Evaluation of prosthetic-valved devices by means of numerical simulations

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The in vivo evaluation of prosthetic device performance is often difficult, if not impossible. In particular, in order to deal with potential problems such as thrombosis, haemolysis, etc., which could arise when a patient undergoes heart valve replacement, a thorough understanding of the blood flow dynamics inside the devices interacting with natural or composite tissues is required. Numerical simulation, combining both computational fluid and structure dynamics, could provide detailed information on such complex problems. In this work, a numerical investigation of the mechanics of two composite aortic prostheses during a cardiac cycle is presented. The numerical tool presented is able to reproduce accurately the flow and structure dynamics of the prostheses. The analysis shows that the vortical structures forming inside the two different grafts do not influence the kinematics of a bileaflet valve or the main coronary flow, whereas major differences are present for the stress status near the suture line of the coronaries to the prostheses. The results are in agreement with in vitro and in vivo observations found in literature.

Keywords: fluid–structure interaction; immersed boundary; bileaflet valve; Valsalva graft

1. Introduction

The replacement of the natural aortic valve with prosthetic devices (biological and mechanical) is a clinical practice performed diffusely, accounting for roughly 25 per cent of all cardiac operations. Patients receiving an aortic prosthetic valve have excellent chances of long-term survival with a good quality of life [1]. Nevertheless, several issues (e.g. thrombosis, haemolysis and tissue overgrowth) are still encountered and require an enhancement of the overall procedures. In order to deal with such potential problems, a deep understanding of the blood flow dynamics inside the devices, along with improvements of the prostheses and modifications of the surgical techniques, is required.

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Numerical simulations of aortic prostheses

Figure 1. (a) The straight (top) and Valsalva (bottom) grafts are illustrated. The central portion of the Valsalva graft, with vertical rather than horizontal corrugations, is called the skirt, and allows radial expansion under pressure, reproducing more closely the natural aortic geometry. (b) The St Jude bileaflet mechanical valve.

In the case of aortic valve disease associated with aneurysm or dissection of the aortic root and ascending aorta, surgeons replace the aortic valve and the entire ascending aorta simultaneously, with coronary reimplantation [2]. The use of composite valved conduits facilitates and expedites the surgical procedure, with excellent long-term results and an extremely low rate of complications [1]. Recently, some of the authors introduced a new conduit, with the aim of minimizing the tension on coronary ostia anastomoses [3]: the Valsalva graft reproduces a single egg-shaped proximal portion, better reproducing the natural geometry of the aortic root (three sinuses of Valsalva). Figure 1a shows the two prostheses adopted in practice by surgeons. An accurate in vivo evaluation of the flow dynamics and of the tensional status of the prostheses is not yet available in human subjects owing to the difficulties in performing invasive repeated measurements, thus requiring a numerical approach to investigate these aspects.

In this work, we present an accurate numerical investigation of the blood flow dynamics inside two aortic prostheses used in practice, combined with a structural analysis to evaluate the level of stresses on the devices. The tool developed is able to describe the complex and time-varying fluid patterns through a bileaflet mechanical valve interacting with the composite graft walls during a pulsatile cardiac cycle. The two coronaries are included in the model, thus allowing one to evaluate the influence that the aortic root may have on the early coronary flow as well as on the stress level near the suture line with the graft.

2. Modelling methodology and simulation details

The aortic valve considered is the bileaflet St Jude (St Jude Medical Inc., Minneapolis) mechanical prosthesis, with 25 mm diameter (figure 1b). The flat leaflets exhibit a rotation range equal to 53°, with a fully open position of 5° with respect to the streamwise direction (see figure 2 for the axis definition). The valve can be implanted in two different Dacron conduits: the standard straight graft and the Valsalva type. The geometrical model reproducing the
aortic root prostheses is composed of the following parts: the inflow tube and the valve housing, considered rigid; the collar, the skirt and the tube considered deformable and made of Dacron material; and the two coronaries attached to the skirt region of the graft, considered deformable. Figure 2 shows the Valsalva model used in the simulations during functioning: under the pressure of blood, the tissue in the skirted portion of the graft is stretched horizontally, creating a bulged portion. It is important to note that the geometrical model is the same for the straight and Valsalva grafts, the only difference being the different material description in the skirt region. The woven Dacron is described by an orthotropic (transversely isotropic) model. A linear model is considered, as shown by Lee & Wilson [4], with different elastic constants, both in the longitudinal and the circumferential directions. The values used in this work are the following: \( E_x = E_y = 12 \text{ MPa} \), \( E_z = 1.2 \text{ MPa} \), \( \nu_{xz} = \nu_{yz} = 0.15 \), \( \nu_{xy} = 0.1 \), \( G_{xz} = G_{yz} = 5.2 \text{ MPa} \), \( G_{xy} = 0.55 \text{ MPa} \), where \( E_i \) are the Young moduli, \( \nu_{ij} \) are the Poisson ratios and \( G_{ij} \) are the shear moduli. The different behaviour of the two prostheses is obtained by an inversion of the orthotropic directions in the skirt region for the Valsalva graft. A simple isotropic linear elastic model with Young’s modulus equal to 2 MPa is used to describe the natural tissue of the coronaries. The leaflets, made of pyrolytic carbon (\( \rho_l = 2000 \text{ kg m}^{-3} \)), are considered to be rigid with a moment of inertia with respect to the pivot axis of \( I = 7.05 \times 10^{-9} \text{ kg m}^2 \).

