Finite element analysis of TAVI: Impact of native aortic root computational modeling strategies on simulation outcomes

Alice Finotello a, Simone Morganti b,*, Ferdinando Auricchio c

a Department of Experimental Medicine, University of Genova, Largo R. Benzi, 10, 16132 Genova, Italy
b Department of Electrical, Computer, and Biomedical Engineering, University of Pavia, Via Ferrata 5, 27100 Pavia, Italy
c Department of Civil Engineering and Architecture, University of Pavia, Via Ferrata 3, 27100 Pavia, Italy

Article history:
Received 16 February 2017
Revised 21 June 2017
Accepted 25 June 2017

Keywords:
Transcatheter aortic valve implantation
Patient-specific modeling
Finite element analysis

A B S T R A C T

In the last few years, several studies, each with different aim and modeling detail, have been proposed to investigate transcatheter aortic valve implantation (TAVI) with finite elements. The present work focuses on the patient-specific finite element modeling of the aortic valve complex. In particular, we aim at investigating how different modeling strategies in terms of material models/properties and discretization procedures can impact analysis results. Four different choices both for the mesh size (from 20k elements to 200k elements) and for the material model (from rigid to hyperelastic anisotropic) are considered. Different approaches for modeling calculations are also taken into account. Post-operative CT data of the real implant are used as reference solution with the aim of outlining a trade-off between computational model complexity and reliability of the results.

© 2017 IPEM. Published by Elsevier Ltd. All rights reserved.

1. Introduction

Aortic Stenosis (AS) is the most common form of valvular heart disease in developed countries, occurring in 3% of people older than 65 [1]. It is a degenerative disease of the aortic valve, compromising its function of regulating blood flow from the left ventricle to the aorta, with significant consequences on morbidity and mortality of patients, thus representing a current relevant clinical problem.

In the last decade, transcatheter aortic valve implantation (TAVI) has become the established treatment option for patients at high surgical risk, representing nearly the 30% of procedures for elderly patients with severe AS, not suitable candidates for conventional open heart surgery [2]. It is estimated that, since the first-in-man TAVI in 2002, more than 100.000 patients worldwide benefited from this revolutionary procedure [3]. However, despite the clinical success, there are still some complications associated with TAVI; the most relevant being post-operative paravalvular leakage, but also aortic root rupture, prosthesis migration, left bundle branch impairment may occur [4], which are contraindications typically related to the mutual interaction between the device and the aortic root wall.

For clinicians, such complications are difficult to predict due to patient variability, especially in terms of aortic root geometry and distribution and dimension of calcific plaques. For this reason, clinical operators look with enormous interest at tools potentially able to allow the surgeon to select the optimal valve for a specific patient, i.e., tools able to give predictive evaluation of the prosthesis post-operative performance (principally intended as degree of leaflet coaptation and entity of possible paravalvular leakage). Such procedure outcomes depend on the choice of the device, on the adopted implantation strategy, and, of course, on the pre-operative specific native valve configuration [5].

In this context of personalized medicine, patient-specific computational simulations, based on pre-operative images, represent a powerful tool capable to obtain such predictive information about the behavior of the device, both during delivery and after expansion. A detailed review about the state of the art of patient-specific simulations of TAVI is available in Vy et al. [6].

Since the first finite element study of TAVI [7], several authors have proposed different modeling strategies of the percutaneous procedure either to investigate the hemodynamic environment before and after TAVI [8], or to explore the feasibility of TAVI in patient specific morphologies [9]. Computer-based simulations can be employed also to reconstruct the loading forces induced by the stent on the aortic valvular complex [10], as well as to evaluate the radial force produced by the self-expandable or balloon expandable devices [11]. The prediction of the outcomes of percutaneous aortic valve implantation through numerical simulations has been
extensively proposed for the two most common devices: the balloon-expandable Edwards Sapien (Edwards Lifesciences, Irvine, CA, USA) \cite{12–15} and the self expandable Medtronic Corevalve (Medtronic, Minneapolis, MN, USA) \cite{16–19}. However, for this class of very complex analyses, validation still represents a crucial issue.

Only very recently, few papers dealing with the validation of the TAVI finite element simulation framework have been published. Grbic et al. \cite{17} proposed for the first time an automatic procedure to reconstruct patient-specific parametrical aortic valve models and compared the simulation results with postoperative images. However, very simplified aortic root and prosthetic device models were considered. Schultz et al. \cite{20} proposed a validation study (including both Corevalve and Sapien implantation procedures) based on 39 patients; however, their work represents mainly a medical paper and details about the adopted simulation strategy are missing. Finally, Bosmans et al. \cite{16} conducted an interesting study comparing finite element results and postoperative data, considering different aortic wall thickness values and different (simplified) material models.

