Simulation of closed-loop flow in a ventricular assist device coupled with a circulatory system model

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Abstract

This work extends previous open-loop simulations of an idealized, diaphragm-type ventricular assist device by coupling it in a closed loop to a three-dimensional model of the systemic circulatory system. The successes, challenges, and limitations of this model are discussed.

Keywords: Ventricular assist device; VAD; Fluid–structure interaction; Blood flow; Computational fluid dynamics; ADINA; Circulatory system; Closed loop

1. Introduction

Ventricular assist devices (VADs) are medical devices that are connected to failing hearts to help them pump a sufficient amount of blood through the body. Many different types of VADs have been built or are in development. This research is focused on an idealized diaphragm-type VAD design related to a model used at Brunel University, for which some experimental and numerical results are available for comparisons [1–3]. The current VAD model incorporates interaction of the diaphragm with blood flow and is coupled to a three-dimensional representation of the systemic circulatory system in a closed loop.

2. Model

The present VAD model consists of two hemispherical fluid chambers, separated by a flexible circular diaphragm, as shown in Fig. 1. The blood chamber design is the same as that in the Brunel University VAD [3], while the driving fluid chamber was designed to complement the blood chamber. The blood chamber was designed to be twice as large as an actual VAD, with diameters of 140 mm for the hemisphere and 38 mm for the inlet/outlet tubes. A major improvement of the current model over the Brunel University model is the use of a fully interactive flexible diaphragm rather than a rigid hemispherical piston with prescribed motion. In a previous study, simulations of blood flow, with the VAD model operated in open loop, were partly validated versus available experimental results [4]. The present simulations extend this work by connecting the VAD in closed loop to the circulatory system model described below.

The closed-loop model consists of the previously described VAD coupled to a three-component representation of the systemic circulatory system, as shown in Fig. 1. The circulatory system model consists of two elastic tubes, which simulate the total arterial and venous compliances, respectively, and an isotropic porous medium, which simulates the total peripheral resistance.

Simulations were performed using the commercial finite element package ADINA version 8.1. A slightly

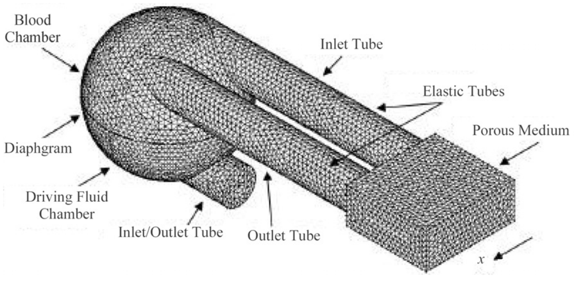

Fig. 1. Closed-loop geometry.

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compressible fluid model was used for blood and our driving fluid, water, due to the difficulty in satisfying a conservation of volume inside a closed system with moving boundaries. The governing equations for the fluid model are the arbitrary-Lagrangian-Eulerian (ALE) forms of the continuity and Navier–Stokes equations for slightly compressible fluids, defined, respectively, as:

$$\frac{\rho}{\kappa}\left(\frac{\partial p}{\partial t}+(\boldsymbol{v}-\boldsymbol{w})\cdot\nabla p\right)+\rho_{\mathrm{m}}\nabla\cdot\boldsymbol{v}=0 \quad (1)$$

$$\rho\frac{\partial \boldsymbol{v}}{\partial t}+\rho(\boldsymbol{v}-\boldsymbol{w})\cdot\nabla\boldsymbol{v}-\nabla\cdot\boldsymbol{\tau}=\boldsymbol{f}^{\mathrm{B}}, \quad (2)$$

where ρ is density, κ is the fluid bulk modulus, p is pressure, t is time, $\boldsymbol{v}$ is the fluid velocity vector, $\boldsymbol{w}$ is the mesh velocity vector, ρ_{m} is a compressible fluid density, defined as $\rho_{\mathrm{m}}=\rho\left(1+\frac{p}{\kappa}\right)$, $\boldsymbol{\tau}$ is the stress tensor, and $\boldsymbol{f}^{\mathrm{B}}$ is the body force per unit volume.

