Computer simulation of the blood flow in a planar configuration for a pulsatile ventricular assist device

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Abstract—**Patients that suffer congestive cardiac insufficiency need to be assisted with assist devices (VAD) as a temporal solution when the cardiac transplantation is not possible. For that reason it is of great interest to know the performance of such devices, especially the potential blood damage that they can produce. In this work, a computational two-dimensional simulation of a novel pulsatile VAD is performed. This new design of VAD has a double acting piston driven electromagnetically at 2.1 Hz. In addition, the VAD has four active valves. The flow velocities, the pressure and the shear stress developed in the blood are analyzed in the chambers and inflow and outflow conduits. The results suggest that the developed flow would not be dangerous for the blood.**

Keywords —**ventricular assist device, blood damage, finite elements, pulsatile flow**

I. Introduction

When a person has cardiac insufficiency your heart cannot provide an adequate blood flow, even under basal conditions (5 l/min and 100 mmHg). Therefore, the patient should be treated with drug therapy, mechanical circulatory assist (MCA) techniques [1], or via a heart transplant. It is well known the shortage of donors for the quantity of people requiring a transplant, thus MCA techniques become very important, either as a temporary solution until the transplant can be done or as therapies to recover the normal cardiac function.

Currently one of the methods of MCA is by using implantable pumps (VADs) connected in parallel with the heart [2,3]. These devices are being used because they provide greater life expectancy and life quality when compared with extracorporeal VADs. One of the most critical points in their design and development is to keep the blood damage at acceptable levels. Devices should cause the least possible levels of hemolysis and platelet activation to the blood [2,3,4].

Hence, the aim of this work is to simulate and to evaluate a new concept of pulsatile VAD operating at frequencies higher than physiological values; this make possible it can be miniaturized. With this purpose, a 2D computational simulation of the VAD is performed and the calculated flow variables (velocities and pressure) are used to estimate the damage on the platelets via the magnitude of shear stress and the time history it acts on these cells.

II. Methodology

A. Description of the simulated VAD

The simulated VAD is a double action volumetric pump. It consists of a piston which is supposed driven via electromagnetic forces (not simulated here), a left (LC) and right (RC) chambers, two inlet (Vi) and two output (Vo) valves and communication channels, as depicted in Figure 1. The piston displaces horizontally with sinusoidal velocity and each career is used to fill one chamber and to propel the fluid in the other (in alternating way). Between the walls and the piston there is a small gap about 100 μm, which is supposed to be maintained electromagnetically. The opening and closing of the valves is performed actively and synchronously with the piston displacement, to allow the fill of a chamber while the other is being emptied in the same piston career. Additionally, this allowing close a valve and then open their antagonistic at the time the piston begins to pump blood, this delay ensures that the two valves of the same camera never are open.

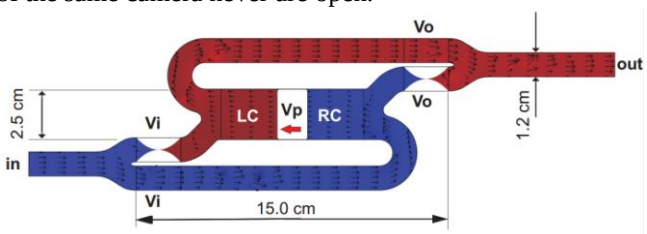

Fig. 1 Description VAD with input (in) and output (out) regions, the piston moving to the left and some representative dimensions.

B. The flow model

A 2D geometry, see Figure 1, which corresponds to a simplification of the 3D case is considered. By considering the blood as an incompressible and Newtonian fluid [5,6], the flow can be modelled by the continuity and Navier-Stokes equations. The set of equations are solved with the finite elements method, but using an appropriate coordinates mapping to model the boundary displacement and the mesh deformation, as described in the next section.

a) Moving domain and mesh deformation:

