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Highlights

• A realistic, robust computational framework to support TEVAR planning is provided
• Structural analysis of stent-graft deployment is performed
• A distance-image approach is adopted to build the CFD mesh
• Post-TEVAR hemodynamics is studied by CFD
• Two patient-specific clinical cases are studied in detail
A novel computational framework to predict patient-specific hemodynamics after TEVAR: integration of structural and fluid-dynamics analysis by image elaboration

R.M. Romarowski¹, E. Faggiano¹, M. Conti¹, A. Reali¹,³,*, S. Morganti², and F. Auricchio¹

¹Dept. Civil Engineering and Architecture, University of Pavia, via Ferrata 3, Pavia, Italy
²Dept. Industrial Engineering and Informatics, University of Pavia, via Ferrata 5, Pavia, Italy
³Inst. Advanced Study, Technical University of Munich, Lichtenbergstraße 2, Garching, Germany

*Corresponding author: A. Reali, alessandro.reali@unipv.it

Abstract

Although Thoracic EndoVascular Aortic Repair (TEVAR) is a consolidated procedure to treat thoracic aortic diseases, it still has relevant complications mainly related to sub-optimal wall apposition of the stent-graft, impairing the post-operative hemodynamics and the clinical outcomes. Accurate stent-graft sizing and patient selection are the key aspects to minimize drawbacks. Unfortunately, current TEVAR planning is only based on geometrical measurements performed on static images, completely neglecting the biomechanical interplay between the stent-graft and the aorta. Despite an extensive literature dealing with bioengineering simulation of endovascular implants, studies on the prediction of the post-TEVAR hemodynamics based on both pre-operative patient-specific aortic anatomy and stent-graft mechanical features are still missing.

The present study aims at providing a realistic and robust computational framework to support TEVAR planning in the clinical practice by predicting the post-operative hemodynamics given a selected stent-graft model and pre-operative medical images of the aorta to be treated. A novel approach based on a distance image aimed at transforming the result of the structural analysis of stent-graft deployment in a volume mesh suitable for a computational study of the post-TEVAR hemodynamics is presented. The study discusses two clinical cases as illustrative examples of the framework application and, for one of the two cases, a comparison of the predicted hemodynamics with a simulation based on real post-operative images is shown.

Such a comparison proves that the proposed computational framework is able to capture the main hemodynamic aspects related to the stent-graft implant. In particular, the use of simulations has confirmed the unsuitability for the endovascular repair of one of the two
patients due to the short proximal landing zone, leading to a high risk of the so-called bird-beak.

The proposed computational framework is shown to be a useful tool to support planning of elective TEVAR, especially in those borderline cases when the sole geometrical analysis of static images is not exhaustive.

**Keywords:** thoracic endovascular aortic repair, virtual surgery, computational fluid dynamics, stent-graft simulation, structural finite element analysis.

## 1 Introduction

Thoracic Endovascular Aortic Repair (TEVAR) is a consolidated procedure to treat thoracic aortic diseases such as aneurysms and dissections, especially in those patients who are unsuitable for standard open surgery [1, 2]. The procedure is performed through a catheter-guided deployment of one or more stent-grafts, which are metallic tubular structures covered by a polymeric skirt. Despite being a low-risk treatment, TEVAR has also some not negligible drawbacks associated with both in-hospital mortality and neurological complications [3]. Most of such complications are related to suboptimal apposition of the device to the aortic wall [4] impairing the post-operative hemodynamics resulting in endoleaks, stent-graft migration, development of a false aneurysm, and malperfusion of peripheral organs [5].

Consequently, the clinical outcome of this minimally invasive technique is strictly related to appropriate patient/stent-graft selection, stent-graft/aorta mechanical interaction, and operator skills. In this context, a quantitative assessment of the biomechanical actions induced in the aortic anatomy due to the stent-graft and vice versa may support a correct planning of the procedure, improving its clinical outcomes. Unfortunately, planning is mainly based on geometrical measurements performed on static images [1], neglecting the aforementioned biomechanical aspects. For this reason, in the last decade, the bioengineering community has promoted the use of numerical simulations to assess in a non-invasive manner the aortic biomechanics and, in particular, the aortic hemodynamics before and after TEVAR [6].

In 2006, Frauenfelder et al. [7] analyzed the hemodynamic changes in abdominal aortic aneurysms (AAA) after stent-graft placement by computational fluid dynamics (CFD) in 11 patients. Both pre-operative and post-operative fluid-dynamic simulations were based on multidetector computed tomography (CT) angiography datasets. Similarly, in 2010, Karmonik et al. [8] used CFD to provide a quantitative assessment of hemodynamic wall forces in Type B Aortic Dissection before and after treatment using pre- and post-operative CT scans and dynamic magnetic resonance imaging (MRI) data. In 2012, Midulla et al. [9] further combined aortic anatomy derived from CT with MRI, used to obtain boundary conditions of the aortic wall movement during the cardiac cycle and, performed CFD analysis to evaluate post-TEVAR aortic blood flow in 20 patients. The mentioned studies are just few examples of the extensive and constant effort to bring aortic fluid-dynamics simulations to the bedside [10, 11].

Although such studies accurately describe the post-operative hemodynamics since based on real post-operative images and measurements, they lack the predictive capability of the simulations, i.e., the quantification of the aortic hemodynamics after the intervention based on pre-operative images and stent-graft design characteristics. To this aim, two major ingredients are necessary:
1) an accurate modeling of both the prosthesis mechanics and its deployment mechanism; and 2) a reliable modeling of the aortic lumen after stent-graft apposition.

