Fluid-structure interaction of heart valve dynamics in comparison to finite-element analysis

Abstract: An established therapy for aortic valve stenosis and insufficiency is the transcatheter aortic valve replacement. By means of numerical simulation the valve dynamics can be investigated to improve the valve prostheses performance. This study examines the influence of the hemodynamic properties on the valve dynamics utilizing fluid-structure interaction (FSI) compared with results of finite-element analysis (FEA). FEA and FSI were conducted using a previously published aortic valve model combined with a new developed model of the aortic root. Boundary conditions for a physiological pressurization were based on measurements of ventricular and aortic pressure from *in vitro* hydrodynamic studies of a commercially available heart valve prosthesis using a pulse duplicator system. A linear elastic behavior was assumed for leaflet material properties and blood was specified as a homogeneous, Newtonian incompressible fluid. The type of fluid domain discretization can be described with an arbitrary Lagrangian-Eulerian formulation. Comparison of significant points of time and the leaflet opening area were used to investigate the valve opening behavior of both analyses. Numerical results show that total valve opening modelled by FEA is faster compared to FSI by a factor of 5. In conclusion the inertia of the fluid, which surrounds the valve leaflets, has an important influence on leaflet deformation. Therefore, fluid dynamics should not be neglected in numerical analysis of heart valve prostheses.

Keywords: Transcatheter aortic valve replacement, fluid-structure interaction, finite-element analysis, Arbitrary Lagrangian-Eulerian formulation

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1 Introduction

The aorta and the left ventricle are separated by the aortic heart valve that prevents the backward flow of blood. The pressure gradient between aorta and ventricle causes the valve opening during ventricular systole and the closure during diastole. As a result, the valve leaflets are constantly exposed to hydrodynamic forces.

Aortic valve stenosis and insufficiency represent the most common heart valve diseases. The number of patients suffering from one of the mentioned diseases is increasing due to the demographic change towards an aging population. An established therapy for these diseases is the minimally invasive treatment by transcatheter aortic valve replacement (TAVR) [1].

To investigate and improve the behavior of aortic valve prostheses, experimental studies are performed to examine the properties of the valve. However, not all mechanical data needed for the optimization of the design of an artificial heart valve can be determined experimentally. For this reason, numerical simulation of valve leaflet dynamics presents an essential tool. With these methods, hydrodynamic properties such as wall shear stress can be calculated. Therefore, numerical results contribute to help to optimize transcatheter heart valve prostheses in order to prevent emerging diseases. A simulation of the dynamic valve mechanics by means of fluid-structure interaction (FSI) provides insights in the interaction of heart valves and blood flow. Previous publications have focused primarily on finite-element analysis (FEA), which neglects the blood flow [2-3]. For this reason, the current study examines the influence of the hemodynamic properties on the valvular deformation by means of FSI and compares these results with a corresponding deformation resulting from FEA.
2 Materials and methods

2.1 Design parameter

The geometric model used in this study consists of a generic aortic root model with three identical leaflets that are based on previously published literature [4-5]. The aortic root was created using computer-aided design (CAD) software Creo Parametric 3.0 (Parametric Technology Corp., Needham, MA, USA). The shape of the aortic root was defined by five geometric parameters (see Table 1).

Table 1: Design parameter describing the aortic root model.

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>(R_o)</td>
<td>Large circle radius</td>
<td>11.01 mm</td>
</tr>
<tr>
<td>(R_i)</td>
<td>Small circle radius</td>
<td>3.67 mm</td>
</tr>
<tr>
<td>(h_s)</td>
<td>Sinus height</td>
<td>23.48 mm</td>
</tr>
<tr>
<td>(h_1)</td>
<td>Distance till max. sinus diameter</td>
<td>5.87 mm</td>
</tr>
<tr>
<td>(D_a)</td>
<td>Aortic/ventricular diameter</td>
<td>24.00 mm</td>
</tr>
</tbody>
</table>

The relation between the geometric parameters is described by an epitrochoid (see Figure 1):

\[
x(\varphi) = (R_0 + R_i) \cdot \cos \varphi - 0.5 \cdot \cos \left(\frac{R_0 + R_i}{R_i} \varphi\right) \tag{1}
\]

\[
y(\varphi) = (R_0 + R_i) \cdot \sin \varphi - 0.5 \cdot \sin \left(\frac{R_0 + R_i}{R_i} \varphi\right) \tag{2}
\]

The aortic sinus was also defined by its abdominal line (see Figure 2), which was determined by the ventricle and aortic diameter, as well as the maximum sinus diameter at 1/4 of the sinus height. The fluid domain inlet was located at the tube 10 mm upstream and the outlet 10 mm downstream of the heart valve model. The diameter of the tube upstream and downstream of the valve model is equivalent to the diameter of the valve model.

2.2 Finite-element analysis

Only the opening part of the cardiac cycle was simulated by FEA. The boundary conditions for physiological pressurization were based on measurements of ventricular and aortic pressure from in vitro hydrodynamic studies of a commercially available heart valve prosthesis using a pulse duplicator system (ViVitro Labs, Inc., Victoria, BC, Canada) according to ISO 5840-3. The pressure boundary condition was applied on the ventricular side of the aortic valve by using the transvalvular pressure gradient. The translational displacement at the commissure was constrained to fix the valve leaflets. Between the neighboring leaflets a frictionless surface-to-surface contact was defined. A linear elastic constitutive law with a Young modulus of \(E = 10\) MPa, a Poisson ratio of \(\nu = 0.46\) and a density of \(\rho = 1000\) kg/m\(^3\) was assumed for aortic valve leaflets. Structural analyses were performed using ANSYS Mechanical APDL (ANSYS, Inc., Canonsburg PA, USA).