When performing this kind of analyses, in fact, many parameters remain uncertain, including, for example, the real patient-specific mechanical properties of the aortic tissue which can only be assumed on a statistical basis \cite{21} without specific histological information (usually not available for patients undergoing TAVI). In fact, while some analysis ingredients (like prosthesis geometry and material properties) are well known in advance, others are not preoperatively available and may have significant impact on simulation outcomes. Several studies, for example, agree that the use of different aortic root material models can deeply affect simulation results \cite{6,18}.

Hence, the aim of the present work is to investigate how different possible modeling strategies of the aortic valve complex may affect the finite element results and what is the balance between acceptable accuracy for clinical purposes and reasonable computational efforts, again for clinical application. In particular, assuming that the device geometry and the material properties are known and that the morphology of the native valve can be reliably reconstructed from computed tomography (CT) images, the main arbitrary modeling choices are represented by the aortic valve and root material models and properties as well as by its discretization strategy, focus of the present paper. Simulation results are compared with a “exact solution” extracted from post-operative medical images.

2. Materials and Methods

An overview of the framework to evaluate TAVI post-procedural outcomes is given in Fig. 1.

A “high-fidelity” model of the prosthetic device is constructed from microCT images. Angio-CT scan data are used for patient-specific reconstruction of the aortic valve complex including calcifications, and intraoperative angiographic measurements are used to correctly replicate with finite elements the real implantation procedure. These (green dots in Fig. 1) are assumed as reliable data included in the developed simulation framework. The impact of arbitrary modeling choices of the aortic district (red dots in Fig. 1) on simulation outcomes is investigated through a comparison between the obtained results and post-operative data.

In the following sections, we provide detailed descriptions of the developed simulation framework, with particular focus on the native valve possible modeling choices.

2.1. Prosthetic model

The prosthetic device chosen by the medical equipe for implantation in the investigated clinical case is a Medtronic Corevalve size

<table>
<thead>
<tr>
<th>Table 1</th>
</tr>
</thead>
<tbody>
<tr>
<td>List of parameters used to reproduce the Nitinol behavior taken from Auricchio et al. \cite{23}.</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Nitinol material parameters</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Austenite Young’s modulus</td>
<td>51,700 MPa</td>
</tr>
<tr>
<td>Austenite Poisson’s Ratio</td>
<td>0.3</td>
</tr>
<tr>
<td>Martensite Young’s modulus</td>
<td>47,800 MPa</td>
</tr>
<tr>
<td>Martensite Poisson’s Ratio</td>
<td>0.3</td>
</tr>
<tr>
<td>Transformation strain</td>
<td>0.063</td>
</tr>
<tr>
<td>Loading</td>
<td>6.527</td>
</tr>
<tr>
<td>Loading start of transformation stress</td>
<td>600 MPa</td>
</tr>
<tr>
<td>Loading end of transformation stress</td>
<td>670 MPa</td>
</tr>
<tr>
<td>Temperature</td>
<td>37°C</td>
</tr>
<tr>
<td>Unloading</td>
<td>6.527</td>
</tr>
<tr>
<td>Unloading start of transformation stress</td>
<td>288 MPa</td>
</tr>
<tr>
<td>Unloading end of transformation stress</td>
<td>254 MPa</td>
</tr>
<tr>
<td>Start of transformation stress (loading in compression)</td>
<td>900 MPa</td>
</tr>
<tr>
<td>Volumetric transformation strain</td>
<td>0.063</td>
</tr>
</tbody>
</table>

29. The geometrical model of the Corevalve prosthesis is created from high-resolution micro-CT images of the real device sample. The reconstructed STL file is imported in Rhinoceros 5.0 (McNeill, WA, USA) where the CAD model of one elemental unit is built (see Fig. 2a).