While a lot of progresses have been made in modeling the stent-graft mechanics and its deployment in (patient-specific) aortic models by structural finite element analysis [12, 13, 14, 15, 16], integration of the stent-graft model after the deployment with the aortic fluid domain is still an open issue to predict fluid-dynamics after TEVAR. As an example, Filipovic et al. [17] examined blood velocity and wall shear stresses in the thoracic aorta with and without an aneurysm, based in a single-patient case and a post-surgery virtual configuration (obtained by resecting the aneurysm and bridging the free boundaries of the aorta with a cylindrical geometry representing the implanted prosthesis). Analogously, Alimohammadi et al. [18] analyzed the hemodynamic effectiveness of aortic dissection treatments via virtual stenting by simulating the stented domain by simply sealing the entry tear and removing the intimal flap.

Other studies by Xiong et al. [19] and Neugebauer et al. [20], proposed to create the virtual post-operative model manipulating the pre-operative surface of the aortic wall by elaboration of the centerline of the vessel based on design parameters of the prosthesis. Although such an approach creates a more accurate representation of the surface when compared to previous studies [17, 18], it has a limited capability of capturing undesired protrusions of the stent-graft into the lumen, as in the case of the bird-beak [4, 21]. Furthermore, comparison of virtual surgery with actual post-operative data is missing in both cases. The combination of virtual stent-graft deployment and CFD analysis has been addressed in 2009 by the group of Figueroa, who compared hemodynamics in the aorta with a virtually implanted stent-graft and the actual post-operative geometry [22]; similarly, the displacement forces acting on the stent-graft have been investigated using follow-up imaging after endovascular repair AAA [23]. Unfortunately, in both studies, the details on how the stent-graft is modeled are not reported. Subsequently, in 2013, Prasad et al. [24] proposed a computational framework for investigating the positional stability of aortic stent-grafts combining Computational Solid Mechanics and Computational Fluid Dynamics. The study focused on abdominal aneurysms and investigates in a parametric manner, several factors that are clinically known to affect stent-graft stability. Although this study is certainly one of the most comprehensive analysis performed in the field of virtual endografting, it considers an idealized model of an aortic abdominal aneurysm and neglects the modeling of the actual stent-graft deployment.

Given such premises, the present study aims at providing a realistic and robust computational framework to support TEVAR planning in clinical practice by predicting post-operative hemodynamics, given a selected stent-graft model to be implanted and the target pre-operative aortic anatomy. In particular, the present work describes the whole computational framework to simulate TEVAR, proposing a novel technique based on distance images for the integration of the stent-graft and the aortic surfaces to generate a CFD-suitable volumetric mesh, following the simulation of stent-graft deployment.

The study focuses on thoracic aortic aneurysms, presenting two clinical cases as illustrative examples of the framework application with particular emphasis on the analysis of borderline cases, when pre-operative planning is not straightforward. The simulation work-flow consists of three main steps: patient-specific simulation of the stent-graft deployment by structural FEA [15], creation of CFD-suitable domain with FEA outcomes based on signed distance functions [25], and CFD analysis to compute post-TEVAR hemodynamics [26]. Moreover, for one of the two clinical cases, a comparison of the predicted hemodynamics with the actual one, computed using
real post-operative images, is presented.

2 Materials and Methods

In this section we provide methodological details of the developed computational framework consisting of three main steps as shown in Figure 1: step 1) realistic simulation of stent-graft deployment; step 2) creation of CFD analysis-suitable domain; step 3) CFD simulations.

2.1 STEP 1: Realistic simulation of stent-graft deployment

The first step consists of the patient-specific structural analysis of the TEVAR implant, which combines all the different sub-steps depicted in Figure 1-step 1: a) medical image processing; b) stent-graft modeling; and c) simulation of stent-graft deployment.

2.1.1 Clinical cases and medical image processing

As illustrative examples of the clinical use of the proposed framework, we refer in the following to two patients, who were admitted to IRCCS Policlinico San Donato (San Donato Milanese, Milan, Italy) with frequent episodes of chest pain.

In both cases, contrast enhanced multislice computed tomography (MSCT) was acquired on a Siemens SOMATOM Definition AS scanner (Siemens Medical Solutions, Erlangen, Germany), with a slice thickness of 1mm, a reconstruction matrix of 512 × 512 pixels, and a final resolution of 0.8mm × 0.8mm × 1mm.

Patient 1 (P1) presented a saccular aneurysm having a maximum diameter of 55 mm just below the aortic arch, whereas Patient 2 (P2) presented an aneurysm with a maximum diameter of 40mm in the same position. The lengths of the proximal landing zone were 24mm and 18mm, and the arches inner radii 37mm and 19mm for P1 and P2, respectively.

TEVAR was performed on P1 with the deployment of one stent-graft Medtronic Valiant 28-24-150 (Medtronic, Santa Rosa, CA, USA), consisting of 8 Nitinol rings covering a polyester skirt to limit blood flow into the sac plus a proximal uncovered ring to improve stability without occluding the left common carotid artery [27]. A post-operative MSCT confirmed the correct placement of the prosthesis and the absence of either migration or endoleak. Such a post-operative scan was performed with a slice thickness of 1mm, a reconstruction matrix of 512 × 512 pixels, and a final resolution of 0.84mm × 0.84mm × 1mm. Instead, due to the complicated morphology of P2’s arch, characterized by a short proximal healthy neck and a high arch angulation, decision to undergo TEVAR was not taken on P2, considering this as a borderline patient not ideally suited for TEVAR. Need of patients’ consent for using their images was waived due to the retrospective nature of the study and the use of anonymized data.