Flow in the porous medium is calculated using the following form of Darcy's law:

$$\mu\boldsymbol{k}^{-1}\cdot\bar{\boldsymbol{v}}=-\nabla p+\boldsymbol{f}^{B} \quad (3)$$

where μ is the fluid viscosity, $\boldsymbol{k}$ is the isotropic permeability tensor, and $\bar{\boldsymbol{v}}$ is the superficial velocity, which is the average velocity over the entire cross-sectional area of the porous medium.

The density and viscosity of water were taken as $\rho = 999\,\mathrm{kg/m^3}$ and $\mu = 1.00\times10^{-3}\,\mathrm{kg/ms}$, with the bulk modulus of water chosen as $\kappa = 1.00\times10^{10}\,\mathrm{kN/m^2}$ to limit the water compressibility. For blood, $\rho = 1105\,\mathrm{kg/m^3}$, $\mu = 4.66\times10^{-3}\,\mathrm{kg/ms}$, and $\kappa = 1.00\times10^{5}\,\mathrm{kN/m^2}$. For the porous medium, the permeability was set to a uniform value of $1.00\times10^{-4}\,\mathrm{m^2}$.

The diaphragm and the two elastic tubes are modeled as linear elastic materials with large displacements and small strains. All three solids are modeled as nearly incompressible with a Poisson ratio of 0.495 and a density of $1000\,\mathrm{kg/m^3}$. For the diaphragm, the Young's modulus is $E_{\mathrm{d}} = 1.00\times10^{8}\,\mathrm{kN/m^2}$ and for the two elastic tubes, it is $E_{\mathrm{t}} = 7.50\times10^{4}\,\mathrm{kN/m^2}$.

In ADINA, the fluid and solid models are defined separately and coupled during the solution process through coincident fluid–structure interaction (FSI) boundaries. The solution to this FSI problem was obtained using an iterative solver.

Flow in the system is driven by a sinusoidal pressure function applied at the entrance of the driving fluid chamber with a maximum amplitude of 215 MPa. This pressure forces water into and out of the driving fluid chamber, causing the diaphragm to deform back and forth and forcing flow around the loop. The reason that such high pressures were used was that at low values of the Young's modulus of the diaphragm, the solution did not converge. While increasing Young's modulus, it became necessary to also increase the driving fluid pressure in order to achieve the desirable diaphragm deformation.

Flow direction is controlled by instantly opening and closing valves at the interfaces between the VAD and the circulatory system model. The fluid geometry is meshed using four-node tetrahedral mini-elements, and the solid geometry is meshed using nine-node mixed interpolation of tensorial components (MITC) shell elements.

3. Results

Representative results of the simulations are presented in Fig. 2 at the two times during the cycle at which the diaphragm has reached its extreme deformations. It can be seen that as the diaphragm deforms into the blood chamber, the walls of the elastic tubes expand, and as the diaphragm deforms into the driving fluid chamber, the walls contract. The closed-loop results for the flow inside the blood chamber have been compared with our previous open-loop results [4]. Fig. 3a shows velocity magnitude variations at the center of the inlet of the blood chamber during one cycle, and Fig. 3b shows the variations of the x-velocity at a point inside the blood chamber. At both of these locations, the closed-loop model predicts much lower blood velocities than does the open-loop model. This difference is due to blood compressibility, and not differences in diaphragm displacement, which, in the closed-loop case, was at most only about 3% lower than that in the open-loop case.