Matlab software (Mathworks Inc, Natick, MA, USA) is used to replicate in polar series the elementary unit in order to obtain the entire description of the device (Fig. 2b). Then, a structured mesh of first-order hexahedral solid elements with a reduced integration scheme is defined for the device model. In particular, C3D8R elements in the Abaqus library (Simulia, Dassault Systèmes, Providence, RI, USA) were used. Approximately 80,000 elements are adopted to discretize the entire structure using three elements in the radial direction to prevent locking issues \cite{22}. Material properties of the Nitinol alloy are considered according to the model proposed by Auricchio et al. \cite{23}. Material parameters are listed in Table 1; the density is set to 6.5 e-9 T mm^-2. In this study, the transcatheter valve leaflets are not included since they do not affect the mechanical behavior of the stent and its interaction with the aortic root wall. A cylindrical surface, in the following labeled as catheter, is built and used in the numerical analysis to reproduce the crimping technique. The catheter is defined through a surface obtained by sweeping a cylindrical section having a radius length equal to 22 mm and meshed using 11,040 quadrilateral surface element with reduced integration (SFM3D4R). It is modeled as a rigid material with a density equal to 6.7e-9 T mm^-3. A frictionless contact is defined between the outer Corevalve surface and the inner surface of the catheter, while a self-contact formulation is used for the stent.

2.2. Native aortic root model

Cardio-synchronized CT images of a 76 year-old male patient acquired at IRCCS Policlinico San Donato (Italy) in the diastolic phase with a Siemens MedCom Volume CT (pixel spacing: 0.621/0.621; slice thickness: 1 mm) are used as starting point to create a patient-specific model of the aortic valve complex, consisting of aortic root wall, valvular leaflets, and calcific plaques (see Fig. 3). The aortic wall surface is extracted with Itk-Snap 3.0 software (www.itksnap.org) and processed with an in-house Matlab code. Since it has been proven that the vessel wall thickness induces negligible effects on the deformed valve configuration (a 6% maximal diameter deviation occurs when the thickness of the aortic root is doubled \cite{16}), for simplicity, a constant thickness of 2.5 mm is considered to recreate the outer profile of the wall. The resulting volume is then discretized using C3D4 tetrahedral elements. Native leaflets are geometrically reconstructed following the procedure described in Morganti et al. \cite{13} and modeled with...
Fig. 1. Workflow of the computational framework to evaluate TAVI post-procedural outcomes. While certain parameters are assumed as given (green dots), others are investigated in terms of impact on analysis results (red dots). Yellow boxes refer to the required medical input data. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article).

Fig. 2. (a) Elementary unit geometric model, which lies on a surface determining the curve profile (focus is given to the adopted mesh); (b) Entire reconstructed stent.

Fig. 3. (a) The surface extracted with the Itk-Snap software is overlapped to the aortic wall model; (b) The model of the aortic valve district which consists of aortic root wall, valvular leaflets, and calcific plaques.

4-node shell elements with reduced integration (S4R) and constant thickness of 0.5 mm. Frictionless contact is considered between the aortic root and the leaflets; self-contact is applied to the leaflets.

**Discretization strategy**

We consider four different mesh sizes for the discretization of the aortic root wall and leaflets. As showed in Table 2, the number of elements is 22,370, 39,932, 84,259, and 207,801. We name the
Table 2
Mesh refinement on the aortic root wall and leaflets is performed. Four different meshes are considered, ranging from a coarse model to a very fine one.

<table>
<thead>
<tr>
<th>Discretization strategy</th>
<th>M1</th>
<th>M2</th>
<th>M3</th>
<th>M4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Description</td>
<td>Coarse</td>
<td>Medium</td>
<td>Fine</td>
<td>Very fine</td>
</tr>
<tr>
<td>Number of elements</td>
<td>22,370</td>
<td>39,932</td>
<td>84,259</td>
<td>207,801</td>
</tr>
<tr>
<td>Approximate element size [mm]</td>
<td>1.8</td>
<td>1.5</td>
<td>1.0</td>
<td>0.6</td>
</tr>
</tbody>
</table>

![Fig. 4. Experimental data (black and white dots representing the circumferential and longitudinal behavior, respectively) are plotted versus model predictions.](image)

**Material models**

To evaluate the effect of the material model on the aortic root post-operative configuration, four options for modeling the native aortic root wall and valve are taken into consideration: a rigid configuration (R), an isotropic linear elastic (LE) material model, a non-isotropic hyperelastic (HE) model, and an anisotropic Holzapfel–Gasser–Odgen (HGO) model. Beside the first very simplified rigid option, the material parameters of each considered model are calibrated with experimental data obtained from biaxial tests [24]. The linear elastic model is fitted by considering only the circumferential response since the aortic root is supposed to experience mainly a circumferential dilatation when the device is implanted. Results of the fitting procedure are illustrated in Fig. 4. The impact of the material model is investigated adopting for all simulations the mesh M3.