Using the Vascular Modeling Toolkit library (VMTK, www.vmtk.org) [28], aortas were segmented from the annulus to the diaphragm together with the root of the supraortic branches; in particular, the segmentation of three MSCT scans was performed: 1) pre-operative scan of P1; 2) pre-operative scan of P2; 3) post-operative scan of P1. The first two segmentations were used as starting point of the simulation framework for virtual TEVAR, while the latter was used as
a reference in the comparative analysis between the virtual prediction and the actual surgical outcome. First, a level-set segmentation was applied to extract the region where the contrast dye was released according to the approach proposed in [28] using the following parameters: initialization of colliding fronts, 100 iterations, propagation scaling of 0.3, curvature scaling of 1.0, and advection scaling of 1.0. Once the segmentation was obtained, it was converted into a triangulated surface representation using the Marching Cubes method [29]. Then a Taubin surface smoothing [30] with 30 iterations and passband of 0.01 was applied to both pre- and post-operative surfaces. Finally, inflow and outflow sections were created by interactively cutting the original surface by planes perpendicular to the lumen longitudinal axis.

2.1.2 Stent-graft and aortic wall modeling

Following the procedure described in [15] stent-graft models were created to define meshes suitable for structural FEA using Abaqus/Explicit v. 6.16 (Simulia, Dassault Systèmes, Providence, RI, USA).

For P1, the stent-graft model imitates the implanted device, i.e., Medtronic Valiant 28-24-150. The mesh of the model consisted of 81,449 nodes connected by two sets of elements: one composed by 22,937 linear brick elements with reduced integration (C3D8R) representing the struts and one of 76,942 triangular membrane elements (M3D3) representing the fabric coverage. In the case of P2, who did not undergo surgery, a Medtronic Valiant 26-26-100 stent-graft was proposed, sized by the analysis of the diameter of the proximal neck and aneurysm extension following the indications discussed with clinical operators. The mesh of the stent-graft model in such a case consisted of 66,698 nodes, 19,759 brick elements, and 60,361 membrane elements. Nitinol of the stent-graft struts was modeled as a pseudoelastic material whereas the stent-graft fabric, made of woven polyester, was assumed to be linear elastic as proposed earlier in literature [31].

Following a preliminary analysis (see Appendix - section 7.1) aimed at evaluating the impact of a rigid-wall hypothesis on the final stent-graft configuration after the deployment, we considered the aorta as a rigid body described by a rigid surface resembling the pre-operative lumen. This approach has already proven to give reliable and accurate results [15]. Moreover, in order to reduce the computational cost of the simulation, we performed a surface remeshing, using the corresponding VMTK module, to minimize the number of elements that describe the surface without losing geometrical accuracy; the remeshed triangular surface was then imported in Abaqus. The final aortic model consisted of a mesh of 6,926 nodes connected by 13,698 triangular elements (type R3D3) and 3,470 nodes connected by 6,396 elements for patient P1 and P2, respectively.

2.1.3 Virtual stent-graft deployment

The input file for the structural FEA was generated by means of an in-house Python script allowing the user to select the given stent-graft model, from a virtual library of designs, and the proximal landing point within the diseased vessel. The selection of the landing point in case of P1 was based on the actual post-operative images, while in the case of P2 the choice was based on surgeon’s advice by visual inspection of the pre-operative surface reconstruction.

The numerical analysis of the stent-graft deployment in the aorta is a non-linear problem involving large deformations and contact, which was solved with the commercial software Abaqus/Explicit.
To avoid spurious inertial effects, all the simulations were run under a quasi-static regime: the ratio of kinetic energy (ALLKE) to internal energy (ALLIE) was monitored along the whole simulation to be below the threshold of 10% following the recommendations of the software’s manual. The steps of the deployment simulation for the case of patient P1 were depicted in Figure 2: following the approach proposed by Auricchio et al. [15], the stent-graft was first crimped by a catheter and curved from a straight position to the vessel centerline; in this step, the contact interaction between the catheter and the struts of the prosthesis was considered frictionless. Once the stent was in place, a uniform enlargement of the catheter surface along its length allowed the stent-graft re-expansion to simulate its deployment. To do so, a contact pair between the stent-graft struts and the luminal surface of the artery was activated.

Approximate position of Figure 2.

To assess the quality of the virtually created geometry, the segments of the aorta where the stent-graft was placed were compared between the real post-operative reconstruction and the virtual one for the case of patient P1. Both surfaces were clipped and the prosthesis surface was then registered with the iterative closest point algorithm [32]. Once the geometries were superposed, point-wise distances were calculated between each of the points of the surface and a colormap was created to illustrate the differences using the corresponding VMTK modules. The post-deployment configuration of the stent-graft fabric, resulting from the numerical analysis, was then exported in stereolithographic format as an input for the second step of the framework described in the following.

2.2 STEP 2: creation of CFD analysis-suitable domain

CFD analyses require the definition of a suitable mesh describing the fluid-domain. Meshing vessel-like geometries is often a cumbersome operation, especially if the shape of the artery cannot be straightforwardly associated to a conforming, cylinder-like geometry, as it happens in our cases and in most clinical cases of TEVAR [33]. In fact, the fluid-domain in such situations is bounded partially by the native aortic lumen, partially by the stent-graft, and potentially by both of them in case of prosthesis malapposition. Moreover, boundaries are obviously not known a priori in the case of the predictive simulation of a stent-graft implant; for this reason, the structural analysis of the stent-graft deployment, described in the previous section, was used as input of this second step of the framework (see Figure 1, step 2).

