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# Modeling Aortic Valve Closure under the Action of a Ventricular Assist Device

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*Abstract*—**The support of a failing heart with pump devices has been an essential element in cardiac health care for several decades. It is therefore important to understand the left ventricular response to the pumping action of these devices when connected to the native heart. Furthermore, monitoring of aortic valve opening and closure is important in avoiding valve stenosis and thrombogenesis during pump support. This paper reports the first steps in simulating the effects of outlet pump pressure on aortic valve closure of the heart assisted by an implantable blood pump. A two-dimensional fluid structure interaction aortic valve model is presented with blood flow in left ventricular chamber using the Arbitrary Lagrangian–Eulerian Finite Element Method formulation to predict the AV closure during outflow of blood from the left ventricle into the left ventricular assist device (LVAD).**

## I. Introduction

ONE in five people develop some degree of heart failure (HF) in their life time in the United States [1]. Currently, left ventricular assist devices (LVADs) represent alternative therapies for congestive HF patients [2]. Due to the limited number of donor hearts, LVADs are often used as a bridge to transplant, or more frequently nowadays, a destination therapy [3]. In order to improve LVAD pumping efficiency, a full understanding of its effects on the hemodynamics of the left ventricle (LV), as well as the timing of aortic valve (AV) closure and opening, are essential [4].

Development of valve abnormalities after LVAD placement is common among patients with advanced heart failure, likely due the fact that the LVAD induces different hemodynamics by changing the direction of blood flow from the apex of the heart, largely bypassing the LV, directly to the aorta. This abnormal recirculation modifies pressure and stress on the AV, leading to remodeling of the valve [5]. AV functional problems, including aortic insufficiency (AI), prolonged valve closure, and aortic stenosis (AS) (an abnormal narrowing of the AV), are more prevalent in the LVAD patient after implantation [6, 7]. Furthermore, 50% of patients with pulsatile LVADs develop AI or AS within 6–12 months of LVAD implantation [8]. Similarly, a major consequence of the LVAD is excessive flow, resulting in diminished and infrequent AV opening, which could also extend to the systolic phase [9]. Currently, MRI and Doppler ultrasound are the most popular diagnostic methods to evaluate LVAD following implantation [10].

Alternatively, computational modeling of blood flow has been used extensively to study AV hemodynamics. Several Finite Element (FE) models of the left ventricle and aortic valve have been developed to simulate leaflet motion due to blood flow through the heart valves. Early studies of heart valve fluid-structure interaction (FSI) models includes the work of Peskin [11], who used an immersed boundary method to describe a 2D model of the left heart. Watanabe *et al.* [12] addressed the limitations of the Peskin approach, using instead the arbitrary Lagrangian–Eulerian (ALE) approach to simulate cardiac mechanics during cardiovascular flow. Alternately, de Hart *et al.* [13, 14] used a fictitious domain (FD) method to simulate closure of a flexible heart valve during systole. Loon *et al.* [15], extended this approach by linking fluid dynamics with adaptive meshing for the flexible leaflet. Stijnen *et al.* [16] also simulated the dynamic behavior of a two-dimensional moving rigid heart valve using the FD method. Lastly, McCormick *et al.* formulated a FSI model of the left ventricle with LVAD using a modification of the Newton–Raphson/line algorithm and optimizing the interpolation scheme at the fluid–solid boundaries [4, 17].

However, these models did not include simulation of AV closure during LVAD support. Understanding how the AV behaves under LVAD support may allow the detection of the open/closed state of the valve from LVAD pressure/flow sensor transducers alone, paving the way for more sophisticated pump control algorithms which take into account the AV state.

In this paper we present a 2D aortic valve computational model capable of simulating the aortic valve dynamics (closure/opening) under LVAD support, which can be useful

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in the design and evaluation of physiological pump control algorithms to ensure patient safety and comfort, as well as evaluating aortic valve pathologies.

TABLE I
MODEL PARAMETERS

| Parameter | Value |
|---|---|
| Blood density (kg/m$^3$) | $1.06 \times 10^3$ |
| Blood viscosity (Pa.s) | $2.70 \times 10^{-3}$ |
| Leaflet density (kg/m$^3$) | $1.06 \times 10^3$ |
| Aortic leaflet Young's modulus (Pa) | $1.0 \times 10^6$ |
| Leaflet Poisson's ratio | 0.49 |

## II. METHODS

### A. Model description

A two-dimensional FSI model of aortic valve dynamics was implemented using COMSOL Multiphysics (COMSOL AB, Sweden, Version 4.3a), using the ALE method first proposed by Donea *et al.* [18]. ALE is most commonly used in aerospace Engineering for simulating the interaction between moving objects and fluid flow. However, in this paper, ALE is used to apply forces exerted on the deforming structure (aortic valve) due to the flow of fluid (blood). This method includes a continuous adaptation of the mesh without modifying the mesh topology, and is used to describe the fluid using an Eulerian (i.e. moving coordinate system) formulation, and the structure in a Lagrangian (i.e fixed coordinate system) formulation.

