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

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ABSTRACT

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 20 k elements to 200 k elements) and for the material model (from rigid to hyperelastic anisotropic) are considered. Different approaches for modeling calcifications 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.

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

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extensively proposed for the two most common devices: the balloon-expandable Edwards Sapien (Edwards Lifesciences, Irvine, CA, USA) [12–15] and the self expandable Medtronic Corevalve (Medtronic, Minneapolis, MN, USA) [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. [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. [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. [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 [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 pre-operatively 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 [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 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 (McNell, WA, USA) where the CAD model of one elementary unit is built (see Fig. 2a).

Table 1
List of parameters used to reproduce the Nitinol behavior taken from Auricchio et al. [23].

Nitinol material parameters	
Austenite Young's modulus	51,700 MPa
Austenite Poisson's Ratio	0.3
Martensite Young's modulus	47,800 M Pa
Martensite Poisson's Ratio	0.3
Transformation strain	0.063
Loading	6.527
Loading start of transformation stress	600 M Pa
Loading end of transformation stress	670 MPa
Temperature	37 °C
Unloading	6.527
Unloading start of transformation stress	288 MPa
Unloading end of transformation stress	254 MPa
Start of transformation stress (loading in compression)	900 MPa
Volumetric transformation strain	0.063

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 [22]. Material properties of the Nitinol alloy are considered according to the model proposed by Auricchio et al. [23]. Material parameters are listed in Table 1; the density is set to 6.5 e^{-9} T mm^{-3}. 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 [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. [13] and modeled with