2.2.1 Cylinder mapping on the stent-graft

As a preliminary stage of the construction of a CFD analysis-suitable domain from the structural apposition results, we worked with classical CAD functions like Non-Uniform Rational B-Splines (NURBS) and performed a mapping procedure in order to obtain a smooth description of the implanted device, much easier to manipulate in subsequent steps, but potentially very detailed. Mathematical details of this mapping are reported herein due to the paramount importance of this step for obtaining a conforming surface, which is one of the novelties of our work. A generic surface can be described adopting a NURBS representation as follows:
\[ J(\xi, \eta) = \sum_{i=1}^{n} \sum_{j=1}^{m} N_{i,p}(\xi) M_{j,q}(\eta) B_{i,j} W_{ij} \]

(1)

where \( N_{i,p}(\xi) \) and \( M_{j,q}(\eta) \) are B-spline basis functions of degree \( p \) and \( q \) respectively, while \( n \) and \( m \) are the number of control points in the parametric directions, and \( W_{ij} \) are proper weights [34].

Following [35] the idea was to start from a simple primitive NURBS geometry resembling the shape to be mapped. In our case, a cylinder-like shape was swept along the centerline of the implanted device (see Figure 3, A). Then, keeping fixed the number of control points per direction \( n \) and \( m \), the B-spline basis functions \( N_{i,p}(\xi) \) and \( M_{j,q}(\eta) \) and the weights of the primitive geometry, the mapping procedure consists in finding the optimal position of the control points \( B_{i,j} \) such that the distance between the target surface and the mapped one are minimized, in a least square sense. The target surface (i.e., the implanted prosthesis) was evaluated at a set of \( n_s \) sampling points (with \( n_s > n \cdot m \)) as depicted in Figure 3, B. Increasing the number of control points \( (n \) and \( m) \) of the primitive geometry through simple refinement procedures [34] allowed a more accurate representation of the implanted device geometry. Then, the cylindrical surface representing the stent-graft was extruded radially, following the inner normal of the surface: a constant value of 0.4 mm, resembling the strut diameter. Finally, a thin volumetric annular stent model was generated (see Figure 3, C and D).

2.2.2 CFD mesh generation

The creation of a single conforming surface mesh representing the volume where the blood flows after the stent-graft deployment is one of the main innovations of the present study. We used the concept of signed distance function [25]. Given a surface \( S : \mathbb{R}^2 \to \mathbb{R}^3 \) a signed distance function to the surface \( S \) is defined as the scalar function \( f_S(x) : \mathbb{R}^3 \to \mathbb{R} \) representing the distance from \( x \) to the nearest point on the surface \( S \), assuming negative values for \( x \) inside the space bounded by the surface and positive outside. On the surface, the signed distance function assumes zero values. From the signed distance function is always possible to reconstruct the original surface extracting its zero values, i.e., \( S = \{ x \in \mathbb{R}^3 : f_S(x) = 0 \} \).

We call distance image a discretization of the signed distance function over a set of voxels, i.e., each voxel of the distance image assumes the value of the signed distance function in its location. Our idea was to construct two distance images with the same coincident voxels, representing the annular stent and the vessel, and to appropriately merge them in order to obtain a combined distance image from which the final surface can be extracted.

First, two “empty” 3D images were created with spatial boundaries that match the vessel bounding box. These two images were discretized with a voxel size of 0.4mm × 0.4mm × 0.4mm in order to have enough spatial resolution to capture the thickness of the annular stent. Each element of this image has its own spatial coordinates as well as an empty value where the distances will be stored.
Secondly, by iterating through all voxels, each distance with the most proximal point in the surfaces (either the annular stent or the vessel) were stored. This step was performed with the VTK Image Processing library [36]. For the aorta, the distance image assumed the standard convention with negative values inside the aorta and positive outside as seen in Figure 4, A. In the case of the annular stent, the standard sign convention was inverted and the distance image yielded positive values inside the thickness of the stent and negative outside (see Figure 4, B).

Thirdly, the two images were combined using the maximum operator in order to obtain a final image with negative values inside the lumen and positive values both outside the lumen and inside the volume of the stent (see Figure 4, C).

As can be seen in Figure 4, C, the negative values of the combined distance image represent the union of vessel and stent but are not the area we are searching for. Indeed, also the left subclavian artery, which is closed by the implant, and the aneurysm sac, which is excluded by the implant, assume negative values although they should be excluded from the final surface.

The final surfaces were then constructed extracting the negative values by means of a region-growing method with a starting seed inside the lumen within the newly-constructed combined distance image [37]. With this technique, the front proceeds as long as it encounters a zero value, which stops it; in this way, only the regions in which the blood can flow (starting from an inside seeding) were included in the final surface geometry (see Figure 4, D).

Approximate position of Figure 4.

2.3 STEP 3: CFD simulations

Three CFD simulations were performed, one for each of the following cases: 1) virtual post-TEVAR analysis of P1; 2) actual post-TEVAR analysis of P1; 3) virtual post-TEVAR analysis of P2. For the sake of simplicity, the simulations are labeled in the following as P1-virtual, P1-actual, and P2-virtual, respectively.