The incompressible Navier–Stokes equations, governing the motion of the blood considered as a Newtonian fluid [5], are solved using a fractional-step approach [6]. The time-integration procedure employs a variable time step, with a very high temporal resolution, in this case ranging from 2 to 200 \( \mu \text{s} \) during a cardiac cycle. The points inside the two coronary channels are considered as a porous medium, so as to modulate in time the porosity and obtain a mainly diastolic coronary flow as described by de Tullio et al. [7]. Moving and deforming geometries inside the flow are considered using the immersed boundary approach [8]. Ordinary differential equations, governing the rotation of the two leaflets about their pivots, are solved, together with the fluid equations by an implicit scheme, using an iterative procedure. A strong fluid–structure interaction
Figure 3. (a) Physiological flowrate (solid line with triangles) and pressure (solid line) waveforms imposed at the inlet section. (b) Leaflets kinematics for the St Jude valve in straight (solid lines) and Valsalva (dashed lines) graft. (Online version in colour.)

coupling is required to ensure stability and robustness of the simulation over the whole cardiac cycle owing to the high acceleration of the leaflets. Details on the method are given in de Tullio et al. [9]. The finite-element commercial software ANSYS Multiphysics™ (http://www.ansys.com), solving the deforming roots, is coupled with the fluid solver in a segregated weak approach: for each time step, the solution of the structural solver is needed only once, using pressure and viscous loads obtained by the flow solution. This reduces the computational cost of the procedure, and allows one to use optimized solvers for both the fluid and the structure. The overall scheme consists of solving, for each time step, the fluid and leaflet equations simultaneously, using the root position and velocity of the previous time step as boundary conditions; then, the structure solver is advanced in time under the loads evaluated by the flow solution, thus providing the new root configuration for the next time step.

Five complete cardiac cycles are simulated. The cardiac output considered is about 5 l min⁻¹ at a fixed beat rate of 70 beats per minute, resulting in a stroke volume of about 72 ml. The blood density is set to ρ_b = 1060 kg m⁻³. The peak Reynolds number, based upon the bulk velocity at the peak inflow, \( U = 0.95 \text{ m s}^{-1} \), the inflow tube diameter, and the blood kinematic viscosity, \( \nu = 3.04 \times 10^{-6} \text{ m}^2 \text{ s}^{-1} \), is about 7800. A background cylindrical structured grid is used, having 217 × 165 × 250 nodes in the azimuthal, radial and axial directions, respectively. This spatial resolution has already been checked by de Tullio et al. [9] against grid refinement benchmarks to be adequate to resolve all the flow scales at this Reynolds number. Physiological pressure and velocity profiles are imposed at the inflow section as shown in figure 3a. The deformable structure is fully constrained at the level of the mechanical valve, simulating the non-compliant artificial valve ring. In order to include the longitudinal stresses owing to the axial extension of the distal part of the aorta, a longitudinal maximum displacement of about 5 mm [10], modulated in time by the pressure curve in figure 3a, is applied to the nodes corresponding to the outlet section of the graft.
3. Results

