XIV International Conference on Computational Plasticity. Fundamentals and Applications
COMPLAS 2017
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PERFORMANCE EVALUATION OF A PULSATILE VENTRICULAR ASSIST DEVICE UNDER NON PHYSIOLOGIC PUMPING FREQUENCIES BY MEANS FEM AND 2D APPROACH.

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Key words: Ventricular assist device, blood damage, platelet activation, pulsatile flow.

Abstract. A ventricular assist device (VAD) is a blood pump that works in parallel with heart. It is used as a mechanical assistance for patients that suffer cardiac insufficiency: as a therapy, as a bridge to transplant or to extend life. The blood flow simulation into VAD is of great interest for the design and evaluation, mainly before building the prototypes. In previous works, by means of blood flow simulation, was evaluated a new concept of implantable VAD consisting on a pump with a double effect piston, driven without contact and four active valves. In this work, the flow into VAD is analyzed for four frequencies values: 1.05, 2.10, 3.15 and 4.20 Hz. The former is the physiologic frequency, the second allows the basal flow rate (5 l/min), while the others are higher in order to assure an increase in flow rates. The analysis is carried out comparing variables as velocity and pressure distribution into VAD and evaluating blood damage due to acting shear stress over cells. The blood flow simulation is performed on a 2D simplified geometry using COMSOL Multiphysics software to resolve Navier-Stokes and continuity equations, assuming blood as a Newtonian incompressible fluid. The blood damage is evaluated by means of platelet activation state index and a cumulative damage model. The global variables as flow rate, force and power to impel fluid, are shown in agreement with theoretical predictions. The risk of blood damage raises for higher frequencies, however, the predictions shown that the VAD analyzed is comparable and best to other VAD and mechanical heart valves.

1 INTRODUCTION

Heart failure (HF) affects heart avoiding the accomplishment of its function; this is currently one of the greater pandemics in the world [1], mainly in the occidental world. People suffer HF can be assisted with different treatments, mainly when the pathology is in advanced stages. A possible one for patients with HF is the mechanical circulatory assistance (MCA), which has been accepted as a therapeutic option, as a bridge to heart transplant or in cases where this is not possible; even more when the number of people affected by HF increases and the number of donors for a heart transplant (HT) relatively diminishes. However, HT is the more accepted option by the medical community [2].

Ventricular assist devices are pumps that are placed in parallel to the heart, usually pumping blood from the left ventricle into the aorta (LVAD). Implantables VAD are given more attention because they can be fully implanted and allow long-term care (from months to some years), as well as to improve the quality of life because they allow patients to do simple daily activities. A VAD is a therapeutic option to be used as a definitive solution, as a bridge to HT or, in some cases, as a treatment to recover the normal function of the damaged heart.

In previous works [3, 4, 5], using a numerical simulation of blood flow, a new concept of VAD was analyzed in a simplified way. This is a double-acting pump with a non-contact and external driven piston (i.e. electromagnetically) and four active valves. In this work, the flow in the same device for four operating frequencies *(f)* (1.05, 2.10, 3.15 and 4.20 Hz) by comparing the fields of velocity and pressure and the applied power, is studied. In addition, blood damage caused by the shear forces for $f=2.10$ *Hz* and $f=4.20$ *Hz* are compared. Blood flow is simulated into a simplified two-dimensional (2D) geometry using the COMSOL Multiphysics software, solving the Navier-Stokes and continuity equations, assuming blood as an incompressible and Newtonian fluid. Blood damage is assessed by calculating the platelet activation state (PAS) using a cumulative damage model [6, 7].

2 METODOLOGY

2.1 Description of the simulated VAD

Starting from the concept of a simple design volumetric pump and taking into account the functional VAD's characteristics, a two-dimensional (2D) approach is made. This approach involves the simulation of blood flow to investigate its possible damage. The simplified VAD geometry is presented in Figure 1; it has a piston, two chambers, right and left (LC and RC), input and exit ducts, two inlet valves (Vi) and two output valves (Vo). Piston without contact would demand high electromagnetic technology to be driven. The 2D geometry is selected to acquire preliminary knowledge using standard computational resources, as a guide for further investigation with more realistic geometries.