According to He et al. [38], flow extensions were connected at the endpoint of each input/output branch of the aorta and, for each case, the triangular discrete surface describing the aortic lumen was turned into volumetric mesh of linear tetrahedral elements using VMTK [28]. Mesh size varied case-by-case, ranging from 1.2M to 7M elements (see Table 1), in order to ensure the convergence solution at the systolic peak, capturing the main local features of the velocity field with local refinement when necessary.

The unsteady Navier-Stokes equations for incompressible fluids were considered; blood was modeled as a Newtonian, homogeneous, and incompressible fluid with a viscosity of 0.0035 Pa·s and a density of 1060 kg/m³.

The vessel was considered rigid imposing a corresponding no-slip condition at lumen wall. The same inlet/outlet boundary conditions are used in all the three simulations; such conditions were derived from post-operative medical images of P1. In particular, 2D CINE phase contrast magnetic resonance images (PC-MRI) were acquired from P1 after TEVAR using a Siemens MAGNETOM Aera scanner (Siemens Medical Solutions, Erlangen, Germany). Five oblique image slices were positioned in the mid-ascending aorta, the descending aorta at the height of the diaphragm, and in the proximal sections of the three supraortic vessels. The temporal resolution was 30 samples/beat with a pixel resolution of 2.08mm × 2.08mm. Venc was chosen equal to 150 cm/s;
TR/TE: 37.1/2.5ms. The flow rate from PC-MRI was imposed as boundary condition in the ascending aorta with a peak flow of 423 ml/s and a stroke volume of 83 ml assuming a flat velocity profile. Outflow boundaries consisted of three element Windkessel circuits also tuned in a patient specific way with a procedure resembling [39] which also accounted for patient-specific cuff pressure.

Time step was chosen equal to 0.75ms and 6 cycles were simulated to ensure converge of both velocity and pressure field. Flow was assumed to be laminar but a numerical viscosity was added due to the high Reynolds number mostly present in the ascending aorta. The average Reynolds numbers were 1554, 1347, 1670 for P1-virtual, P1-actual, and P2-virtual, respectively. The Womersley number was 25 in all three cases. To perform the simulations, we uses the open-source finite element C++ library LifeV (www.lifev.org) [40]; as a trade-off between accuracy and computational cost, we used mixed P1bubble-P1 elements, providing a piecewise continuous linear interpolation enriched by a cubic bubble for the velocity, and a piecewise continuous linear approximation for the pressure. The results of CFD simulations were post-processed using Paraview v. 4.4 (Kitware, FR).

3 Results

3.1 STEP 1: Realistic simulation of stent-graft deployment

The simulation of stent-graft deployment was performed successfully in both of the cases under investigation in an average time of 6.5 hours using a multicore Intel Xeon E5-4620-2.20 GHz processor.

As shown in Figure 5, A, the simulation of stent-graft deployment in the pre-operative model of the aorta for P1 was able to capture the main aspects of the actual post-operative configuration. In fact, the simulation showed that the device membrane totally covers the left subclavian artery ostium, which is indeed excluded by the circulation, calling for a surgical bypass as part of the clinical procedure. At the same time, the simulation was also able to predict the actual position of the proximal uncovered ring of the stent-graft which allowed to keep the patency of the left carotid artery, while increasing the stability of the implant in the landing zone. Moreover, in the inner curvature of the arch, it is possible to visualize that the numerical results appropriately reflect the localized kinking of the stent-graft (Figure 5, B); such a stent-graft kink is also present in the distal zone of the implant.

Approximate position of Figure 5.

Results of stent-graft deployment for P2, similarly to what was observed for P1, showed that the left subclavian artery was completely covered by the stent-graft fabric while the left carotid artery remained patent. In such a case, the simulation revealed a non optimal apposition of the proximal part of the prosthesis to the aortic wall which is particularly remarkable in the inner curvature of the arch, immediately at the proximal neck of the aneurysm sac (please refer to the Appendix - section 7.2 for further details).
3.2 STEP 2: creation of CFD analysis-suitable domain

The simplification of the stent was straightforward and there was agreement between the shape and coverage of the virtually deployed graft and the reconstructed cylinder as it can be appreciated in Figure 3, (D). The quantitative comparison of the virtual post-operative lumen shape (P1-virtual) with the actual one (P1-actual), is reported in Figure 6 in terms of point-wise distances between the surfaces. The magnitude of such a distance had an average of 1.31mm; the highest discrepancy (almost 5mm) was located near the distal sealing zone, in the posterior side, where the aorta tapers significantly.

As previously discussed, the structural simulation has shown that P2 is an example of stent-graft malapposition suggesting a high risk of bird-beak. Indeed, this case presented a device protrusion inside the lumen that our approach was able to capture, reconstructing at the same time a conforming surface ready for fluid simulations (see Figure 4, C and D).

3.3 STEP 3: CFD simulations

All simulations were run on a multi-core AMD Opteron 6272-2.1GHz processor using 64 cores; simulation time to reach systolic peak varied approximately from 6 hours (P1-virtual and P1-actual) up to 60 hours (P2-virtual).

Figure 7 depicts the magnitude of the flow velocity and the pressure distribution along the aorta of P1 at the systolic peak (20% of the cardiac cycle) for both P1-virtual and P1-actual analyses. Systolic peak was considered to be the more relevant for our purposes; please refer to the Appendix (section 7.3) for the comparison of pressure and velocity fields in different instants of the cardiac cycle of P1.

