2nd International Conference on Mathematical and Computational Biomedical Engineering – CMBE2011
March 30 – April 1, 2011, Washington D.C., USA
P. Nithiarasu and R. Löhner (eds)

THE INFLUENCE OF THE UPSTREAM BOUNDARY CONDITION IN THE NUMERICAL SIMULATION OF THE OPENING OF AN AORTIC BMHV

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SUMMARY

In this paper, the influence of the upstream boundary condition for the numerical simulation of an aortic Bileaflet Mechanical Heart Valve (BMHV) is studied. Two types of upstream boundary conditions are discussed and evaluated. First, an inflow velocity profile is imposed at the inlet of the valve. Secondly, a geometrical boundary condition is used, which implies that the flow rate is governed by the geometrical contraction of the left-ventricle (LV). Both boundary conditions are used to simulate a 3D case with the same BMHV. The change in time of the LV volume is calculated such that the flow rate through the valve is identical in both cases. The dynamics of the BMHV are modelled using fluid-structure interaction (FSI) and only the opening phase of the valve is simulated. The simulations show that although the results for the two cases are similar, differences occur in the leaflet movement. In particular, when using the velocity profile, the leaflets impact the blocking mechanism at their open position with a 25% larger angular velocity. Therefore, when one wants to simulate the dynamics of such an impact, the upstream boundary condition needs to be chosen carefully.

Key Words: *Boundary condition, Left-ventricle, Bileaflet Mechanical Heart Valve, FSI*

1. INTRODUCTION

When numerically simulating Bileaflet Mechanical Heart Valves (BMHVs), several types of upstream boundary conditions can be used [1,2]. However, since the dynamic movement of the valve leaflets is driven by the resulting flow field, the imposed boundary condition needs to be chosen carefully.

Commonly, an inflow velocity profile is imposed at the inlet [1]. Another approach is to implement a geometrical boundary condition. For a BMHV in the aortic position, this can be done by a contracting left-ventricle (LV), as is discussed in [2].

In this paper, the use of these two upstream boundary conditions is discussed and evaluated through 3D numerical simulations.

2. METHODS

In this section, the used 3D cases are discussed. First, the numerical simulation of the dynamics of a BMHV is described. Subsequently, the details of the used boundary conditions are discussed.

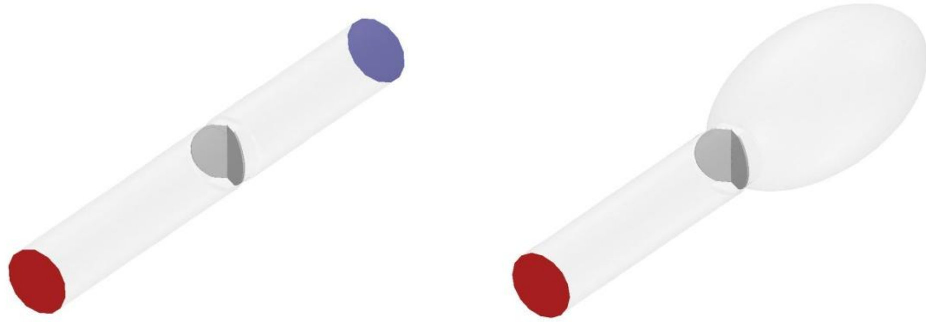
2.1 Fluid-Structure Interaction simulation of the BMHV

The numerical simulation of a BMHV is a complex Fluid-Structure Interaction (FSI) problem because the movement of the leaflets strongly interacts with the surrounding fluid motion and, therefore, the dynamic equilibrium at the fluid-structure interface needs to be taken into account.
The dynamics of the BMHV with rigid leaflets is calculated by a recently developed FSI algorithm [1]. This strong coupling algorithm uses separated solvers for the flow and the structural domain. It predicts the moments (and thus the angular accelerations) for the next coupling iteration through a linearization of Newton's Second Law with a finite difference approximation of the Jacobian. The components of this Jacobian are the derivatives of the moments (exerted by the flow on the leaflets) with respect to changes in leaflet angular accelerations. The Jacobian is numerically derived from the flow solver by variations of the leaflet positions. A more detailed description of the FSI algorithm can be found in [1].
The BMHV used in the simulations, is a simplified model of the 25mm ATS Open Pivot Standard Heart Valve in aortic position with the orifice inner diameter measuring 20.8mm [1].

2.2 Boundary Conditions

Downstream of the valve, the geometry consists of a rigid straight tube with diameter 22mm. A pressure is imposed at the outlet boundary, since in a rigid geometry the pressure level does not affect the flow field (only the pressure gradient appears in the equations).