The equations (1) and (2) set the harmonic motion imposed to the piston, *Xp* and *Vp* being the displacement and velocity, respectively and where *A*= 2.0 cm is half the piston stroke and *f* = 2.1 Hz. This motion produces a deformation of the flow domain and to capture it, an Arbitrary Langrangian-Eulerian (ALE) and deformable mesh techniques are implemented [5]. In addition, the opening and closing of the valves are simulatedby imposing displacements to the respective boundaries but preventing the collapse of the mesh when the gap reaches it minimum thickness. This is similar to the method used in [6] but here a phase shift between the valve motions is introduced to avoid that two valves remain simultaneously opened in the same chamber. The closing time of the valves is set in 25 ms.

$$Xp = A\,\mathrm{sen}(2\pi f t) \quad (1)$$

$$Vp = 2\pi f A\cos(2\pi f t) \quad (1)$$

b) Boundary conditions:

On fixed and moving solid surfaces, no-slip and impermeability boundary conditions are considered. Thus, $\boldsymbol{v}$=0 on the chamber and ducts walls, $\boldsymbol{v}$=*Vp* on the piston wall and $\boldsymbol{v}$=*Vvalve* on the valves boundaries (which is obtained of deriving the functions of the position on the walls of the valves, as in [6]). On the input and output sections, reference pressures are imposed: Pin = 0 and Pout = 13.3 kPa (~ 100 mmHg), which approaches the mean aortic pressure. Furthermore, constant values are assumed for the viscosity and blood density: $1.060\text{x}10^{3}$ kg/m^{3} and $3.5\text{x}10^{-3}$ Pa s, respectively [4,6,8].

C. Numerical simulation:

The 2D geometry is discretized with a mesh of approximately 220,000 triangular elements of P3-P2 type, with sizes between $2\text{x}10^{-4}$ m and $5\text{x}10^{-3}$m. These values show to be suitable according to the performed mesh refinement tests. The simulation interval is 1.0 s which corresponds to 2.1 cycles of the piston motion (starting from the centre to the right) with maximum time-steps of $5\text{x}10^{-4}$ s.

The model was solved with the software COMSOL Multiphysics 4.4 running on a PC with an Intel Core i5 2500K 3.3 GHz processor and 16 GB of RAM. A fully-implicit, fully coupled Newton's method is chosen to linearize the equations and a direct solver (PARDISO) is used to solve the resulting linear equation system in each iteration. The total simulation time is about 96 hours. A standard Galerkin mixed formulation is used, with SUPG stabilization. Also, to start the simulation a higher viscosity is used which quickly assumed its desired value, while velocities and pressure are initialized in zero. Therefore, the first half cycle is not considered for the analysis.

D. Model of blood damage

Red blood cell (RBC) tolerates shear stresses of the of 150-250 Pa order during intervals of 100 s, whilst the platelets (Pl) tolerate approximately 20 Pa in the same time interval [7]. Therefore, platelet activation (PA) is the most relevant variable to indicate blood damage.

The modelling of the PA phenomenon is complex and depends on physical, chemical and biological factors. By considering only the physical aspects, several studies [3,4,6,8] coincide that the platelet activation state (PAS) could be predicted using a model based on the rate at which the shear stress is applied. These models should consider, also, the history of shear stress acting on the cells. Thus, in this work the model proposed Nobili et al. [8,9] was adopted. For its application, the path of a set of Pl flowing in the domain must be known. For that, it is supposed that the Pl moves like virtual mass-less particles and its trajectory is calculated by integration of the velocity field. Then, the viscous stress over the path (τ) is evaluated for each Pl (particle) by using the equation (3) [10]. Finally, the shear stress history in used in the equation (4) to evaluate the quantity PAS_n, that is the PAS for each Pl, where the constants *a*, *b* and *C* were extracted from [8]. A global quantity can be computed by taking the average of PAS_n, *PAS mean*, over a set of Pl released at the same time in a given region. In this work, two groups of 20 virtual particles representing the Pls were released from rest: one group at t=0.15 s to flow through RC and the other at t=0.39 s to flow through LC.