Results in both cases showed that velocity was higher in the supra-aortic branches, in the inner portion of the arch, and in the distal part of the aorta covered by the stent-graft. The magnitude of the velocity in these regions was higher in the virtual model, having a larger extension as well, when compared to the actual one. The maximum value computed for P1-virtual in the proximal landing zone of the endograft is 168 cm/s whereas P1-actual indicated a value of 146 cm/s; regarding the distal landing zone, P1-virtual indicated 185 cm/s as maximum magnitude of velocity, while in the case of P1-actual the computed value was 161 cm/s.

Systolic pressure along the aorta ranged from 99 to 122 mmHg in the virtual case and from 100 to 120 mmHg in the actual one; a gradient is observed from the ascending to the descending aorta with sound values in both cases.

A further comparison of the computed blood flow profile between the virtual and the actual model is reported in Figure 8, where the contour plots of the flow velocity magnitude at systolic peak in the proximal, middle, and distal cross-sectional planes perpendicular to the stent-graft rings are depicted.

In the proximal slice, velocity profiles are in both cases skewed towards the inner curvature of the arch. Due to the different shape of the predicted and the post-operative geometries, velocity peaks were on the inferior-right and inferior-left parts of the aorta, respectively.
At the middle plane, flow was mostly flat across the slice with an acceleration zone at the outer curvature and a small deceleration zone at the inner curvature; the latter effect was more remarkable in the CFD results of P1-actual due to the uneven profile of the lumen cross-section.

Finally, the distal slice matched the recirculation zone of the distal ring and lower velocity magnitudes were observed towards the inner curvature while high velocity zones surrounded the posterior part of the vessel. Obviously, such a comparative analysis is limited to patient P1 because patient P2 did not undergo TEVAR.

Approximate position of Figure 8.

In order to compare the hemodynamic conditions of the two clinical cases, streamlines of the velocity in the region of interest are shown in Figure 9. In the ascending and descending parts of both aortas, streamlines followed an organized pattern suggesting that the presence of the stent-graft did not disturb the flow neither proximally nor distally to the implant. In both cases, the left-subclavian artery was excluded from the analysis because entirely covered by the stent-graft membrane, while the brachiocephalic trunk and the left carotid artery were characterized by high velocity flow. Flow velocity was significantly higher in P2 along all the aortic region covered by the stent-graft, whereas P1 had higher velocity areas only in the proximal and distal curves of the prosthesis. Flow was fairly organized inside P1 stent-graft whereas the bulging in P2 created recirculation at the inner curvature. Interestingly, vortical structures were seen on the distal landing zone of the stent-graft, with a higher intensity in P2. Following the streamlines, this secondary flow seemed to be originated by the portion of blood impinging the outer wall of the vessel. It is worth noting that the stent-graft protrusion present in P2 did not lead to a significant flow disturbance near the inner curvature of the arch.

Approximate position of Figure 9.

4 Discussion

In the present study, we introduced a computational framework to support the planning of TEVAR for thoracic aortic aneurysms; such a simulation tool computes the post-operative hemodynamics relying on pre-operative images, accounting also for the simulation of the stent-graft deployment. In particular, two clinical cases were discussed emphasizing the flexibility of the proposed tool to tackle real patient-specific scenarios.

The framework integrated three main steps: 1) simulation of stent-graft deployment by structural FEA; 2) generation of CFD-suitable mesh of the post-operative aortic configuration after the stent-graft deployment; 3) CFD analysis to assess the post-TEVAR hemodynamics.

The potential use of structural FEA as a preoperative planning tool of endovascular repair has been proven by several publications, which validate the simulation outcomes comparing the numerical results with in-vitro experiments [16] or with the actual geometry of the deployed stent-graft(s) retrieved from post-operative images. Most of these studies deal with the ascending aorta [15] and abdominal aorta [14], while the descending thoracic aorta and the arch are not extensively investigated [13]. Our study fills this gap further confirming the predictive capability of this type of simulations also in the descending thoracic aorta, discussing two clinical cases of
saccular aneurysms. The comparison of the stent-graft configuration after the virtual deployment matches well the actual shape of the prosthesis implanted in the real clinical case. In particular, simulations were able to highlight the risk of bird-beak in P2, which was considered a borderline patient for endovascular repair due to the short proximal aneurysm neck, limiting the landing zone, and the acute arch angulation, challenging the stent-graft conformability.

We have also presented approach based on distance images merging the vessel and stent-graft geometries to create CFD-suitable meshes. Such an approach constituted the main novelty of the present study since it allows to overcome the limitations of other simplistic approaches proposed in literature, where a direct modification of the aortic surface is performed to simulate endografting [17, 19, 20]. Indeed, we achieved a description of the aorta/endograft configuration which would be difficult to obtain using simple boolean operations such as union, intersection or difference between surfaces. This feature is particularly important when dealing with complications like bird-beak or occluded arterial branches. To reach our proposed goal, we transformed the final stent-graft geometry into an annular cylinder-like structure, able, if desired, to follow even wrinkles and geometric details of the implanted device. This created a much smoother geometry compatible with a fitted mesh.

