Left: schematic design with dimensions of the silicone phantom and experimental setup. Right: pictures of the phantom and the experimental setup over the angiographic imaging device.

3.2. In vitro experiment and image acquisition

In order to simulate blood flow inside the phantom after inserting the different stents, this last one was connected to a pulsatile flow pump. The flow pump used is hand-made and consists of a steady flow pump connected to a pulsatile one. This pulsatile pump is formed by a piston imposing a sinusoidal (1 Hz frequency) movement over the steady flow. The fluid circulation was set at 3 cc/s and 37 °C. The phantom was connected to the flow pump in serial and the outlets of the phantom to a fluid reservoir (Fig. 3a). Both stents were released inside the phantom. The fluid properties measured experimentally showed a viscosity of 3.5 cP and density of 1056 kg/m³. After inserting the stents, the phantom was scanned using the angiographic imaging device. A non-ionic iodinated contrast agent (Iopamiro 300, Bracco, Milano, Italy) was injected at the location of the phantom and the feeding system of the pump (20 cm before the bifurcation) for the visualization of the flow in the phantom with the two different stents. The contrast agent injection was coupled with the system at 3 cc/s during 2 s and with 1 s delay. All acquisitions where performed at 15 fps from a frontal view. The field of view (FOV) was set to 15 cm and the table height was 99 cm.

As the contrast agent is injected manually, the exact curve for the contrast density at the inlet is not available. This information was extracted from the images by analyzing the image intensity at the inlet overtime (Fig. 3a). For this, a reference region of interest (ROI) was subtracted from the rest of the sequence thus eliminating the background. The resulting ROI image intensity was averaged to obtain a time–density curve. The same procedure was followed to extract the time–density curve at the aneurysm region. Both curves where normalized to the interval [0, 1] where 0 represents no contrast and 1 represents maximum intensity in that region.

3.3. FVS parameter selection

As described above, variables \( \chi \) and \( \psi \) are used to balance the influence of \( \mathbf{f}_1 \) and \( \mathbf{f}_2 \), respectively. Then, \( \psi = 1 - \chi \) is considered, thus linking both into a single variable. This parameter is related to the “stiffness” of the stent struts (longitudinal and angular). In the case of shape memory alloy (e.g., Nitinol) stents, this property depends on the stent design and radial thickness, requiring a parameter tuning for different stent designs. In the case of braided stents, which are manufactured as a fabric of thin threads, each thread is allowed to slide over crossing threads. Thus, the strut size is not fixed. For this reason, no length or angle constraints are applied allowing the mesh to adapt with more freedom. In all, the parameters required by a FVS stent model remain 3: \( \chi \), \( \alpha \) and \( p \).

In order to find the FVS parameters for each stent, both of them were released freely (i.e., without a vessel geometry) and compared to the geometry obtained from the micro-CFT of the same stent (presented in Fig. 4). This process was used to optimize the model parameters. For Stent 1, the parameters where set to \( \chi = 0.5 \), \( \alpha = 0.1 \) and \( p = 0.5 \). For Stent 2, the parameters where set to \( \chi = 0.8 \), \( \alpha = 0.9 \) and \( p = 0.3 \).

3.4. Computational fluid dynamics and virtual angiography

For the generation of the CFD models, both the vessel and the stent surface meshes where used. As we are interested in modeling the blood flow in the lumen of the vessel containing a stent, the stent geometry is explicitly meshed and its volume (i.e., the stent struts) is represented as a void within the vessel lumen. A non-slip boundary condition is considered over the surface of the stent and on the surface of the vessel (see Fig. 4c). A total of 10 and 13 million unstructured tetrahedral elements were generated in ICEM-CFD (ANSYS Inc., Berkeley, CA, USA) for Stent 1 and Stent 2, respectively. The volumetric elements were transferred to the finite volume-based commercial CFD code, CFX (ANSYS Inc., Berkeley, CA, USA). The Navier–Stokes equations were solved with a second-order backward Euler scheme under the assumption of incompressible, laminar and Newtonian fluid inside a rigid wall boundary. The viscosity and the density of the fluid were matched with the experiments at 3.5 cP and 1056 kg/m³, respectively. A pulsatile velocity waveform of a 1 Hz sinus with an average flow rate of 3 cc/s was imposed at the inlet. The outlets were set to zero pressure (both outlets are at the same distance from the aneurysm). In order to mimic the contrast injection effect on the flow rate for each case, the shape of the contrast density curve extracted from the corresponding image sequence was re-scaled to match the volume of 6 cc and superimposed to the flow at the inlet. The same time–density curve was imposed at the inlet and the scalar transport equation was time resolved.

The resulting time–density curves in the aneurysm were aligned with the corresponding curve obtained from the images by the first non-zero value. As a result, a 3D map of contrast agent concentration in each point of the volumetric mesh was obtained for the two models. The unstructured grid mesh was interpolated to a structured grid of dimensions 151 × 87 × 251 resulting in a voxel spacing of 0.25 mm³. For the creation of digitally reconstructed radiographs (DRR) an in-house software was used.