- Concerning the rigid configuration, a rigid body constraint is assigned to the aortic wall elements, whereas deformable tissue properties are attributed to the leaflets.
- For the LE model we use a Young’s modulus of 2 MPa and a Poisson’s ratio ν of 0.475.
- With respect to the hyperelastic (HE) model we choose a six-order reduced polynomial constitutive model as highlighted by

$$\mathbf{U} = \frac{1}{\mathbf{D}_1} \sum_{i=1}^{N} \mathbf{D}_i (\mathbf{f}_i - 1)^{2i}$$

where \(\mathbf{f}_i\) is the elastic volume ratio and \(\mathbf{D}_i\) the first deviatoric strain invariant. \(\mathbf{C}_{ij}, \mathbf{D}_i\) are material parameters and \(N\), set to 6 in our case, determines the order of the strain energy potential. The aortic tissue is assumed to be nearly incompressible and a Poisson’s ratio of 0.475 is considered. In Table 3 the constant values resulting from the fitting procedure described in Aurichio et al. [25] are reported.

- The strain energy function of the adopted HGO constitutive model is reported in Eq. (2).

$$U = C_{10} (I_1 - 3) + \frac{1}{\mathbf{D}_1} \left( \frac{(\mathbf{f}_i - 1)^2}{2} - \ln \mathbf{f}_i \right) + \frac{k_1}{2k_2} \sum_{a=1}^{N} \psi_{a2}(\mathbf{\tau}_a)^2 - 1$$

with \(\mathbf{\tau}_a\):

$$\mathbf{\tau}_a = k(\mathbf{I}_1 - 3) + (1 - 3\kappa)(\mathbf{\tau}_{\text{area}} - 1)$$

\(C_{10}\) is used to describe the material matrix and \(k\) characterizes the level of dispersion of the collagen fiber orientations. \(k_1\) is a positive material constant with the dimensions of stress whereas \(k_2\) is a dimensionless temperature-dependent material parameter. \(N=2\) is the number of fiber families, which have mean orientations described by the vectors \(A_1\) and \(A_2\) in the reference configuration. Following the fitting procedure implemented in [25], we assess \(C_{10}=13.23\ \text{kPa}\), \(k_1=54.406\ \text{kPa}\) and \(k_2=85.01\). Fiber preferred orientation angle \(\pm \beta\) is set equal to 30. A value of \(k=0\) is used in the numerical simulation and it accounts for no fiber dispersion.

**Calcifications**

It is well known, at least to the medical community, that calcifications may drastically affect the efficacy of TAVI procedure [26]. However, from available CT images, it is still very difficult to attribute specific material properties to patient’s calcifications.
Therefore, we consider three different cases: in the first case, we simply neglect calcifications (named NC in the following); in the second case, we assign linear elastic properties $E = 10\,\text{MPa}$ and $\nu = 0.35$ (C,10MPa) [9]; in the third case, we assign again linear elastic properties but modeling a much stiffer material reproducing hydroxypapite behavior (C,60GPa). In particular, the following parameters have been used: $E = 60\,\text{GPa}$ and $\nu = 0.3$, as reported in Wang et al. [27]. In the two last cases, a semi-automatic gradient-based level set method with threshold initialization, implemented in the VMTK (www.vmtk.org) library, is employed to recognize the calcific blocks (Hounsfield Units $> 900$). A fine mesh of approximately 50,000 linear tetrahedral elements (C3D4) for all the calcifications is used. By means of a kinematic coupling constraint technique, interaction between calcific blocks and leaflets is defined. A frictionless general contact is used to handle the interactions between calcifications and the aortic root inner surface.

2.3. Positioning strategy from angiographic images

Morganti et al. [19] demonstrated that the prosthesis positioning has noteworthy effects on post-operative outcomes. For this reason, Synedra View Personal 3.4 (Synedra Information Technologies, Innsbruck, Austria) is employed to analyze the intra-operative angiographic images performed during the intervention in order to extract the real depth of implantation $d$, i.e., the distance of the distal margin of the stent from the aortic annulus (Fig. 5a), and the tilt angle of the device relative to the root axis $\varphi$. Since it is not trivial to define the axis of the aortic root from angiographies, as illustrated in Fig. 5b, the $\varphi$ angle is calculated starting from the $\varphi$ angle formed by the longitudinal axis of the device and the straight line representing the aortic annulus. The following values were obtained: $d = 4.7\,\text{mm}$ and $\varphi = 3.2\,\text{deg}$ and considered for stent placement simulation. We neglect the pressure gradients since, in the clinical practice, rapid pacing is applied during implant deployment.

