Advanced Haemodynamic Modelling of Transcatheter Aortic Valve Implantation: Insights into Leaflet Thrombosis and Blood Flow Dynamics

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Abstract

Aortic stenosis (AS) is a common heart valve condition, particularly in the aging population, with Transcatheter Aortic Valve Implantation (TAVI) widely established as the standard intervention. Despite its growing adoption, TAVI is associated with potential complications such as leaflet thrombosis, which manifests on CT as hypoattenuated leaflet thickening (HALT). Traditional post-procedure evaluations offer limited information on the haemodynamic factors that contribute to thrombus development. In this context, computational modelling, especially fluid-structure interaction (FSI) techniques, offers a powerful tool for analysing blood flow dynamics and valve behaviour. Nevertheless, current research lacks detailed patient-specific modelling that includes the actual implanted prosthetic valve. This study seeks to bridge that gap by exploring the haemodynamic conditions contributing to thrombus formation through advanced FSI simulations based on patient-specific geometries. We introduce our framework, which integrates the geometric modelling of the implanted TAV, reconstruction of individual aortic anatomies, and FSI-based simulation to evaluate the interplay between valve motion and blood flow.

1. Introduction

Aortic stenosis (AS) is the leading valvular heart disorder in the Western world, closely associated with ageing [1, 2]. It involves thickening and distortion of the valve leaflets, along with calcium build-up in the aortic root and valve. Transcatheter Aortic Valve Implantation (TAVI) is now the preferred initial treatment recommended by the European Society of Cardiology for patients 70 years or older or those deemed at high risk for open-heart valve surgery, matching Surgical Aortic Valve Replacement (SAVR) in early outcomes regardless of risk score [3]. With an aging population and an increase in life expectancy, the demand for TAVI is expected to surge. Although safety and efficacy have been demonstrated, it is crucial to recognize associated risks such as paravalvular leakage (PVL) and leaflet thrombosis, detectable through hypoattenuation and leaflet thickening (HALT) on computed tomography (CT) scans, which raise concerns of stroke and heart failure [4, 5]. HALT can clearly be seen on CT scans of 10-15% of patients [6]. Current methods for evaluating post-TAVI thrombus formation lack comprehensive haemodynamic insights. To address this gap, the application of computational modelling holds great potential in investigating haemodynamic and mechanical problems associated with TAVI implantation. In silico fluid-structure interaction (FSI) simulations present a robust approach towards understanding mechanical and haemodynamic interactions after TAVI implantation. While FSI holds significant promise, there is a lack of published studies investigating the interplay between blood flow dynamics and patient-specific anatomical structures, including the implanted prosthetic valve [7–9]. Additionally, realistic geometries of a 3D model of the TAV device are essential for precise computational modelling. However, the literature has placed limited emphasis on research dedicated to refining accurate geometries [10–13]. The primary objective of this study is to investigate the role of the haemodynamics surrounding the valve in order to better understand thrombus formation. This will be achieved through computational modelling of CT data of patients who underwent TAVI. The proposed pipeline comprises three main steps: reconstruction of TAV devices, reconstruction of patient-specific geometry, and FSI simulation.

2. Methods

2.1. Device reconstruction

The TAVI device utilized in this study belongs to the SAPIEN family of transcatheter heart valves manufactured by Edwards Lifesciences (Edwards Lifesciences, Irvine, CA, USA). The device consists of three peri-

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cardium leaflets attached to a metallic frame, with an outer skirt wrapped around the frame to improve contact with the native valve and reduce the risk of PVL. In a functional state, the leaflets have a complex shape that is difficult to replicate in CAD software. In this study, the leaflets were carefully separated from the device by cutting the sutures and attachment points, preserving their integrity. The flattened leaflets were then imaged using the AxioScan system (Oberkochen, Baden-Württemberg, Germany) with a pixel size of 0.879 × 0.879 microns. In most FSI simulations, the valve is assumed to start in a closed position to aid convergence. The 2D image was essential for determining the leaflet's geometric shape. To accurately model valve closure, a contact mechanics simulation was performed, assuming the 2D representation corresponded to a fully open valve. The closure of the valve was simulated using the Shell Nonlinear Analysis Programs (ShNAPr) library, which utilizes isogeometric discretization for shell structures [14]. Each leaflet was modelled as a multipatch NURBS surface consisting of four patches of quadrilateral elements and a thickness of 0.0386 cm. These non-matching multipatch NURBS surfaces were coupled using the framework PENGoLINS based on FEniCS [15]. A St. Venant-Kirchhoff (SVK) material model was applied with a Young's modulus of 1e7 and Poisson's ratio of 0.45. Pinned boundary conditions were applied to the attached edge. The dynamic simulation ran in 300 steps, applying a constant pressure of 5 mmHg with a time step of 1e−4 s.