The computational time for the CFD simulations ranged from 72 to 96 h for Stent 1 and Stent 2, respectively. It is important to consider that a total of 8 s of full CFD simulations with scalar transport

Fig. 4. Iso-surface reconstruction from micro-CT image of the two commercial stents used for the validation. Together with each stent is presented the corresponding stent cell design used for the virtual deployment. On the right is presented the computational mesh used for the CFD simulations highlighting the embedded stent.
for the contrast agent was simulated. For this computations, a computational cluster with 8 nodes, 8 cores each, and 4 Gb memory per node was used to run the simulations using ANSYS-CFX 12.

4. Results

The methodology proposed in this work has been specifically tailored for its use in the clinical environment. In this context, we are interested in a method that can provide an accurate estimation of the configuration of different stents in an anatomically accurate geometry within seconds.

4.1. In vitro vs. virtual release

Fig. 5 presents a comparison of the results obtained from the real stents released in the phantom and the FVS counterparts. Fig. 5a presents the results for Stent 1 (release time was 7 s in 92 iterations) and Fig. 5b the results for Stent 2 (release time was 59 s in 226 iterations). As the radiopacity of these stents is very low and the radiopaque markers are located in the stent tips, is hard to visualize the final shape of the stent in these images. In the first image the stent is highlighted on the image obtained from the angiographic imaging device. The second image corresponds to a close-up of the angiographic image of the phantom. In the third image is presented the phantom digital model with the virtual stent. This image also presents a close-up of the stent in the phantom where the detail of the stent apposition on the wall and the stent geometry across the aneurysm neck can be observed in mode detail. In the last two images is presented the image of the phantom with the real stent with the manually overlayed virtual stent. A similar resemblance between both stents can be observed. The attachment of the stent to the phantom inner wall is more accurate for Stent 1 than for Stent 2. This is observed mainly in the region of the vessel bifurcation and aneurysm.

It can be observed that, for the FVS method, Stent 2 enters slightly more into the aneurysm than Stent 1. This might be due to the fact that Stent 2 is a braided stent, thus not having any constraints on the length of the struts. In this way, the deformable model has more freedom to adapt to the vascular geometry. A similar, but more pronounced, behavior can be observed in the in vitro released stent, where Stent 2 shows the larger differences with Stent 1.

4.2. Contrast fill-in/wash-out

Fig. 6 presents the results from the in vitro flow experiments and CFD simulations. Five time steps from the 8 s simulations, 2 before contrast density peak (fill-in) and 3 after it (wash-out), are presented. The image corresponding to the same instant in time was selected for the in vitro experiment (left) and virtual angiogram (right). The visualization angle in the virtual angiograms was manually adjusted to match the viewpoint from the phantom scans. The plot in each figure presents the contrast density in the inlet (blue) and the contrast density in the aneurysm obtained from the images (pink), and the contrast density in the aneurysm obtained from the CFD simulation (green). Stents are not visible in the virtual angiograms.

In the contrast density curves for Stent 1, contrast density curve of the virtual angiograms resembles the in vitro experiments. The curves match well in the fill-in period differing more during wash-out where the contrast is transported away from the aneurysm faster in the CFD than in the in vitro experiment. During the early fill-in period, a good match in the flow patterns is observed in Fig. 6a, where a jet impinging on the right side of the aneurysm dome can be observed in the experimental data as well as in the CFD simulation. Also, a non-uniform filling of the aneurysm can be observed during fill-in and wash-out. Nevertheless, contrast is mostly homogeneous near the peak contrast density. Regarding the contrast distribution in the parent vessel during fill-in, a slightly higher density can be observed on the left branch (a), showing a good correlation between in vitro and CFD. On the other hand, more contrast is transported on the right branch during wash-out (d).
For Stent 2, a good similarity between in vitro experiments and CFD simulation can also be observed. Similar to Stent 1, both curves match well during fill-in but they separate during washout. Observing the flow during fill-in, a more uniform distribution of contrast is present and no single impinging location can be clearly identified. A non-uniform distribution of contrast is observed during fill-in and wash-out. Also, as can be observed in Fig. 6b the contrast distribution is different between Stent 1 and Stent 2, mainly observable during fill-in period. Finally, considering the downstream contrast distribution, a similar behavior to Stent 1 can be observed where contrast is directed towards the left branch during fill-in (b) and towards the right branch during wash-out (d and e). A good correlation between in vitro and CFD experiments is therefore observed.