2.4. Simulation details

Following the real procedure, the stent release simulation reproduces the progressive expansion of the device due to the progressive sliding from the bottom to the top of the constraining catheter. Nodes at the bottom side of the stent are constrained to prevent longitudinal translations of the prosthetic device. Self general contact is defined for the stent. Frictionless contact is defined to model the interaction between the inner surface of the sliding catheter and the stent as well as the outer surface of the stent and all valvular structures. The time-period of the device expansion analysis is set to 0.4s. A semi-automatic mass-scaling strategy is used to speed up the analysis. During the entire simulation, kinetic energy is monitored to ensure that the ratio of kinetic energy to internal energy remains less than 10%.

2.5. Comparison with post-operative CT

Post-operative CT of the patient is performed on a Siemens MedCom Volume (0.4238/0.4238 pixel spacing; 3 mm slice thickness). VMTK open-source software is used to extract a 3D representation of the implanted stent after processing DICOM images. We register each resulting deformed stent surface coming from the simulation and compare it to the real reconstructed one using the Iterative Closest Point algorithm [28]. Different quantitative analyses are then performed to evaluate the reliability of simulation results. Only the stent is considered for two reasons: (i) the valve is not visible from CT images, (ii) we reasonably assume that the stent governs the post-operative performance of the valve. The following quantities are thus computed to evaluate differences between the measured patient outcome and the simulations:

- **Distance map.** By means of the VMTK tool, the distance between corresponding points of the two stent surfaces, i.e., the implanted and simulated ones, is computed.
- **Mean distance.** On cross-sections at three different characteristic levels ($L_1, L_2, L_3$ as shown in Fig. 6). Matlab software is used for this purpose: the centerline of the stent structure is built; then the three planes associated with the transversal section are identified and the nodes lying on those planes are selected and fitted with ellipses. For each of the three levels, the mean distance between the ellipse corresponding to the simulated stent and the ellipse related to the extracted one is calculated.
- **Stent eccentricity.** Eccentricity of the stents is measured at levels $L_1, L_2, L_3$. The adopted formula reads: $e = c/a$ where $c$ and $a$ are the major and the minor axis, respectively. For each plane, the differences between eccentricities of the simulated and the extracted stents are investigated.

Fig. 5. Measurements of the real device delivery: (a) depth of implantation and (b) implantation angle.

Fig. 6. Measurements are taken considering three characteristic planes at three different levels L1, L2 and L3.
Table 4
Mean distance results, from the real extracted stent to the simulated ones, are reported.

<table>
<thead>
<tr>
<th>Mean Distance [mm]</th>
<th>Level L1</th>
<th>Level L2</th>
<th>Level L3</th>
</tr>
</thead>
<tbody>
<tr>
<td>NC</td>
<td>0.7</td>
<td>1.45</td>
<td>1.2</td>
</tr>
<tr>
<td>( C_{1.0}) MPa</td>
<td>0.6</td>
<td>1.1</td>
<td>1.1</td>
</tr>
<tr>
<td>( C_{6.0}) MPa</td>
<td>1.15</td>
<td>1.4</td>
<td>1.6</td>
</tr>
</tbody>
</table>

Remark. Stent eccentricity measure should be considered very carefully when checking the agreement of a simulation result with the real configuration. In fact, if we consider two identical ellipses lying on a plane and rotated by \( \pi/2 \), they present the same eccentricity, though they are representing very different results. In any case, circularity and axial symmetry of the device (given by the eccentricity measure) is a very important parameter to predict the quality of postoperative coaptation and, thus, prosthesis performance.

Finally, the Von-Mises stress pattern on the aortic root wall due to the interaction with the expanded device is also measured and considered for comparative purposes among the different simulated cases.

3. Results

The quantities computed to describe stent deformation (i.e., distance map, mean distance, and stent eccentricity) as well as stress distributions on the aortic root wall are reported in this section with the aim of highlighting the impact of different modeling strategies of the aortic valve complex in terms of calcifications, mesh size, and adopted material model on simulation outcomes. All the simulations are performed using Abaqus Explicit solver on 12 CPUs. Total CPU time of each simulation is reported in Fig. 7.

3.1. Presence of calcifications

In order to assess whether the calcifications effectively impact on the configuration of the deployed stent, three situations (namely NC, \( C_{1.0}\) MPa, and \( C_{6.0}\) MPa) are analyzed and compared. For each configuration, the 3D reconstruction of the post-operative stent is registered and compared with the computer simulation outcome (see Fig. 8).