The quantitative comparison of the virtual post-operative lumen and the actual one retrieved from medical imaging presented differences of up to 5mm in very limited zones of the endograft, which may be caused by the displacement of the aorta during the surgical procedure making a perfect registration of the surfaces difficult. However, the overall mean value of the absolute difference was around 1 mm which is very close to the spatial resolution of the MSCT scan. Flow patterns resulting from the predictive analysis were in close agreement with the post-operative simulation in P1. The velocity field in the whole aortic geometry in the systolic phase was qualitatively similar, but velocity was higher in the virtually reconstructed domain; such a difference may be caused by the artifact in the post-operative MSCT created by the stent-graft, which may lead to an overestimation of the lumen diameter. This agreement was also confirmed when cross-sectional flow profiles within the stent-graft body were analyzed. Combining the geometries with our strategy can create a virtual surgery domain in which details such as endoleak and bird-beak configuration, appearing mainly in angulated arches, can be represented in the final mesh (see for example Figure 4, D). Flow patterns in the virtual vessel show differences between the patients, even though they have similar morphologies. Recirculation areas observed in P2 within the body of the stent-graft might be due to the arch angulation. However, such a pattern has not yet proved to be harmful on a long-term basis. Similarly, although the kinking of the bird beak in the proximal fixation point of the stent-graft is accurately captured by our model, particular flow disturbances were not observed. Conversely, low velocity is seen in the distal fixation zone which can be explained by the diameter mismatch between the tapered aorta and the uniform endoprosthesis. This could be a hint on why stent-grafts often migrate proximally rather than distally [41], even though it is not proven yet. These results underline the value of using the proposed framework to highlight and avoid potential drawbacks of TEVAR, even in the distal part of the stent-graft, limiting the need for secondary surgical intervention [42].

5 Limitations

The following limitations are important to be addressed when interpreting the results.
In the present study, we exploited post-operative PC-MRI data to set both the inflow rate and to calibrate the outlet boundary conditions of the CFD analysis in a patient-specific manner even though the tool aims at predicting the post-operative hemodynamics relying ideally on sole pre-operative data. Further developments of the present study should overcome such a limitation modeling the whole systemic circulation as proposed by Balossino et al. [43], who addressed this topic for the analysis of carotid hemodynamics describing how the peripheral arterial model should be adapted to predict the hemodynamic impact of vascular surgery. Moreover, we used the same boundary conditions in all the three CFD simulations; such a limitation is probably leading to an overestimation of the velocity magnitude in P2. Future studies should systematically include acquisition of patient-specific flow data from medical images [44].

Although the present study tackles a real clinical problem and is based on the analysis of patient-specific data, the investigation of two specific cases limits the extension of our medical conclusions to a more general scenario. Regarding the validation of the hemodynamics, a bigger cohort of patients have to be analyzed before bringing the tool to the bedside. Sensitivity analysis with variations of the boundary conditions should be also performed to give further significance to the differences between the predicted and the post-operative values.

Including fluid-structure interaction (FSI) in the flow simulations would lead to more realistic results, mostly because the velocity field is significantly affected by the wall compliance [45]. To overcome this issue, dedicated solvers for aortic FSI should be used [46, 47] with a particular attention on correctly representing in the model both vessel and prosthesis stiffness [48].

The computational costs of our simulations are still not compatible with clinically emergent procedures requiring quick decisions but remains suitable for elective cases of TEVAR or to further investigate borderline patients where the sole geometric analysis of the aorta morphology based on CTA is not enough.

With respect to the impact of using a rigid wall model in the stent-graft deployment, albeit the results reported in the Appendix are convincing, other constitutive models and material properties could be explored to see whether the final positioning is sensitive to the model of choice. Furthermore, other boundary conditions for the vessel (e.g., Robin conditions) could be used to reliably represent the surrounding tissues.

6 Conclusions

In this work we used structural FEA for the simulation of the prosthesis deployment integrated with CFD to predict patient-specific TEVAR hemodynamics in a realistic, post-operative model of the aorta. Considering our results, we demonstrated that stent-graft and vessel surfaces could be accurately merged for an appropriate predictive reconstruction of the fluid-domain, suitable for CFD analysis. The framework has been benchmarked with two clinical cases: for one of the two cases, the comparison of the predicted hemodynamics with the actual one computed using real post-operative images provide acceptable agreement. This framework can thus constitute another tool to provide information to the surgeon about borderline patients during the pre-operative planning in elective setting, as shown for one of the two clinical cases, where proximal landing zone was barely acceptable for endovascular treatment. As a future development, the framework will be benchmarked with a larger number of clinical cases; in particular, special attention will be payed to the impact of landing zone angulation [49] on the final stent-graft
apposition [50].

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7 Appendix

7.1 Impact of vessel modeling on FEA results

It is worth noting in this context that endovascular surgeons choose prostheses with a 10%-15% of radial oversizing to improve fixation by pushing the aorta eccentrically [1]. It is therefore important to analyze whether the use of a rigid vessel constitutes a significant difference both in the deployment and the fluid-domain reconstruction phases of our simulation procedure. To this aim, we compared the outcome of the virtual deployment using a rigid wall model of the aorta and a deformable one. Following the workflow discussed in the present paper, the same device used in P1 was deployed in a Mooney-Rivlin hyper-elastic vessel where material properties were retrieved from [51] and 3D triangular shell elements with reduced integration (S3R) were used; the mesh consisted of 6,926 nodes and 13,698 elements. Then, the cylinder-like structure and the fluid-domain were created for the new geometry following the procedure discussed in Section 2.2. The displacements of the aortic annulus, the ends of supra-aortic branches, and the distal end of the descending aorta were set to zero as boundary condition for the structural FEA. Results of the two simulations is reported in Figure 10, where the final configuration of the stent-graft within the aorta is shown for both cases.