5. Discussion

In this work we developed the Fast Virtual Stenting (FVS) method, based on an extension of simplex deformable models with stent-specific geometrical constraints. Preliminary results for this method have been presented (Larrabide et al., 2008) and an extension of the methodology and proper validation is proposed in the current work. There are two main reasons for the use of simplex deformable models. On the one hand, they carry implicit information of local characteristics of the mesh (e.g., surface Gaussian curvature, surface normal and barycentric coordinates of the nodes) facilitating the implementation of a regularized deformation of the mesh. On the other hand, the mesh deformation process requires a low computational cost, it is easy to implement and straightforward to parallelize. As the main purpose of this method is its clinical use, the computational time is an issue to be considered. On this regard, its computational cost depends linearly on the number of simplex nodes and the total number of geometrical constraints. The two stent models used in this work required less than one minute (7 s for Stent 1 and 59 s for Stent 2) for the release. Execution time can vary depending on the complexity of the stent under consideration (more simplex nodes, more constraints, etc.). A computational implementation of this method could be easily adapted to run in multi-thread architectures (multi-core CPUs or GPU hardware) for further speed-up.
Regarding the use of FEA to model intravascular stents, a lot of work has been done so far. However, severe difficulties arise when assessing the behavior of the stent in a real arterial anatomy and idealized geometries are frequently used (McGarry et al., 2004; Gu et al., 2005; Migliavacca et al., 2005; Timmins et al., 2007). One of the main difficulties is fitting the stent model inside real geometries, which are very tortuous presenting high curvature and torsion. In these conditions, achieving a plausible initial configuration of the crimped (i.e., closed inside the catheter) stent in the vessel is, when possible, an ad-hoc and very time consuming process. Another obstacle is that in complex geometries, the associated stent-vasculature wall contact problem is in many cases very unstable or even impossible to solve numerically.

In this work, the method is assessed in a vascular geometry corresponding to a cerebral artery presenting a terminal aneurysm. As it can be observed in Fig. 5, after release, the stent position provided by the method is close to the real one given its simple formulation and low execution time. The configurations of the virtual and real stent are similar regarding the aneurysm neck coverage and its apposition on the vessel wall. These factors, which can be observed from the outcome of the FVS method, can have a large impact on the therapeutic outcome (Ebrahimi et al., 2007; Alfke et al., 2004). Although the location of contrast injection within the parent vessel of the aneurysm was accurately modeled, differences in the contrast distribution can still be observed between the in vivo experiments and CFD models. This could be due to a different contrast distribution at the injection site which could not be considered as it is not available from the images. Regarding the stent effect on flow, the non-constant distribution of contrast during filling and wash-out on both outlet branches is an indicator of the effect of the stent on flow. The aneurysm contrast distribution during filling is clearly different for each stent. Still, the difference between in vitro and CFD for the same stent is small.

The fact that the FVS method can be used in patient-specific vascular geometries has the potential to give the clinicians information that is not otherwise available before treatment (Karmonik et al., 2005). By having a 3D representation of different stent models inside the vessel geometry, the clinician could visualize the process which one provides the best coverage of the aneurysm ostium or release the stent in different positions to assess the sensitivity of the stent position on the vessel relative to its release location and/or rotation. The development of proper visualization tools merging the released stent inside the vascular model in a representation that is natural to the clinician is also important for its usability. As an example, the use of 2D DRR that visualize the stent inside the geometry could be an alternative for enhanced visualization of the treatment prior to intervention.

As shown in this work, the combination of FVS method with CFD could become a powerful analysis tool to assess treatment alternatives preoperatively (Ford et al., 2005; Cebral and Löhrer, 2005). Despite the fact that CFD analysis involves heavy computations, these technologies could revolutionize the way in which elective interventions are planned and optimized (Cebral and Löhrer, 2005; Appanaboyina et al., 2007; Radaelli et al., 2008). In particular, the generation of virtual angiograms from CFD provides additional information in a way that is familiar to clinicians. Following the work of Sadasivan et al. (2002, 2009), CFD derived contrast time–density curves could be used to assess the hemodynamic changes induced by a given device in advance of its placement. For this reason, the use of these techniques within the intervention of acute patients (e.g., suffering a SAH) is currently not possible due to time constraints. In such cases, a proper visualization of virtually deployed stents in the patients own anatomy would provide the clinician with visual feedback for different treatment alternatives (e.g., different stents, different release positions) in a matter of seconds. However, the computational times do allow its use for therapy planning when patients diagnosed with IA are scheduled for treatment with several days or weeks in advance, as is typically done in clinical practice.

6. Conclusions and future work

In this work a novel methodology for virtual stent release is proposed. The FVS method is based on simplex deformable models with constraints. These constraints take into account the stent design (strut disposition), the strut length, the angle between the struts, the strut width and the stent length and diameter. These characteristics of the stent are sufficient to recover the macroscopic shape of the stent if we are not interested in the detailed mechanical behavior of the stent (e.g., internal stress/strain state of the material). From the presented results we can say that a good resemblance can be observed between the real stent and the virtually released counterpart. We can also state that in vitro and CFD experiments agree, specially for the quantitative comparison of the contrast density curves.

For a further evaluation of the FVS method and its limitations, a study comparing it with FEA results would be appropriate. This will allow to better understand the limitations of this model and quantify the errors that can be attributed to its simplifications. The development of proper visualization tools merging the released stent inside the vascular model in a representation that is natural to the clinician is also important for its usability. As an example, the use of 2D DRR that visualize the stent inside the geometry could be an alternative for enhanced visualization of the treatment prior to intervention.

Acknowledgments

This work was partially supported within the CENIT-CDEAM project funded by the Spanish CDTI and partly within the framework of the @neurIST Project (IST-2005-027703), which is co-financed by the European Commission within the IST Program of the Sixth Framework Program.

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