In Table 4, the mean distances between simulated and extracted models for the levels L1, L2, and L3 are reported.

Eccentricity values are reported in Fig. 9.

The deformed geometries of the aortic root wall obtained from the three different simulations are plotted in Fig. 10, providing the tential state of the aortic wall. Color scale shows stress in MPa. It should be noted that the calcified blocks cause stress concentrations on the aortic wall at the level of the sinuses (see Fig. 10(b)).

3.2. Simulation results varying aortic root mesh

We then evaluate the deployment of a Corevalve size 29, varying the adopted strategy to discretize the patient’s aortic root wall and leaflets. The final stent configurations at the end of the expansion simulation are investigated in comparison with the shape of the real implanted device (named “real”). Moreover, the different stress patterns are computed and represented.

3.2.1. Deformed configuration of the deployed stent

The deformed geometries of the stents obtained from the four different simulations are represented in Fig. 11: the distribution of the point-wise distance, from the real implanted device to each simulated configuration, is shown.

As reported in Table 5, mean distance for all the models and for each one of the three characteristic levels (L1, L2, L3) is computed.

We also measured eccentricities of the ellipses. The bar graphs depicted in Fig. 12 show the comparison between the real implanted device eccentricity and the measurements computed on the virtually deployed stents.

3.2.2. Von–Mises stress on the aortic root

In Fig. 13, the results of the stent placement for the different models are reported in terms of Von–Mises stress distribution on the native aortic root wall. These results are referred to the last frame of the expansion step. For each one of the four final configurations, the average value of the Von–Mises stress of the elements is also determined. To avoid the results being affected by isolated peaks of stress, we exclude the 1% of the volume with higher stress values, and we evaluate the average Von–Mises stress \( \bar{\sigma}_M \) of the remaining elements discretizing the 99 percentile of the original aortic root volume. The obtained results are: \( \bar{\sigma}_{M_1} = 17.2 \) kPa, \( \bar{\sigma}_{M_2} = 14.8 \) kPa, \( \bar{\sigma}_{M_3} = 13.2 \) kPa and \( \bar{\sigma}_{M_4} = 13.06 \) kPa for models M1, M2, M3 and M4, respectively. In Fig. 14 the resulted mean stresses are then plotted against the normalized index representing mesh size.

3.3. Simulation results varying native root material model

We evaluate the deployment of a Corevalve size 29 varying the adopted strategy for the material model of the patient’s aortic tissue. The final stent configurations at the end of the expansion simulation are investigated in comparison with the shape of the real implanted device. Moreover, the different stress patterns are computed and represented.

3.3.1. Final configuration of the deployed stent

Taking the stent extracted from the post-operative CT as reference surface, the stent surfaces resulting at the end of the implantation simulations are superimposed to the reference one and the
Fig. 8. Distance map between post-operative real configuration and simulated ones: (a) NC, (b) C_{10} MPa, (c) C_{60} GPa. Minimum and maximum values are set to 0.0 and 3.0 mm, respectively; black-colored areas are related to higher values.

Fig. 9. Differences between eccentricities, at levels L1, L2, and L3 are shown.

Contour plot of their relative distance is computed. In Fig. 15 the distribution of the pointwise distance to the real implanted device in the R, LE, HE and HGO cases is shown.

Table 6
Mean distance between real implanted stent and simulated one for all considered material models, measured at levels L1, L2 and L3.

<table>
<thead>
<tr>
<th>Material</th>
<th>Level L1 (mm)</th>
<th>Level L2 (mm)</th>
<th>Level L3 (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>R</td>
<td>2.5</td>
<td>1.6</td>
<td>1.8</td>
</tr>
<tr>
<td>LE</td>
<td>0.35</td>
<td>0.8</td>
<td>0.8</td>
</tr>
<tr>
<td>HE</td>
<td>0.6</td>
<td>1.1</td>
<td>1.1</td>
</tr>
<tr>
<td>HGO</td>
<td>0.6</td>
<td>1.1</td>
<td>1.2</td>
</tr>
</tbody>
</table>

In Table 6, results concerning mean distances for all the tested material models are provided.

Bar plots of the eccentricities are finally reported in Fig. 16.