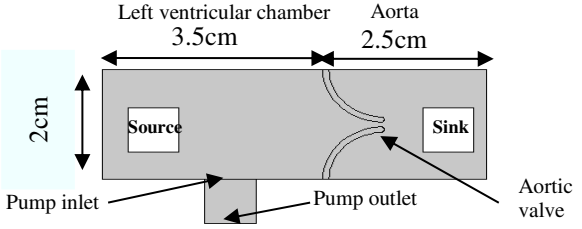


Figure. 1. Two-dimensional representation of the aortic valve and pump cannula.

The model geometry consists of a horizontal flow channel represent the ventricle, and narrow curved structures represent the AV leaflets. The aortic valve consists of two flexible leaflets of length 1.25 cm and thickness 0.1 cm. A thin-walled cannula of width 0.8 cm and length 0.8 cm is inserted into the ventricular chamber wall. The model also included blood outflow from the LV which was produced from inlet boundaries placed at the left end of the channel; aortic outflow was modeled as outlet boundaries as illustrated in figure 1. In this study our objective was to analyze the motion of the aortic valve in a highly-simplified model of the left ventricle, without taking into account the effects of heart wall contraction.

### B. Model Equations

We characterise blood pulsatile flow as being laminar, Newtonian, viscous and incompressible. The fluid is described by the Navier-Stokes equation (1) and the mass continuity equation for incompressible flow (2).

$$\rho\left(\frac{\partial \boldsymbol{u}_f}{\partial t} + (\boldsymbol{u}_{f-}\boldsymbol{u}_m\right) - \nabla \boldsymbol{u}_f) = \nabla.[-p\mathbf{I} + \mu(\nabla \boldsymbol{u}_f + (\nabla \boldsymbol{u}_f)^T] \quad (1)$$

$$\rho \nabla . \boldsymbol{u}_f = 0 \quad (2)$$

where $\rho$ is the fluid density; $\boldsymbol{u}_f$ is the velocity of the fluid in the fixed Lagrangian coordinate system, $\boldsymbol{u}_m$ is the velocity of the fluid in the moving Eulerian coordinate system, $p$ is the pressure and $\mathbf{I}$ is the unit tensor.

The fluid flows into the model from the left source (inlet) boundaries. At this entrance, the flow is assumed to have fully developed a laminar profile, changing with time as described in the next section. The valves are modeled as a linear elastic material, which was formulated as an isotropic Hookean elastic solid expressed using Einsteinian indicial notation as:

$$\varepsilon_{ij} = \frac{1+v}{E}\sigma_{ij} - \frac{v}{E}\sigma_{\alpha\alpha}\delta_{ij} \quad (3)$$

where $E$ is the Young's modulus, $v$ is Poisson's ratio, $\varepsilon, \sigma$ are the strain and stress respectively, and $\delta_{ij}$ represents of the Kronecker-delta tensor such that:

$$\delta_{ij} = \begin{cases} 0 & i \neq j \\ 1 & i = j \end{cases}$$

### C. Boundary Conditions

The fluid flow boundary conditions of the model included the following: laminar inflow conditions were applied to simulate aortic valve opening during systole and valve closure during diastole; a sinusoidal velocity pattern with a period of 1Hz and amplitude 0.12 m/s was applied at the inlet boundaries as shown in fig. 2a, 0 mmHg pressure was specified at the sink outlet boundaries, 0.014 m/s average velocity was applied at the outflow of the pump cannula, the fixed walls of the model were set to be no-slip boundaries, and the valve leaflets were assigned a 'moving-wall' boundary condition, whereby the velocity the fluid normal to these leaflet boundaries was set equal to the velocity of the moving wall and the tangential velocity was set to zero. Finally, the stress on the valve leaflets was set to equal to the fluid stress (pressure plus viscous stress), with the leaflet root boundaries set to a fixed displacement of zero.

### D. Computational settings

The FSI simulations were performed on a computational workstation using a 64-bit windows platform with 3.20GHz processor utilizing an Intel Core i7-3930K processor, with an applicable memory allocation of 32 GB. The fully-meshed model exhibited approximately 15,000 degrees of freedom. opened.

## III. Results

We observed that our model behavior could effectively be broken up into two phases: aortic valve closed and open, as shown from the outlet pump pressure in fig. 2b The opening phase was characterized by the formation of pressure waves travelling between the source boundaries and the pump outlet during the early stage of filling. Moreover, the pumping action kept the aortic valve pressures lower. However, during the second phase, this pressure dropped when the aortic valve closed by t=1.467s resulting in open phase duration of 0.467s.

Fig. 2b shows the development of the outlet pump pressure pattern when the pump was turned on. A low fluid velocity was observed at the source boundaries during the closing phase (t=0.4s), rising sharply and accelerating while the aortic valve was in its closed state (t=0.4-0.675s), reaching its maximum value of 0.025m/s at t=0.675 s.