2.2. Patient-specific model

A patient-specific model was developed using pre-TAVI CT scans acquired using a Siemens SOMATOM Force (Siemens Medical Solutions, Erlangen, Germany), with a tube voltage of 120 kV, collimation width of 0.6 mm, slice thickness of 1 mm, and pixel spacing of 0.916 mm × 0.916 mm. From these scans, a 3D model of the aorta was generated by automatically segmenting the pre-TAVI CT images with TotalSegmentator (see Fig. 1) [16]. The aortic centreline was subsequently extracted automatically using Mimics 26.0 (Materialise Inc., Leuven, Belgium) to serve as a reference for device positioning. To facilitate accurate placement of the TAV device, an interactive platform was developed using Grasshopper and Rhinoceros 3D 8.0 (Robert McNeel & Associates, Seattle, WA, USA). This tool allows for manual manipulation of the TAV model along the patient-specific centreline, including rotational adjustments for commissure alignment. The platform streamlines the modelling workflow by directly generating the necessary mesh files, including NURBS patches for the valve leaflets, the skirt geometry, and both fluid and solid meshes of the aortic domain. The volumetric mesh of the aorta is generated using Gmsh 4.10.5 (http://gmsh.info/) and consists of approximately 1 million tetrahedral elements. The solid domain is composed of two layers of structured elements, while the fluid domain features a central unstructured layer surrounded by two boundary layers of structured elements. To improve accuracy, the elements in the vicinity of the valve are refined, and the fluid domain is further resolved with boundary layer elements to better capture near-wall flow features. The aortic wall thickness is set to 0.2 cm [17]. Additionally, the inflow section of the aorta is constructed as an extruded cylindrical extension, aligned with the valve's position relative to the aortic sinuses.

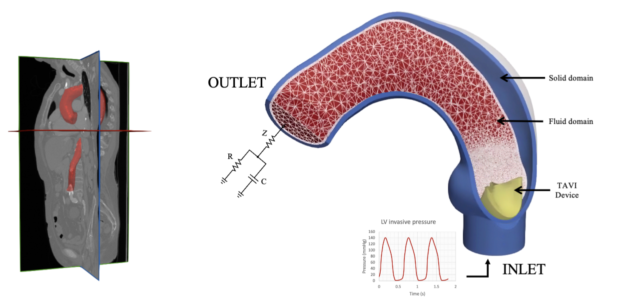

Figure 1. Pre-TAVI CT scan of the patient with the segmented aorta highlighted (left). Simulation model showing the fluid and solid domain meshes, including the valve geometry (right).

2.3. Simulation set-up

The FSI analysis was carried out using the library Coupling, via Dynamic Augmented Lagrangian (DAL) of Fluids with Immersed Shell (CouDALFISh), built on top of FEniCS [14]. The simulations employed isogeometric analysis and variational multiscale methods to accurately capture the interaction between the blood flow and the valve leaflets. Figure 1 illustrates a schematic representation of the FSI simulation setup, including boundary conditions, domain meshes, and valve placement.

The aortic wall was modelled as a neo-Hookean hyperelastic solid, incorporating mass damping (c = 1e-4 s^{-1}) and a dilational penalty to accurately capture energy dissipation from interactions with surrounding tissues. Sliding boundary conditions were applied at the inflow and outflow of the solid domain, while the region interfacing with the prosthetic valve stent was fixed in place. Leaflet contact was managed using a penalty function incorporating nonlocal regularization. The trileaflet valve, including skirting, was represented using a B-spline model with clamped boundary conditions at the bottom and sides and material properties defined by the isotropic Lee-Sacks model [18]. Blood flow was modelled as an incompressible Newtonian fluid with a dynamic viscosity of μ_f =

3e−2 g/cm^3 and density ρ_f = 1 g/cm^3. Boundary conditions included a left ventricular patient-specific pressure profile at the inlet, a three-element Windkessel model at the outlet and a spring-like condition at the solid wall. The simulation ran for three cardiac cycles with a time step of 2e−4 s.

To calibrate the Windkessel model efficiently, we developed a reduced-order surrogate model that reproduces the key dynamics of the FSI setup at a fraction of the cost. It combines a 0D lumped valve, a 1D elastic aortic segment, and the same three-element Windkessel model used in the FSI simulations. Driven by left ventricular pressure, the system propagates flow through the valve and aorta before absorption into the Windkessel circulation. Model parameters are tuned against measured aortic pressure, first for the valve and then for the Windkessel, with ongoing work exploring simultaneous fitting of all parameters.