$$\tau = \frac{1}{2}\sqrt{\left(2\mu\frac{\partial u}{\partial x} - 2\mu\frac{\partial v}{\partial y}\right)^2 + 4\left[\mu\left(\frac{\partial u}{\partial y} + \frac{\partial v}{\partial x}\right)\right]^2} \quad (3)$$

$$PAS_n = \sum_{i=1}^{N} C\; a\left[\sum_{j=1}^{i}(\tau(t_j)^{\frac{b}{a}}\Delta t_j + D_o)\right]^{a-1}\tau(t_i)^{b/a}\Delta t_i \quad (4)$$

III. Results

A. Flow rate and velocity in the VAD

For the supposed operating conditions, the pump impels an average flow rate (2D) of 39 cm^{3}/(s cm), which is equivalent to 4.7 l/min, by assuming a width of 2.0 cm for the VAD. Figure 2 depicts the flow rate at the outlet section. The instants where the flow rate is negative are due to the volume swept by the valves during its motion, which is not negligible respect to the volume displaced by the piston. The result of the velocity field inside the VAD is shown in Figures 3 and 4, at the instant where the piston is moving with maximum speed toward the left and to the right, respectively. Speeds of the order of 3.00 m/s were found, and

at some cases the maximum speed reaches 6.50 m/s in the thin channels left by the valves when its antagonist is closing.

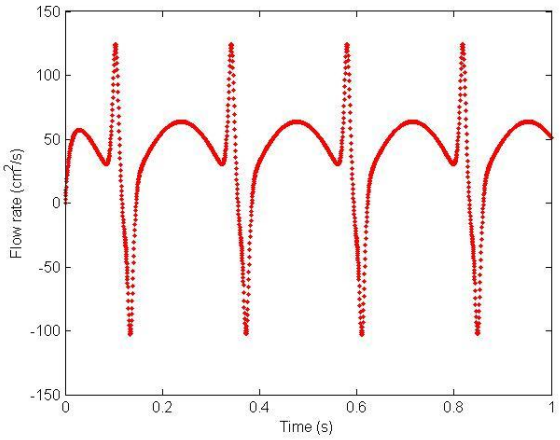

Fig. 2 Output flow rate.

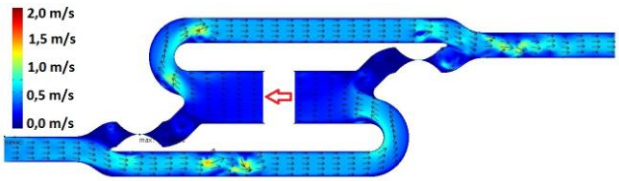

Fig. 3 Velocity field at 238 ms, with maximum Vp (0.27 m/s); at this instant the maximum speed registered in the fluid is 1.73 m/s.

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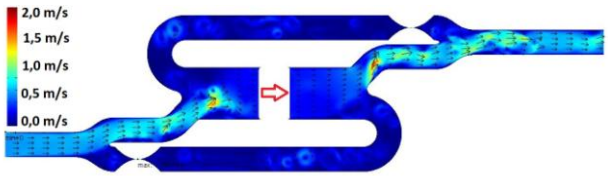

Fig. 4 Velocity field at 476 ms, with maximum Vp (0.27 m/s); at this instant the maximum speed registered in fluid is 1.71 m/s.

B. Pressure

The piston action increases the pressure in one chamber above Pout, it allowing the ejection of the fluid through the output; simultaneously, it decreases the pressure in the other chamber (below the zero reference value) to allow its filling. These effects can be observed in Figure 5, where the pressure inside the LC is depicted, while for the RC the behaviour is similar but shifted half of a cycle. It can be observed positive and negative peaks, around 25 kPa and -10 kPa respectively. Also, one detail of the pressure field can be observed in the Figure 1 for t = 238 ms; at this instant the pressure inside LC approximately equals Pout (red), while inside RC the pressure is close to Pin (blue). This corresponds with the instant where the piston reaches half of its career towards the left.