3.3.2. Von-Mises stress on the aortic root.

Von-Mises stress distribution on the aortic wall after the Corevalve stent deployment is represented in Fig. 17. As shown, both the HE and the HGO material models lead to reduced stress values on the aortic root wall with respect to the LE material model.
4. Discussion

Transcatheter aortic valve implantation is becoming the standard procedure to resolve aortic stenosis in high-risk patients. The growing success of this minimally invasive technique has consequently attracted the interests of many researchers who developed advanced computational models potentially able to support preop-erative planning. Currently, in fact, there still occur a number of possible complications (ranging from paravalvular leakage to left bundle branch impairment, valve migration, or even ventriculoaortic junction rupture), which strictly depend on the native valve configuration (i.e., dimension, geometry, extension of calcifications), as well as on the device choice and adopted implantation strategy. In this context, computer-based simulations represent a powerful tool that can provide ambitious predictive information.

Fig. 13. The analysis results show the Von-Mises stress distribution on the aortic root wall. Figures from (a) to (d) refer, respectively, to the models M1, M2, M3 and M4. The values are calculated in MPa and the same color scale is used for all the images.
driving the surgeon during the decision-making and procedure planning process. For this reason, in the last decade, many contributions have been published on this topic. However, when modeling such a complex procedure, some analysis ingredients are not “a priori” fixed or known and can only be assumed, either because they are almost impossible to measure (e.g., biological tissue mechanical parameters), or because they are computational parameters (e.g., discretization size). It is very well known that these parameters can drastically affect the simulation outcome; however, it is not still clear and consolidated what is the measure of such an impact and how different modeling choices can really change clinical indications. Indeed, the aim of the present work is to shed some light on these modeling aspects, to support future studies in the context of personalized medicine based on computational predictions for TAVI.

As depicted in Fig. 1, there are some analysis ingredients that can be reasonably assumed as known (e.g., geometric and material details of the device to implant), and others that really depend on specific modeling choices. In particular, beside the aortic root geometry that can be accurately extracted from CT images using standard techniques, there are some modeling parameters that are arbitrary. In this paper, we focus on the three of them that in our opinion can more significantly affect simulation outcomes: (i) Valve calcifications, (ii) biological tissue parameters and models, (iii) aortic root discretization strategy. Regarding prosthesis implantation site, it has been already demonstrated that it affects simulation results [19]. However, in this work, we do not aim at predicting postoperative performance to support preoperative planning, and therefore we assume to know the device implantation position from available angiographic images. This made possible to compare simulation results with a “ground truth” from postoperative CT images.

Von-Mises stress distribution on the aortic root wall is also considered as an important outcome of the simulation since it can be associated to parameters of medical interest, like the risk
of triggering inflammatory processes or annulus injuries, as well as the risk of conduction branches impairment [15,29]. In fact, stresses induced on the aortic wall are analysed in numerous studies [9,15,30]. Of course, the obtained solutions are related to the adopted constitutive model and no exact or reference stress solutions are available. Thus, we can only assess a comparative comparison highlighting the impact of the constitutive modeling choice on the stress solution, which is the aim of the proposed stress results.

The presence of calcifications is unequivocally recognized as one of the most important factors determining (satisfactory or poor) postoperative results. We aimed at investigating not only the impact of modeling (or not) calcifications (which has already been faced by Russ et al. [18]) but we also tried to verify whether different plaque stiffness assumptions significantly alter the results or not. CT images can give information about the geometric extension of calcifications as well as about the density of the scanned materials. However, with current machine resolution, it is difficult to classify calcifications in terms of density and almost impossible to obtain reliable indications about their material properties. We thus choose to model calcium taking very different properties from the literature (E = 10 MPa [9] and E = 60 GPa [27]) and analyze the obtained simulation outcomes. We observe that, if we do not include calcifications in the native valve model, we obtain a less deformed stent configuration (see, for example, eccentricity measures very close to 1 in Fig. 9) suggesting better valve performance than in the real case. This is expected since calcifications are responsible of local device deformation affecting circularity and, consequently, valve coaptation. The distance measure (between really and virtually implanted stent) highlighted quite significant differences in three regions of the device, as also confirmed by the mean distance measures reported in Table 4. Varying the material properties of calcifications also significantly impact on the results: Fig. 8(c) shows that using very stiff material properties induces even greater variance from the real solution than the “no calcifications” case, while Fig. 9 is highlighting comparable results with the other simulated cases in terms of stent eccentricity at the three considered levels. The stress pattern on the aortic root wall is highly influenced by calcification modeling choices (Fig. 10), suggesting that, if the aim is to evaluate either the damage or even rupture of the aortic annulus, or the triggering of possible inflammatory processes within the aortic root, as well as the impairment of the cardiac electric bundles, particular attention should be paid in the modeling of calcifications.

