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Realistic virtual intracranial stenting and computational fluid dynamics for treatment analysis

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ABSTRACT

In order to support the decisions of medical experts and to develop better stent designs, the availability of a simulation tool for virtual stenting would be extremely useful. An innovative virtual stenting technique is described in this work, which is directly applicable for complex patient-specific geometries. A basilar tip aneurysm provided for the Virtual Intracranial Stenting Challenge 2010 is considered to demonstrate the advantages of this approach. A free-form deformation is introduced for a wall-tight stent deployment. Numerical flow simulations on sufficiently fine computational meshes are performed for different configurations in order to characterize the inflow rate into the aneurysm and the corresponding residence time in the aneurysm sac. A Neuroform and a SILK stent have been deployed at various locations and the computed residence times have been evaluated and compared, demonstrating the advantage associated with a lower stent porosity. It has been found that the SILK stent leads to a large increase in the residence time and to a significant reduction in the maximum wall shear stress in the aneurysm sac. This is only observed when placing the stent in the appropriate position, showing that virtual stenting might be employed for operation support.

1. Introduction

1.1. The Virtual Intracranial Stenting Challenge

Each year, several groups of engineers associated with medical practitioners all around the world participate in a challenge relying only on computational techniques to assess the effect of stents for the treatment of cerebral aneurysms, see, e.g., Radaelli et al. (2008) or Cavazzuti et al. (2010). The main goal is to improve modeling techniques. Ultimately, the purpose is to provide valuable information to clinicians, so that both hemodynamic and anatomical factors could be considered before planning treatment.

Under the auspices of the LINC and Seventh Intracranial Stenting (ICS10) Meeting, the Fourth Virtual Intracranial Stenting Challenge (VISC10)[1] was held in Houston, TX, September 13–16, 2010. Our research team from the University of Magdeburg "Otto von Guericke" in Germany has contributed to this challenge.

1.2. Medical background

At present, a neurosurgeon must rely only on his/her own experience when deciding how to choose and where to place a stent to obtain an optimal treatment, i.e., flow alteration and consecutive thrombosis of the aneurysmal lumen. Even if this procedure is obviously successful, the support of a simulation tool during this critical process might help for untypical configurations and may serve as a mechanism facilitating and supporting the expert decision. Furthermore, an accurate simulation of blood flow before and after aneurysm treatment is necessary as a step towards the development of better and/or alternative stent designs.

The successful treatment of an aneurysm by an implant should change the hemodynamics in the aneurysm sac, producing thrombogenic conditions, i.e., reducing the flow velocity and elongating the stasis. Numerical flow simulations based on computational fluid dynamics (CFD) may provide all important hemodynamic quantities. A first, qualitative examination can be performed by plotting selected contours of the shear stress, pressure or vector fields of the blood velocity. However, a quantitative analysis is essential in order to accurately quantify the effect of stent deployment and to select between possible alternatives.

For this purpose, the flow stasis can be computed by considering the residence time within the sac, as it has been done by Seshadhri et al. (2011) for idealized and in Kim et al. (2008) for

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[1] http://www.ics-meeting.net/ics_visc10.html

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both idealized and patient-specific geometries. Stent deployment should decrease the blood flow entering the aneurysm by excluding it from the arterial circulatory system. Several previous studies (Burleson and Turitto, 1996; Lanzino et al., 1999; Liepsch, 1986; Wakhloo et al., 1998) have already demonstrated that increasing the aneurysmal flow residence time can produce thrombus formation in cerebral aneurysms. Hence, to stimulate aneurysmal thrombosis, the increase in the stasis in the aneurysm must be targeted (Liou and Liou, 2004). This is the reason why residence time is used as a major indicator of stasis in the present work.

1.3. Patient configuration with a basilar tip aneurysm

The organizers of the VISC10 challenge have provided to all participants the raw data of a basilar tip aneurysm found in a real patient. The considered geometry with the saccular aneurysm and the anatomy of the adjacent vessel tree is depicted in Fig. 1 after an accurate three-dimensional reconstruction by our group. In a further step, the quality of the surface mesh has been improved using an advancing front remeshing algorithm (Schöberl, 1997) in order to facilitate later generation of a high-quality volumetric mesh, while preserving all geometrical details of the vascular system.

The aneurysm develops from the distal part of the basilar artery, the so-called basilar tip. The ostium is relatively wide compared to the dome and comprises the proximal parts (P1-segments) of both posterior cerebral arteries, especially that of the right side. Due to this geometry, an endovascular treatment with coiling alone would be difficult, if not impossible, without occlusion of the right P1-segment including the perforating arteries, that arise from this part of the posterior cerebral artery. Endovascular coiling is safe only after placement of a stent. On the other hand, aneurysm occlusion might also be achieved by deployment of a flow diverter without coiling, as will be considered in what follows.

2. Materials and methods

The main motivation of this study is to investigate in a quantitative and realistic manner the effect of stent deployment in real, patient-specific geometries. In this manner, it should ultimately become possible to prevent the rupture of intracranial aneurysms by supporting in the best possible way the decision of medical experts. At the same time, a numerical tool for virtual stenting would be useful to guide the development and optimization of new implant designs. In this section, all methods needed to solve such a problem are described and used considering the configuration of the VISC10 challenge.