3. Results

3.1. Valve closure

The contact mechanics simulation successfully produced a closed valve state, with a consistently observed small gap between the leaflets as can be seen in Figure 2. This residual gap is attributed to the challenges of modelling finite-thickness shell elements and fine-tuning contact parameters. Although the SVK material model can exhibit instability under compressive loading, the thin-shell formulation—focused on in-plane stresses—effectively avoided such issues. The leaflets responded rapidly to increasing pressure, demonstrating reliable closure behaviour, though the precise threshold pressure was sensitive to the resolution of loading steps. The emergence of curvature near attachment points at low pressures further highlighted the critical role of enforcing clamped boundary conditions.

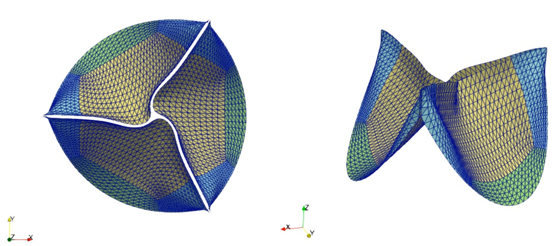

Figure 2. Closed configuration of the trileaflet valve obtained from the contact mechanics simulation. The geometry reflects the result of pressure-driven closure.

3.2. FSI

Figure 3 illustrates the FSI results for one representative patient, shown here for the second cardiac cycle of a three-beat simulation (the first cycle was discarded to allow numerical stabilization). The valve dynamics follow the expected sequence: at end-diastole the valve is fully closed, with minimal forward flow (panel A); early-systole shows the initiation of a narrow high-velocity jet through the central orifice (panel B); peak-systole is characterized by a concentrated high-speed jet across the valve leaflets and the development of recirculating regions downstream (panel C); during deceleration, the jet begins to break up and vortical structures dominate the ascending aorta (panel D); and finally, in early-diastole, the valve closes but persistent low-velocity flow remains near the commissures and leaflet edges (panel E). These regions of near-wall stagnation are of particular interest, as they may contribute to flow stasis and thrombus formation. Stress distributions further reveal that the valve does not close symmetrically, with the highest mechanical stresses concentrated at the commissural attachment zones and leaflet side edges. Overall, the simulation captures both the global haemodynamic patterns and the localized biomechanical features that may be relevant for valve durability and thrombogenic risk.

4. Discussion and conclusion

The 0D–1D surrogate model has been useful for rapid outlet BC fitting, though a trade-off between pressure and flow persists: matching pressures can underestimate cardiac output, while improving flow reduces pressure accuracy. This may stem from reduced-order assumptions and limited invasive measurements, which are taken only a few centimeters downstream of the valve. In the FSI simulations, outlet pressures are slightly overestimated since they represent distal rather than proximal values, yet the surrogate model provides valuable guidance for parameter tuning. Future work will focus on validating the pipeline against patient data and systematically comparing HALT and non-HALT scenarios to assess predictive capability. Together, the surrogate and FSI models form a promising framework for efficient, patient-specific simulations with potential clinical relevance.

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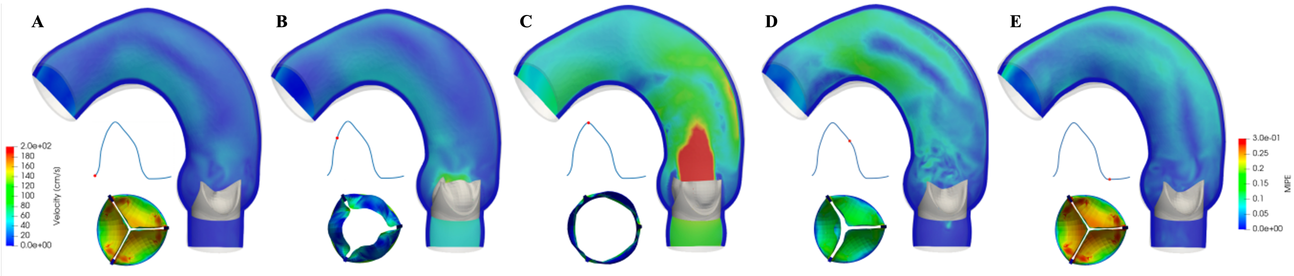

Figure 3. Fluid–structure interaction results for one patient during the second simulated cardiac cycle. (A) End-diastole: valve closed. (B) Early-systole: jet initiation. (C) Peak-systole: narrow high-speed jet with recirculation downstream. (D) Deceleration: jet breakup and vortices. (E) Early-diastole: valve closing with persistent low-velocity regions near commissures; leaflet stresses peak along commissural edges.

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