Approximate position of Figure 10.

The qualitative analysis of such results already indicated a negligible difference, which has been further illustrated in Figure 11, where both the membrane and the struts of the stent-graft computed in the two cases are superimposed.

Approximate position of Figure 11.

Finally, the spatial discrepancy resulted negligible also from a quantitative point of view as shown in Figure 12. In the skirt, the maximum distance between the results of the two approaches was 3.6mm with an average of 0.8mm. Such a difference had also a minimum impact on the
corresponding generation of the fluid-domain: in this case, the maximum distance was 3.5mm and the average was 0.6mm. In conclusion, the average difference between the two approaches had the same order of magnitude of the strut diameter and of the distance image resolution (0.4mm); furthermore, they were below 3% of the stent-graft diameter. We can therefore speculate that using a rigid wall does not affect significantly the results.

7.2 Stent-graft apposition in P2 case

The results of stent-graft deployment for the case of P2 are reported in Figure 13. The simulation revealed a non optimal apposition of the proximal part of the prosthesis to the aortic wall, as highlighted by the contour-plot of the point-wise distance between the stent-graft proximal ring and the aortic wall.

7.3 Pressure and velocity fields along the cardiac cycle in P1

Figure 14 reports the volume rendering of the velocity magnitude and contour plot of the pressure resulting from P1-virtual and P1-actual along the cardiac cycle. The comparison of the pressure distribution between the two cases under analysis shows agreement along all the considered four time instants (i.e., 8% - early systole, 20% - systolic peak, 35% - early diastole, 47% - late diastole). Velocity magnitude resulted slightly overestimated in all the snapshots in the virtual results with the outcome of the CFD simulations using the actual geometry. Given such considerations, we focused our attention to the systolic peak.

References


8 Tables

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Table 1: CFD mesh details for each investigated case. DOF: degrees of freedom.

9 Figures

Figure 1: Steps of the proposed framework. STEP 1 aims at virtually simulating prosthesis deployment, STEP 2 consists of the merging of the stent-graft and the vessel model to create a CFD-suitable volume mesh; STEP 3 deals with the quantification of post-operative hemodynamics by CFD analysis.
Figure 2: Successive steps of the virtual deployment procedure in P1. A: the landing point is selected based on surgeon’s choice. B and C: stent-graft is simultaneously crimped and bent to match the vessel centerline. D: the catheter is enlarged until the prosthesis contacts the inner vessel wall.

Figure 3: Mapping of the stent-graft surface from a cylinder. A: primitive NURBS cylinder following the stent-graft centerline. B: original deployed geometry with its sample points. C: extrusion of the NURBS cylinder representing the skirt. D: red section from C showing the annular stent (blue) together with the final position of the skirt (red) and the struts (black). S: superior, I: inferior, R: right and L: left.
Figure 4: Distance images for the two cases under investigation. A: vessel, B: volumetric stent and C: combined image of A and B. D: internal view of the final extracted surface. Zero level is highlighted in white.

Figure 5: A: Geometrical comparison of the post-operative stent-graft configuration predicted by the structural FEA (left) and the actual one obtained by image segmentation (right) in P1. B: Zoom view at the level of the inner curvature of the aortic arch highlighting the localized kinking of the prosthesis predicted by the simulation (left) and present also in the actual post-operative image elaboration (right).
Figure 6: A: Geometrical comparison of the post-operative lumen reconstructed by the proposed framework (left) and the actual one obtained by image segmentation (right) in P1. B. Contour plot of pointwise distance (magnitude) between virtual post-operative model and the actual lumen surface.

Figure 7: Volume rendering of the velocity magnitude and contour plot of the pressure resulting from P1-virtual and P1-actual. The reported values refer to the systolic peak.
Figure 8: Velocity at the systolic peak in different slices of the stent-graft in P1. Cutting planes correspond to the proximal plane of the first ring, the proximal plane of the fifth ring, and the distal plane of the eight ring. The magnitude of the velocity is normalized to its maximum value (cm/s) which is reported for each slice. Directions of the slice with respect to its position in the aortic lumen are indicated (S: superior, I: inferior, L: left, R: right, A: anterior and P: posterior).

Figure 9: Streamlines of the velocity field in the predicted aortas at the systolic peak.
Figure 10: Configuration of the stent-graft virtually deployed in P1 within a rigid-wall model of the aorta (left) and in a deformable one (right).

Figure 11: Membrane (on top line) and struts (on bottom line) of the stent-graft deployed in a rigid-wall model of the aorta - in light grey - superimposed to the results obtained using deformable model - in red.
Figure 12: Top: Lateral views of the contour plot describing the magnitude of the surface distance between the stent-graft deployed in a rigid-wall aortic model and in a deformable one. Bottom: Lateral views of the contour plot describing the distance between the reconstructions of the post-operative lumen in the same cases.

Figure 13: Results of the simulation of stent-graft deployment in P2. Left: final configuration of the stent-graft released in the pre-operative aortic model. Right: zoom view of the proximal part of the endovascular implant emphasizing the bird-beak. The contour-plot of the point-wise distance between the stent-graft proximal ring and the aortic wall shows the malapposition of the device especially in the inner curvature of the aorta.
Figure 14: Volume rendering of the velocity magnitude and contour plot of the pressure resulting from P1-virtual and P1-actual along the cardiac cycle, from top to bottom, the following instants of the cardiac cycle are depicted: 8% - early systole, 20% - systolic peak, 35% - early diastole, 47% - late diastole.