The image segmentation was first performed using the automatic pipeline developed by our group (Moench et al., 2011). The computational geometry has been smoothed while keeping all important features and exported as an STL surface mesh. The stent geometries are reproduced using the CAD software Autodesk Inventor (Autodesk, San Rafael, USA). A SILK (Balt International, Montmorency, France) flow diverter and a Neuroform (Boston Scientific Neurovascular, Fremont, CA) stent were placed in two different positions (AD and AE vessel branching in Fig. 1). The resulting deployment, as described later, comes in direct wall-tight contact with the vessel walls in spite of the resulting, complex deformation. The porosity of the applied stents was evaluated integrating the area when unrolling the stent onto a plane. The Neuroform stent had a high porosity of 84% (i.e., a low strut density), while the SILK stent had a considerably lower porosity of 60% (higher strut density). Further comparisons of these implants can be found for instance in Seshadhri et al. (2011).

2.1. Stent deployment

Virtual stent deployment has been already considered successfully in the nineties (Aenis et al., 1997) for simplified configurations (usually, a straight pipe). The objective of the present paper is to develop a virtual stenting technique applicable for real patient geometries. The goal of virtual stent deployment is to first find a rigid transformation corresponding to stent extension followed by a non-rigid deformation, aligning the deployed stent geometry by following the centerline of the vessel geometry. This deformation leads to a tight fit of the stent so that it is in direct contact with the vessel walls. Prior publications (Appanaboyina et al., 2009; Larrabide et al., 2012) rely on a deformable shape model (see, e.g., McInerney and Terzopoulos, 1993, 1996) used to deform a cylindrical support surface. An alternative approach (Gundert et al., 2011) represents the stent as a solid model and applies constructive solid geometry (CSG) operations for deployment. Augsburger et al. (2011) proposed a different technique based on the modeling of the flow diverter as a porous medium. A recent approach (Bernardini et al., 2011, 2012) simulates deployment based on a finite element model of the stent and the vessel. In contrast to this, the approach proposed here applies a non-rigid registration based on a free-form deformation. According to the authors' knowledge, such a free-form mesh deformation technique has never been applied for virtual stenting yet.

This new approach is conceptually easier, involves less free parameters, and leads to physically plausible results as well. The main objective is to speed up virtual stent deployment while reproducing the true conditions of operation planning. A fast process is important because the expertise of a medical specialist is required, who is typically available only for a short period of time. Therefore, the computational tools should show interactive response, if possible in real-time. Obviously, the goal of a highly efficient virtual deployment is in conflict with the goal of having a simulation as realistic as possible. However, a truly exact description of stent deployment would only be possible when knowing in detail all material properties of the vessel walls. Unfortunately, there is at present no method available to measure these properties for practical cases. Hence, a phenomenological approach appears to be the best possible solution. The presented approach finds in a very short time a configuration, which corresponds to the experience of medical practitioners, guided by years of corresponding interventions.

The deployment proceeds in two steps. First, a rough, large scale deformation makes the stent follow the smoothed centerline curve extracted from the vessel data. Second, this initial deformation is refined iteratively such that the stent geometry gets closer and closer to the real vessel boundaries. Both steps rely on free-form deformations (Sederberg and Parry, 1986) based on trivariate triharmonic splines, similar to the thin-plate splines (Bookstein, 1989; Duchon, 1977) employed in 2D.

For determining the parameters of the deformation the stent model is temporarily replaced by a cylindrical geometry that incorporates the convex hull of the stent before deployment. The resulting deformation is finally applied by replacing the cylindrical template with the real stent model. A similar approach was taken by Appanaboyina et al. (2009), whereas Larrabide et al. (2012) assume a set of rings as a template.

An innovative virtual stent deployment using a free-form deformation is introduced in this work (Supplementary data). Further details of this method are described in Botsch and Kobbelt (2005).

2.2. Computational mesh

The wall-tight stent deployment is exemplified in Fig. 2 for a SILK flow diverter. The SILK stent is composed of a dense strut network with 48 wires (8 wires have a diameter of 50 μm and 40 have a diameter of 30 μm). The obtained virtual stent deformation is illustrated considering both outside and inside views.

All computational meshes needed for the later finite volume simulations have been generated using the commercial tool ANSYS IcemCFD (Ansys Inc., Canonsburg,

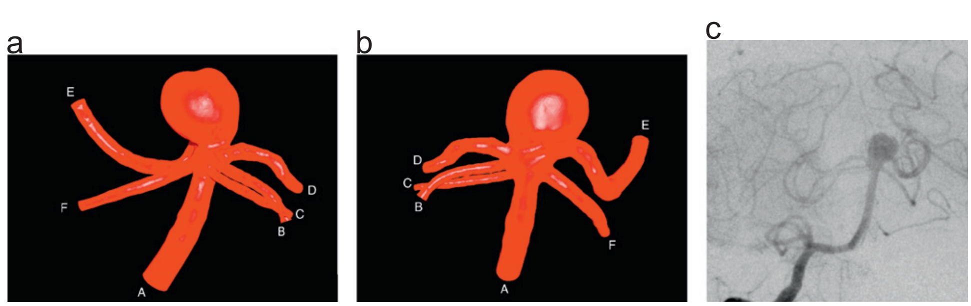

Fig. 1. (a,b) The anatomical model of the computational configuration after reconstruction. (c) Original 2D DSA image of the basilar tip aneurysm provided by the organizers of VISC10.