The double effect piston is assumed with periodical movement, in each run, piston pumps blood from one chamber and suctions it in the other, while two valves are open and two others are closed. The movement of each valve depends only of an actuator that controls the closing and reopening speed. The actuators move valves normally to flow in the input and exit conduits, they open and close in a few milliseconds assuming no secondary effects as rebound or vibration. The simulation of valves opening and closing, is done using moving meshes, with appropriate functions that allow a valve to be in synchronism with the piston movement, as is described in a previous work [5].

Figure 1 shows a Cartesian two-dimensional simplified diagram of VAD, where the piston is moving to the left, pumping blood from the left chamber to the outlet (red region), while the right chamber receives blood from the input (blue region). In this case, the upper Vi and lower Vo are closed to assure net flow rate. When the piston reaches the left end, the lower Vi and the upper Vo quickly close and the upper Vi and the lower Vo quickly open. The piston starts moving to the right, pumping blood from the CD to the exit and suction it from the inlet to the IC.

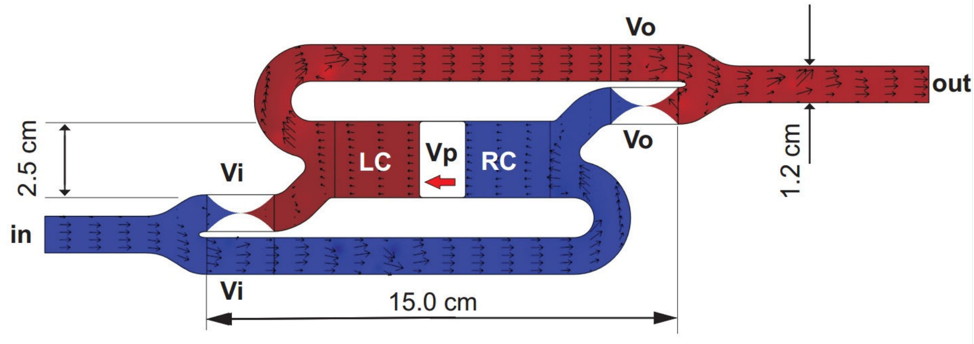


Figure 1: Description of VAD with input (in) and output (out) regions, the piston is moving to the left. There are some representative dimensions.

2.2 The model of blood flow

As was said, numerical simulations of blood flow are made in a simplified 2D plane geometry. Considering blood as an incompressible and Newtonian fluid [8], the Navier-Stokes and continuity equations with the governing law of motion for piston and valves movements, are solved using an appropriate coordinate mapping as described in the next section.

2.2.1 Moving domain and mesh deformation:

Equation 1 and 2 represent a simple harmonic motion (*Xp*), which is appropriate for this kind of pump, where *A=2.0 cm* is the amplitude in each chamber, *Vp* is the harmonic velocity of piston and *f* is the frequency of movement; the parameter of interest in this work. The movement of each valves, is implemented with ad hoc functions that avoid the collapse of the elements [3, 5]. This motion produces a deformation of the flow domain and to trace it, an Arbitrary Langrangian-Eulerian (ALE) deformable mesh technique is implemented [10] [9].

$$Xp = A\sin(2\pi f) \tag{1}$$

$$Vp = 2\pi f A\cos(2\pi f) \tag{2}$$

2.2.2 Boundary conditions for blood flow

On fixed and moving solid surfaces, no-slip 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 are imposed [5]. 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: 1.060x10^3 kg/m^3 and 3.5x10^{-3} Pa·s, respectively [3, 10, 7].

2.2.3 Numerical simulation

The 2D geometry is divided with a mesh of Ne (see table 1) triangular elements P3-P2 kind, whose size varies according to each adopted f. The maximum and minimum dimensions of elements where taken according to a suitable refinement test. The maximum time step is reduced as f increases. Table 1 shows these parameters for each of four simulations. The simulated time intervals are 1.2 s for 1.05 Hz, 1.0 s for 2.10 Hz, 0.5 s for 3.15 Hz and 0.5 s for 4.20 Hz. The simulation time interval is selected to complete at least one operating cycle to do possible the computation and comparison of blood damage.

The system of equations is solved simultaneously through a monolithic scheme. In order to start the simulation, a higher viscosity is used while all variables are initialized with zeros. Although the initial higher viscosity quickly descends to the assumed value, in all the simulations the first half-cycle is not considered for the analysis.