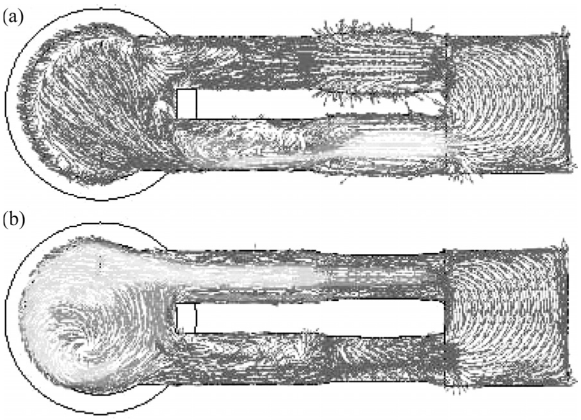

Fig. 2. Velocity vectors at (a) end of ejection and (b) end of filling.

It must be noted that in both the open- and closed-loop cases, the driving pressure used was the same,

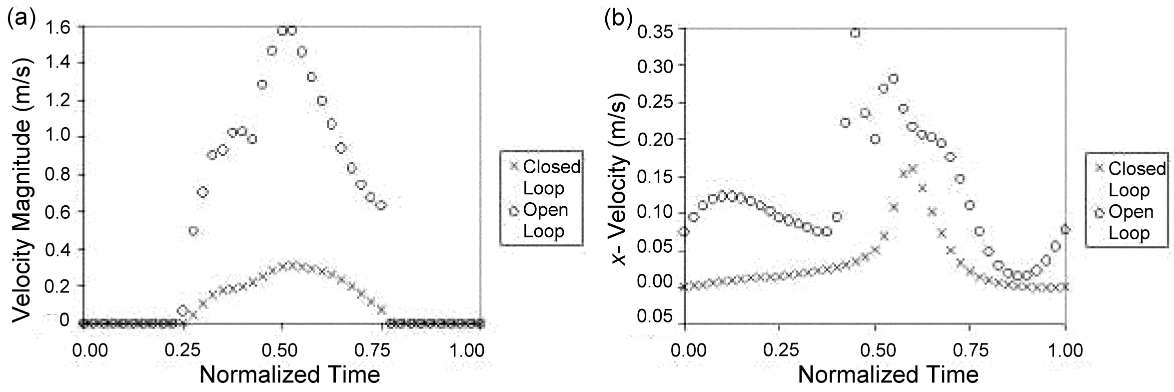

Fig. 3. Comparison of open- and closed-loop results. (a) Velocity magnitudes at the blood chamber inlet; (b) x-velocities at a point inside the blood chamber.

namely sinusoidal with an amplitude of 215 MPa. However, while in the open-loop case the pressure in the VAD was of the order of 1 kPa, in the closed-loop case it was much higher, of the order of 10 MPa. This pressure, although sufficiently smaller than the driving fluid pressure to allow appreciable diaphragm deformation, was high enough to generate large compression of 'blood', thus reducing its flow in the VAD.

Another consequence of using the slightly compressible fluid model is that the elastic tubes could not be calibrated to the total arterial and venous compliance values found in the literature. To achieve the desired motion of the elastic walls, the Young's modulus of elasticity, E, of the walls was set to be less than the bulk modulus of the fluid, κ. If E was set to be greater than κ, then the fluid would have compressed rather than forcing the walls to expand. In addition, the resistance of the porous medium was set to be much larger than the total peripheral resistance in the circulatory system in order to get a decrease in velocity through the medium. Attempts to increase further the value of κ in order to reduce the fluid compressibility, as well as to reduce the magnitudes of the input pressure function and the Young's modulus of the diaphragm, were unsuccessful, as the solution did not converge.

4. Conclusions

Velocity values were found to be much lower in the closed-loop simulations than in previous open-loop simulations. This decrease in velocity is due to the much higher pressure magnitudes inside the loop, leading to compression of the blood. This slightly compressible fluid model limits the ability to obtain an accurate solution with a closed-loop model. To improve the quality of the solution with this model, a method must be found to relax the slightly compressible fluid requirement.

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