2.3 Fluid-structure interaction analysis

The hemodynamic analyses utilizing FSI was defined and prepared by using ANSYS Workbench. For solving the structural part of the simulation ANSYS Mechanical APDL was used and for fluid dynamics ANSYS FLUENT was used. The boundary condition for the physiological pressure curve was equal to the one used for structural analysis but was applied to the inlet of the aortic root. The blood passing through was specified as a homogeneous, Newtonian incompressible fluid with a density of \(\rho = 1060\) kg/m\(^3\) and a viscosity of \(\eta = 0.0035\) Pa s.

For reduction of computational time, only a third of the entire valve model was calculated and reassembled by symmetry conditions. To get precise results at the interface a mesh conserving method was primary used. The fluid mesh
was adapted to the deforming structure for each time step. The mesh deformation was realized by using the arbitrary Lagrangian-Eulerian method. The motion of the fluid mesh was accounted by the ALE formulation of the momentum equation:

$$\frac{d}{dt} \int_V \rho \ u \ dv + \int_S [(u - u_g) \cdot n] \ dS = \int_V f \ dv \ .$$  (3)

The flow velocity is denoted by \( u \), \( V \) is the control volume, \( S \) is the control volume’s boundary, \( \sigma \) is the stress tensor, \( f \) are the body forces, \( n \) is the normal vector and \( u_g \) is the grid velocity. In order to preserve the integrity of the mesh which is required by the ALE method, a gap with an offset of 100 \( \mu \)m was defined between the coapted leaflets [6]. In case of large deformation, a re-meshing was employed to update poor quality cells. The Navier-Stokes and continuity equation in ALE form were solved using a pressure-based solver. The flow solver used a spatially second-order upwind scheme and a second-order time discretization.

### 3 Results and discussion

#### 3.1 Leaflet displacement

From the analysis of the numerical results of the leaflet deformation seven significant points in time were identified that describe the opening behavior of the leaflets. Figure 3 shows the pressure ratio on the ventricular and aortic side as well as the resulting pressure gradient at the significant time points. The FSI model with streamlines at time point \( t_5 \) is shown in Figure 3(c). Furthermore, the performance of the valve model is illustrated for both analysis types (see Figure 4).

The seven time steps show the opening behavior of the leaflets during FEA and FSI analysis. The time period before leaflets are fully opened in FEA is notably shorter than the opening time of the FSI model. Within 10 ms the opening process of the valve using FEA is completed, while the valve modelled by FSI is still closed. The opening process of the FSI valve lasts five times longer compared to the FEA model. At the time point \( t_6 = 140 \) ms both valves are fully opened.

A potential reason for the earlier valve opening in the FEA model is the disregard of the fluid flow. During the opening process of the leaflets the blood generates an additional resistance due to inertia of the fluid. Therefore, a higher pressure is necessary to open the aortic valve. The FSI approach calculates a higher and inhomogeneous pressure distribution at the proximal leaflet surface compared to the homogeneous pressure boundary used for the FEA.

![Figure 3: Aortic and ventricular pressure and mass flow curves for a cardiac cycle based on in vitro hydrodynamic testing of a commercially available heart valve prosthesis (a) with an extract of the pressure gradient showing six analyzed points in time (b) and the FSI model with streamlines (shown for a third) at time point 5 (c).](image)

![Figure 4: Results of dynamic leaflet behaviour from the FEA and FSI simulation illustrated at seven significant points in time: \( t_0 \) closed valves, \( t_1 \) start of opening FE valve, \( t_2 \) 75% opened FEA valve (FSI valve is still closed), \( t_3 \) start of opening FSI valve, \( t_4 \) buckling leaflets of the FSI valve, \( t_5 \) 60% opened FSI valve and \( t_6 \) fully opened valves (in relation to time).](image)
3.2 Leaflet opening area

The leaflet opening area (LOA) represents a projection of the free-margin of the valve leaflets on a cross-sectional plane. Figure 5 shows the leaflet opening area for the FE and FSI analyses over time normalized to the surface area of a circle defined by the valve diameter.

![Figure 5](image_url)

The comparison of the numerical data calculated with the FEA and the FSI model yields considerable differences for the opening area with a maximum deviation of up to 98%. Only for the time step $t_5$ and $t_6$, LOA for both models converges towards a value of 70%. In addition, the oscillation of the LOA in the FEA model at the beginning of the simulation was investigated. The LOA ranges from 55% to 85% during 90 ms and 104 ms. The leaflets undergo high stress values during opening while the resistance of the environment is inferior. The results of the FSI do not show a similar deformation because the inertia of the fluid is damping the displacement of the leaflets.

4 Conclusion

Within the current study, the time dependent deformation of valve leaflets was simulated using FEA and FSI. The results show considerable differences between the opening behavior of the FEA and FSI valve model. By neglecting the fluid domain the valve modelled by FEA is opening significantly faster compared to the FSI approach. This effect could be demonstrated by the comparison of LOA. Thus, it can be concluded that fluid passing through the valve has an important influence on valve leaflet dynamics. Firstly the pressure boundary applied to the fluid domain inlet for FSI effects different pressure distribution on the leaflets and secondly the presence of the fluid generates resistance against leaflet dynamics. For this reason FSI analysis provides a valuable tool for the optimization of the design of prosthetic heart valve devices. Moreover, FSI models have the potential to overcome the limitations of current FEA models. Within this study only the opening process was simulated. To obtain an assessment whether the FSI is more appropriate to describe the real valve leaflet dynamics than the finite-element analysis, the opening and closing behaviour of the valve model will be analysed experimentally in future studies.

Author Statement

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