Table 1: Parameters for each simulation.

Frequency f [Hz]	1.05	2.10	3.15	4.20
Simulated time period [s]	1.20	1.00	0.50	0.50
Maximum time step [s]	1x10^{-3}	5x10^{-4}	4x10^{-4}	2x10^{-4}
Maximum element size [m]	5 x10^{-3}	5 x10^{-3}	5x10^{-3}	2.5 x10^{-3}
Total number of elements Ne (x10^3)	206	206	206	376

2.3 Global variables calculation

Since the DAV is a volumetric pump, the output flow is obtained by integrating the normal component of the velocity (v_x) in the output section as defined in equation 3. The pumping force and the pumping power are determined by equation 4 and 5, respectively. It should be noted that the area integral for the force, becomes line integral over the contour of the piston for the 2D geometry, where $t_x(\boldsymbol{n},t)$ is the vector of tension x component over piston surface.

$$q(t) = \iint v_x(t)\, dy \tag{3}$$

$$Fx(t) = \oint t_x(\boldsymbol{n},t)\ dl \tag{4}$$

$$Pwr(t) = Fx(t) * Vp \quad (5)$$

2.4 Model of blood damage

The platelet activation phenomenon modeling is a complex task. If considering only physical aspects, platelet activation state (*PAS*) can be predicted using a model based on the rate at which the shear stress is applied. This model should consider, also, the history of shear stress acting on the cells. Thus, in this work the model proposed by Nobili *et al.*[7] is adopted. For its application, the path of a set of PLs drifting in the flow domain must be known. For that purpose, it is supposed that PLs moves massless-like virtual particles and their pathes are computed by integration of velocity field. Then, the equivalent shear stress (τ) is evaluated for each PL (particle) path by equation 6 proposed by De Tulio *et al.*, [11]. Finally, the shear stress history is used in equation 7 to evaluate the PAS_n, that is the PAS for the *n*-PL, where the constants *a, b* and *C* are extracted from Nobili *et al.* [7]. A global quantity *PASmean* can be computed as the average of all PAS_n by means of equation 8, over a set of PLs released at the same time in a given region. In this work, four groups, each one composed for a set of N= 20 particles representing the PLs, are released. Four groups are considered for simulation, two groups corresponding to 2.10 Hz (one for each chamber) and, in the same way, two groups corresponding to 4.20 Hz, see table 2.

$$\tau = \frac{1}{2}\sqrt{\left(2\mu\frac{\partial u}{\partial x} - 2\mu\frac{\partial v}{\partial y}\right)^2 + 4\left[\mu\left(\frac{\partial u}{\partial y} + \frac{\partial v}{\partial x}\right)\right]^2} \quad (6)$$

$$PAS_n = \sum_{i=1}^{N} C\, a\left[\sum_{j=1}^{i}(\tau(t_j)^{\frac{b}{a}}\Delta t_j + D_o)\right]^{a-1} \tau(t_i)^{b/a}\Delta t_i \quad (7)$$

$$PASmean = \frac{1}{N}\sum_{n=1}^{N} PAS_n \quad (8)$$

Table 2: PLs groups for blood damage evaluation.

Groups	1	2	3	4
N - Number of Pls	20	20	20	20
Time initial [ms]	150 ms	150 ms	100 ms	100 ms
Frecuency [Hz]	2.10	2.10	4.20	4.20
Chamber	LC	RC	LC	RC
Initial position	Vi in LC	Vi in RC	Vi in LC	Vi in RC

The model used to evaluate the *PASmean* is the most used by the reserchers' community and is the most cited in literature [6, 7, 10, 5, 12]. It predicts the platelet activation because of shear stress from physiological or artificial source. This cumulative damage model is adapted to represent experimental situations of pulsating shear stress, in this case the model works

appropriately as described by Nobili *et al.*[7]. However, there are situations of high shear stress in which the model can not properly represent the sensitization of PL [12].
It is important to note that numerical simulation of platelet activation is an open study field and there is not an optimal model for all situations of blood flow (natural or artificial). The model used in this study was modified by Sheriff et al. (2013), incorporating new parameters to better represent blood damage in many flow situations. However, these researchers have pointed out model limitations to represent certain in vitro results due to the power law description (see equation 7).