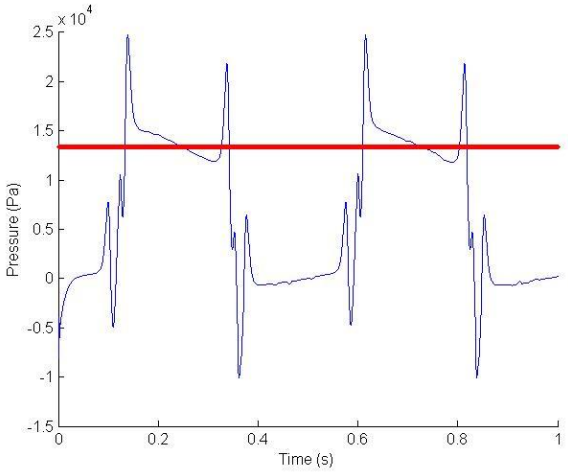

Fig. 5 Pressure (in blue) versus time (in seconds) inside the LC, evaluated at a corner of the piston. The red line indicates the imposed output pressure level: Pout = 13.3kPa.

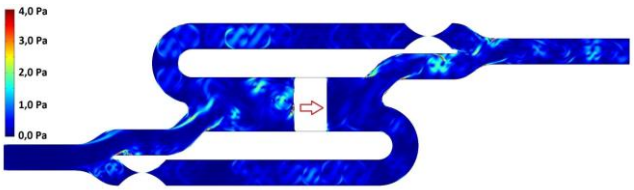

Fig. 6 Shear stress field at 835 ms, when the piston is moving to the right.

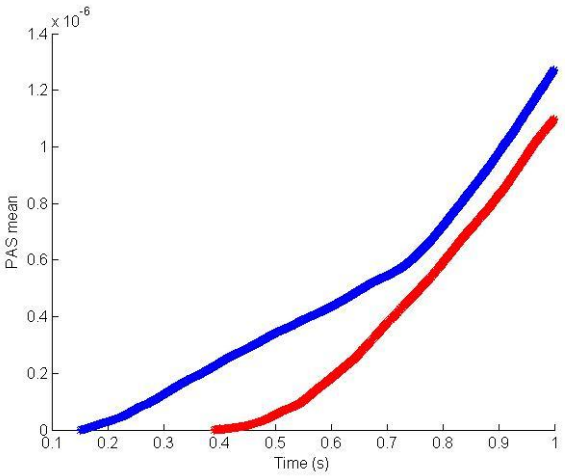

Fig.7 Average platelet activation state for the two groups of Pl: the blue line correspond to those released at 150 ms and the red at 390 ms.

C. Shear stress and platelet activation

As shown in Figure 6, the shear stresses are lower than 4.0 Pa in almost all the domain, except in valve borders, where the maximum stress is 80 Pa and at the curved walls of the channels, where the values are around 20 Pa when the maximum flow rate is reached. The average PAS, *PAS mean*, is obtained through the methodology described to II.D and it can be appreciated in Figure 7. It was evaluated for the two groups of Pls released, and can be noted that the value does not exceeded $2.0x10^{-6}$, which indicate that the platelet activation state has acceptable values [9].

IV. Discussion

The values calculated for pressure and flow rate allow inferring that the proposed VAD concept could have an acceptable performance, in terms of blood impulsion. While the pressure shows high peaks values, this is strongly associated with the valve motion, which has merit to be extensively studied in the future. Additionally, if negative pressure peaks occur, this will cause cavitation, which is undesirable as this causes blood damage.

In addition, the predicted shear stress values reaches tolerable levels for RBC and the maximum values that would be problematic for Pl are located in small areas of valves and in curved walls of the channel. For that reason, the average PAS shows acceptable values, lower than those obtained by Morbiducci et al. [9] for a mechanical heart valve, for example.

While these results are promising, 3D simulations of the proposed VAD are necessary to confirm the predictions, mainly to discard that the values of the shear stresses calculated with the 2D simulations were undervalued [10].

V. Conclusions

The results obtained in this work suggest that the proposed VAD design could sustain the blood flow rate necessary to supplement the function of a diseased heart. In these simulations, for example, it reaches around 4.7 l/min, with average platelet activation state values below the dangerous limits. However, high values of shear stresses are attained in some specific regions and they could indicate a source of cell damage. Nevertheless, these predictions should be corroborated by additional investigations, for example, it conducting 3D simulation analysis on more realistic geometries. These activities are under development.

Acknowledgements

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Conflict of interest

The authors declare no conflict of interest.

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