Upstream of the valve, the flow rate is specified. This is done by the use of two different boundary conditions, which results in two cases, as is visualised in Figure 1.



(*a*) (*b*)

Figure 1: View on the geometry of the simulated cases, with the boundary conditions. Downstream: pressure outlet (in red). Upstream: (*a*) inflow velocity profile (in blue), (*b*) contracting LV.

In the first case, a rigid straight tube (with diameter 22mm) is placed upstream with a velocity profile imposed at the inlet. The used velocity profile is an aortic flow pulse with a time cycle of 1s and is displayed in Figure 3.a. It is the same uniform velocity profile that is used in [1].
The second case consists upstream of a contracting LV, as is done in [2]. The shape of such a LV is usually modelled as a prolate spheroid [3]. The short-to-long-axis ratio of the spheroid is kept constant at 0.5, which is considered the normal reference for a human LV [4]. The change in time of the LV volume is calculated such that the flow rate through the valve is identical in both cases.

An end-diastolic volume of 111ml is chosen. Contraction results in an end-systolic LV volume of approximately 41ml. Both volumes are well within the reference range for healthy men [3].
Both geometries are meshed with approximately 800 000 tetrahedral cells. Blood is modelled as a laminar incompressible Newtonian fluid with density and viscosity equal to respectively 1050 kg/m^3 and 4E-3 Pa·s. A no-slip boundary condition is applied at the walls.

3. RESULTS

The opening phase of the valve leaflets is simulated from t = 0s (begin of systole) to t = 0.125s (peak of systole). The velocity flow field is visualised in Figure 2 on a longitudinal cut plane at t = 0.2s. It shows that the flow through the valve is similar in both geometries.

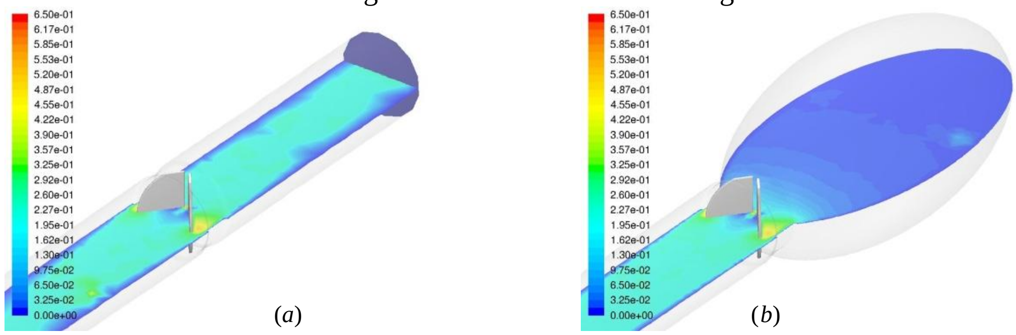


Figure 2: Velocity flow field (in m/s) at *t*=0.2s. (*a*) inflow velocity profile, (*b*) contracting LV.

The movement of the leaflets is depicted in Figure 3. Since the resulting leaflet motion is symmetric, only one of the two leaflets of each case is shown for clarity. The angular positions (Figure 3.a) are calculated relative to the fully opened position. Therefore, 0% refers to the closed position and 100% refers to a fully opened leaflet.