As calcific plaque mechanical characteristics, also aortic root material properties are unknown for patient-specific cases. Histo- logical evidences show that, similarly to large arteries, the aortic root tissue has a three-layered structure, mainly composed by collagen fibers embedded in a ground matrix, consisting of elastin, proteoglycans, and water [31]. However, mechanical characteristics (stiffness, orientation, proportion of constituents, etc...) of the tissue vary from patient to patient and are usually not measurable for patients undergoing TAVI. In the present work, we aim at better understanding how different constitutive models affect the results. Four modeling strategies have been analyzed (from a very simple rigid material to a much more complex hyperelastic non-linear anisotropic one), all fitted to the same human experimental tensile data.

As expected, we found that simulation outcomes are sensitive to the aortic root constitutive model and parameters. The simplest choice of considering the aortic wall rigid leads to very unrealistic results of an overdeformed stent, i.e., eccentricity measure not measurable at all the levels and poor distance between real and virtual device- map due to the very distorted shape of the implanted device (see Figs. 16(a) and 15(a)). Fig. 16 shows that axial symmetry of the implanted device is almost everywhere maintained, independently from the adopted material model. Indeed, Fig. 15 seems to suggest that it is not true that more complex material models (i.e., HGO) lead to more accurate results. Therefore it is useless to adopt very complex constitutive models to represent the behavior of biological tissues, if they are calibrated using experimental data from the literature that are non representative of the specific patient. Being always impossible to know in advance the peculiar properties of a specific patient, a possible solution consists in grouping patients with respect to different factors (age, sex, lifestyle, presence of genetic diseases, etc...) and determine material parameters on a statistical basis. The same considerations can be drawn for stresses, that obviously depend on the adopted material model. However, if, on one side, stress values differ from model to model (see Fig. 17(b)–(d)), the location of high-stress concentrations can be identified, independently from the adopted model.

Finally, in dealing with the discretization strategy, if, on one side, it is recognized that the finer the mesh, the more accurate the obtained results, on the other side, it is not established which is the threshold between reliable clinical indications (from simulation results) and mesh size. Fixing the material model, we then studied the impact of aortic root mesh size on the simulation results. Very coarse meshes, as expected, lead to very poor results, as confirmed by Figs. 11(a) and 12(a). If the main interest is focused on the prediction of the final stent configuration (from which the main device performance parameters depend), there exists a threshold below which inaccurate results are obtained, but also above which negligible differences in terms of solution behavior are attained. In the case under investigation, such a threshold corresponds to M3 (mesh size in the order of 10^5 elements).

Considering M2, M3, and M4, it can be found that mesh refinement has a larger effect on stress values rather than on the overall deformation of the stent. More specifically, eccentricity and mean distance values for the stents of models M2, M3 and M4 are very similar (see Fig. 12), whereas an increase of 11.8% in the average stress is observed between M2 and M4 (see Fig. 14). By contrast, the difference between the mesh M3 and the finest mesh is 1.1%. Therefore, our results show that mesh M3 can be appropriately used for our purposes offering a good compromise between accuracy requirements and computational time.

5. Conclusions

Recent studies have shown that advanced computational simulations can represent powerful and helpful tools able to support interventionalist cardiologists during TAVI procedure planning. However, each study relies on specific assumptions and modeling strategies. In the present paper, we have focused on those that in our opinion represent the most arbitrary choices, all regarding the native pathological aortic root of the patient. The aim has been thus to identify the trade-off between model complexity and clinical reliability of the obtained numerical results. Indications for proper modeling the TAVI procedure have been provided. The impact of different calcification modeling strategies, aortic tissue material modeling approaches, and discretization choices on the simulation results of medical interest has been investigated.

6. Declarations

Competing interests

None declared

Ethical Approval

The patient-specific CT scan data on which the computer model of the native aortic root was part of a retrospective clinical study,
which was carried out in accordance with institutional guidelines. The patient gave written informed consent prior to each examination he/she underwent.

Acknowledgments

The work has been partially supported by iCardioCloud project by Cariplo Foundation (No. 2013–1779) and Lombardy Region (No. 42938382; No. 46554874).

The authors would also acknowledge Dr Anna Ferrara, Dept. of Civil Engineering and Architecture, University of Pavia, Italy, for providing the calibration data of the aortic tissues, and Dr Francesco Bedogni and Dr Nedy Brambilla, Dept. of Cardiology, IRCCS Pol. S. Donato, S. Donato Milanese, Milan, Italy, for providing the clinical case.

References