Figure 4 shows the three-dimensional vortical structures obtained using the \( Q \)-criterion [11], at four different time instants reported in figure 3a (triangles). The figures show how the flow dynamics is mainly influenced by the bileaflet valve. The central vortical structure is due to the shear layer on the leaflet surface, and has the same behaviour in space and time for both prostheses. Both grafts exhibit recirculation regions owing to the different diameters of the tube with respect to the valve. The vortical structures that form in these regions are symmetrical with respect to the valve centre. For the straight graft, the structures move downstream, remaining bounded by the tube, which has a maximum radial deformation of about 1mm with respect to the unloaded configuration. In the case of the Valsalva graft, the vortical structures are larger: the skirt region of the graft enlarges, reaching a maximum radial deformation of about 4mm with respect to the unloaded configuration, thus creating a pseudo-sinus region. At the peak flow rate, owing to the adverse pressure gradient that induces the flow deceleration, the flow becomes turbulent: shear layers are unstable and the vortical structures break down, forming small-scale structures downstream of the valve. After valve closure, the mean flow stops and viscosity dissipates the structures until the new cycle. Although the vortical structures observed within the single egg-shaped sinus are more pronounced than those of the straight graft, their effects on the mobile elements of the mechanical valve are negligible. The opening and closing behaviour of the bileaflet valve is not altered by the graft type, as shown in figure 3b, where the angular position of the two leaflets for the two grafts is shown. High turbulence levels during the flow deceleration are responsible for the very small differences noticeable during valve closure, but this does not alter the overall closing time of the valve.

The numerical mean systolic pressure difference, that is the difference between aortic and ventricular pressure, averaged during the systolic period, is about 5.7mmHg and 5.9mmHg for the straight and Valsalva grafts, respectively. The numerical mean coronary flow obtained is about 135 and 140mlmin\(^{-1}\) for the straight and Valsalva grafts, respectively, again showing very weak influence of the sinus presence.

Figure 5 shows contours of the equivalent scalar stress [9], obtained by the full viscous stress tensor, at two instants of the cycle for the two prostheses. The strong shear layers shed from the valve housing and from the leaflets are evident during the acceleration phase, while the shedding of these layers is clear during the decelerating phase. It is important to note that the viscous stress levels are comparable for both grafts, showing maxima that do not exceed 70N.m\(^{-2}\).

The only noticeable and important difference between the two prostheses is observed for the stresses in the aortic root region. Figure 6 provides the time histories of the von Mises stresses near the coronary-root anastomoses in the circular region shown in figure 2 for both prostheses. The stress level for the Valsalva graft is lower than that of the straight graft for the entire cardiac cycle. In particular, the maximum stress experienced by the straight graft near the coronary suture line is twice that experienced by the Valsalva graft. Owing to the reduced stiffness of the Valsalva graft at the level of the recirculation region, high-frequency stress oscillations during systole are visible in figure 6. These results have relevant clinical implications that are discussed in the next section.

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4. Discussion and conclusions

The vortical structures that are generated by the flow crossing the mechanical valve inside the initial part of the Dacron grafts are difficult to visualize in detail without the help of an accurate computational fluid dynamics approach. In particular, studying the evolution of these structures in space and time is important to determine whether they could affect the dynamics of the mechanical valve, with a direct effect on the transvalvular pressure gradient, thus affecting the expected clinical performance of the valve.
The numerical results show that the hydrodynamic performance of the valve is not altered by the presence of the egg-shaped portion of the Valsalva graft. The opening and closing behaviour of the valve is maintained for both the grafts, as well as the pressure drop across the valve. All these results are in agreement with the \textit{in vitro} evaluations of De Paulis \textit{et al.} [1]. Moreover, the mean coronary flow is only slightly influenced by the presence of the sinus, in agreement with the \textit{in vivo} observations of De Paulis \textit{et al.} [12]. On the other hand, numerical simulations show that the presence of a bulged root section is important to reduce the stress level near the coronary suture line during the entire cardiac cycle, and this is a result that confirms and completes the findings of Weltert \textit{et al.} [13], obtained for a static configuration. The coronary ostia sutured to the straight graft have to withstand the highest dynamic stress levels during the cardiac cycle, while for the Valsalva graft, the highest levels of stresses are transferred from the coronary ostia sutures to the sino-tubular junction. However, in this region, the two portions of Dacron structures are strongly sutured together during manufacturing, therefore the stress has no effect on the patient’s tissues.

The results presented show how the numerical tool developed is able to reproduce accurately the flow and structure dynamics of such complex problems, giving results in agreement with \textit{in vitro} and \textit{in vivo} observations. In particular, concerning composite aortic graft performances, the beneficial effect of the presence of the sinus in reducing the tension at the level of coronary anastomoses is proved. This result may have a great importance in surgical practice, potentially reducing bleeding and pseudoaneurysm formation. Even if the results require clinical confirmation in the real case, since simplifications...
Numerical simulations of aortic prostheses

...are introduced during the simulations, the developed numerical tool is able to give an immediate physical insight and allow for comparative studies between different configurations.

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