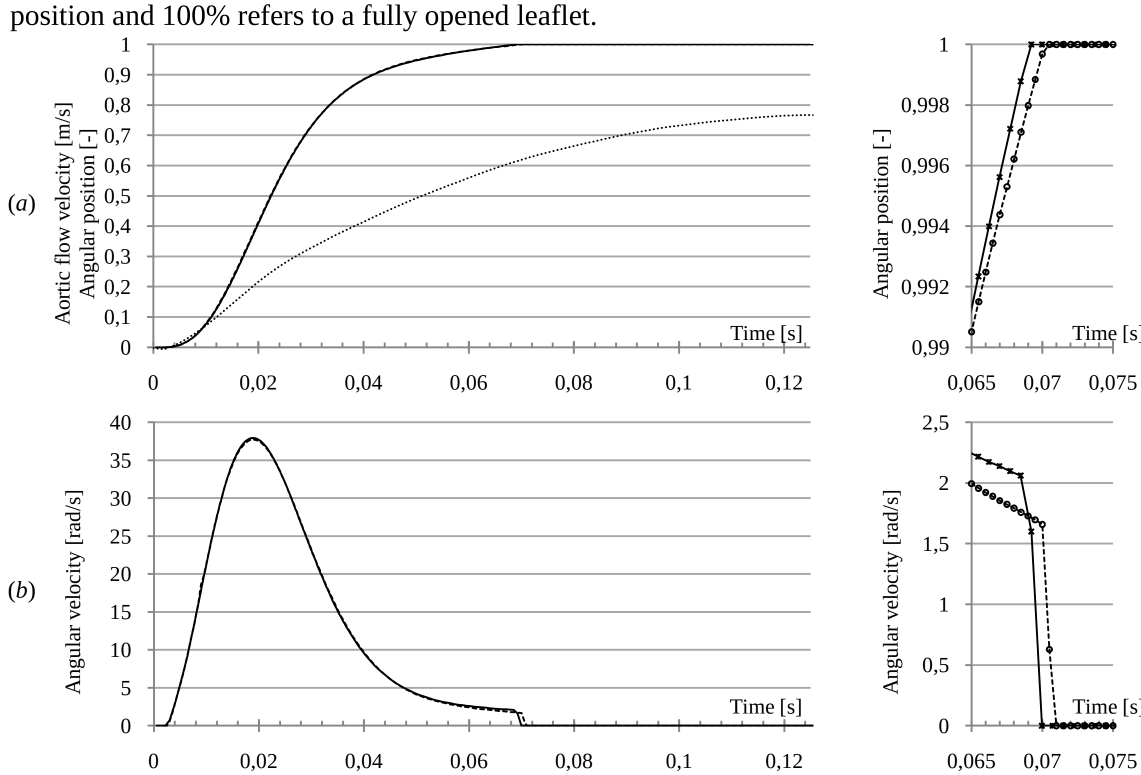


Figure 3: Plot of the aortic flow velocity (*a*) and the leaflet movement: angular position (*a*), angular velocity (*b*). Aortic flow velocity (···), Leaflet with ventricle (---), Leaflet with velocity profile (—). The impact at the open position is zoomed at the right.

It can be seen that although the results for the two cases are similar, differences occur in the leaflet movement. In particular, the leaflet reaches the fully open position a little sooner when using the velocity profile. Furthermore, when using the velocity profile, the leaflet impacts the blocking mechanism with a 25% larger angular velocity which will result in larger stresses in the leaflets. Also, the moments on the leaflets in the open position (after the impact) remain the largest when using the velocity profile (Figure 4).

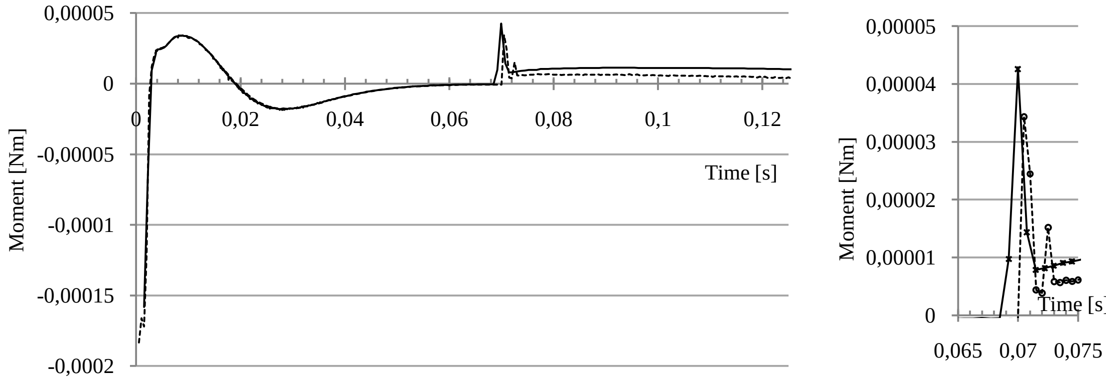


Figure 4: Plot of the moments: Leaflet with ventricle (---), Leaflet with velocity profile (—).
The impact at the open position is zoomed at the right.

4. CONCLUSIONS

In this paper, two types of upstream boundary conditions are used to simulate the dynamics of a BMHV. A first case consists of a rigid straight tube with a velocity profile imposed at its inlet. In the second case, the upstream rigid tube is replaced by a contracting LV. The contraction of the LV at every time level induces a flow rate that is identical to the flow rate due to the velocity profile. It is shown that a change in the specified upstream boundary condition can result in different leaflet motion. In particular, when using the velocity profile, the leaflets impact the blocking mechanism at their open position with a 25% larger angular velocity. Therefore, when one wants to simulate the dynamics of such an impact, the upstream boundary condition needs to be chosen carefully.

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