It can be observed that during this period, blood moves toward the pump inlet due to the suction created by the pump. Meanwhile, the pressure at the pump outlet increases before reaching its maximum value of 185.8 Pa at t=0.85s as illustrated in fig. 2b. However, in this phase, the aortic valve is closed. During the AV opening phase, starting at t=1s, the pump pressure starts to decrease gradually (Fig. 2b). The valve leaflets, as observed from fig. 3, are fully opened at t=1s before starting to close again. The leaflets are fully

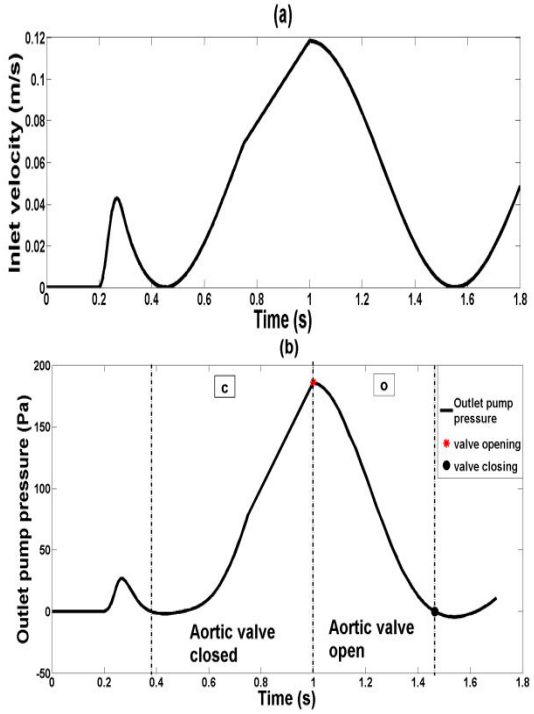


Figure. 2: (a) Sinusoidal velocity profile applied at the inlet to simulate diastolic inflow. (b) Simulated pressure at pump outlet over a single 1 s cardiac cycle, where ' * ' and ' ● ' indicate aortic valve opening and closing times, respectively. Label 'c' represents the period during which the aortic valve was closed, and 'o' is the period in which the valve was open.

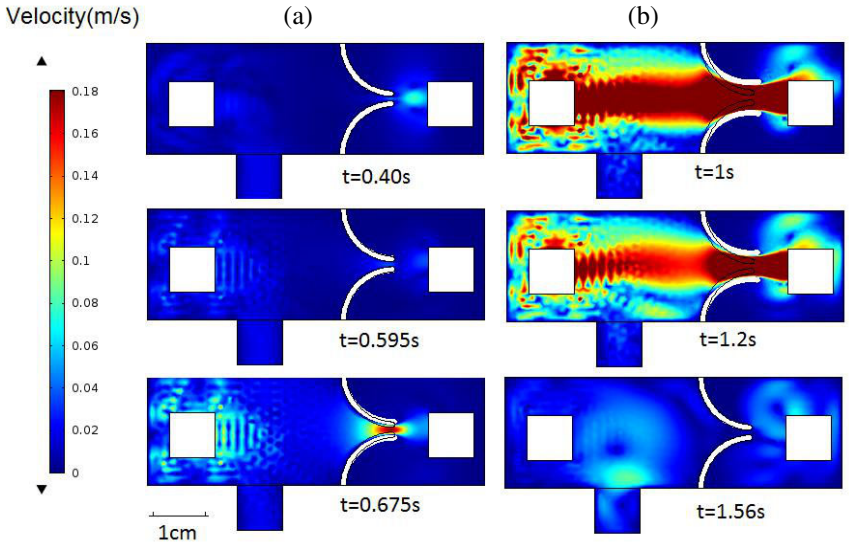


Figure. 3. Snapshots of simulated LV blood velocity magnitude during LVAD support at various phases during the cardiac cycle. (a) Aortic valve at closing phase. (b) Aortic valve at opening phase.

## IV. Discussion

We have presented a two-dimensional FSI model of the aortic valve motion, based on the ALE method. The simulations of this study confirmed that when the AV opens, pressure falls at LVAD inlet. This result is similar to that obtained by Shi *et al*. [19]. However in there model, they did not incorporate any computational fluid dynamics to determine pressure and velocity distribution during LVAD support, ignoring the aortic valve interaction between the blood and moving solid structures. Our simulated pressure variation during AV movement, particularly during its closure, will provide significant insight into LV function under LVAD assistance. Moreover, our simulations offer the potential of improving the accuracy of current LVAD control systems to ensure patient safety and comfort.

## V. Conclusion

Previous FSI models of ventricular function with LVAD support have ignored the impact of AV closure on blood dynamics in the left ventricle, as presented in this paper. Blood flow in a two-dimensional model of the left ventricle with aortic valve was simulated using a finite element model. This model will be used to investigate heart aortic valve closure during LVAD support. Future work includes modeling realistic ventricular geometry with ventricular geometry, as well as adding the contractile activity of the LV in order to simulate an entire cardiac cycle. To date, few studies which considered valve-blood interaction in addition to ventricular wall-blood- interaction have incorporated LVAD pumping action.

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