3 RESULT

3.1 Flow rate and velocity

Figure 2 shows the output flow rate generated by the VAD operating at different frequencies. This flow rate is composed by: the flow rate pumps by the piston whose shape is a "sinusoidal rectified" function; the overflow rate induced for the closing of a respective valve (positive peak) and a backflow rate induced for the opening of the counterpart valve (negative peak). This flow rate is obtained integrating as is indicated by equation 3. On the other hand, in figure 3 is presented a comparison between the ideal flow rate for a cycle (without differential flow rates because valves), represented by the blue line, versus the average flow obtained in each simulation, indicated by red dots.

Figure 4 depicts the magnitude of velocity field when the piston moves to the left at the maximum speed, in the four images the scale is the same (0 to 3 m/s). When the operating f raises, the speed at certain points is greater, as happens at the valve gaps.

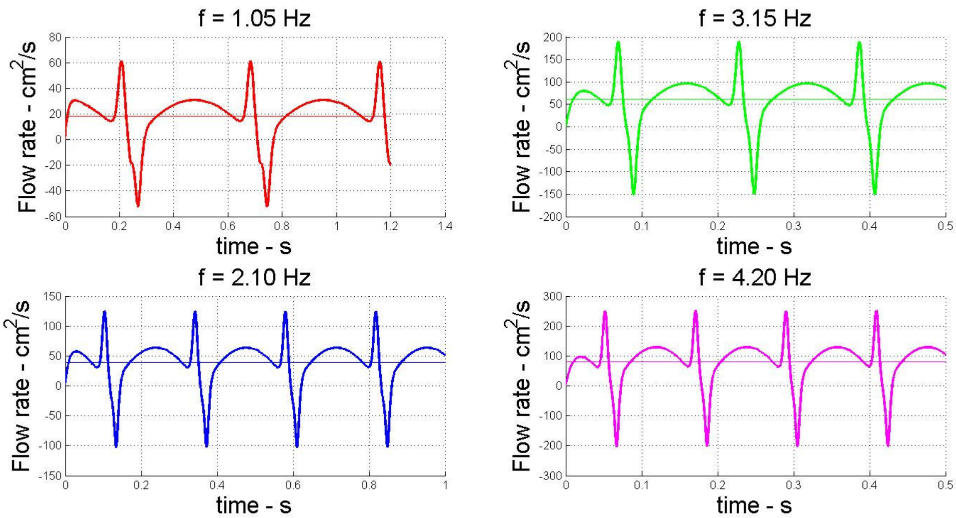


Figure 2: Flow rate for each frequency. The horizontal line indicates the average flow rate.

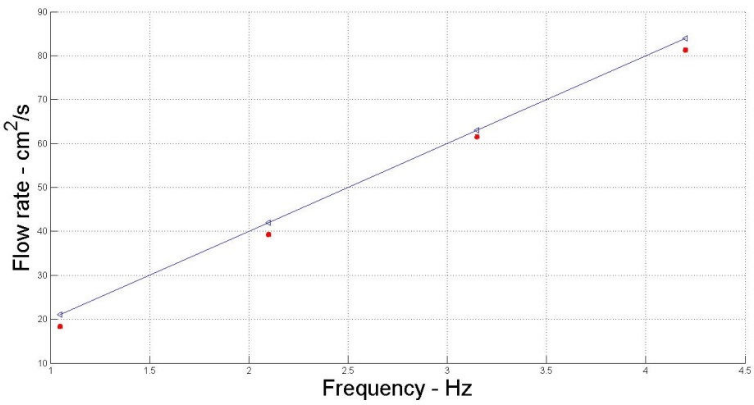
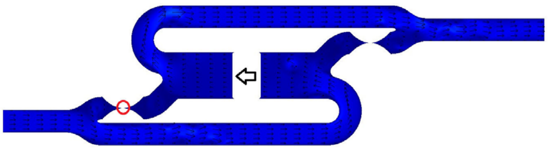
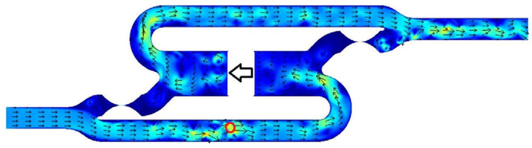


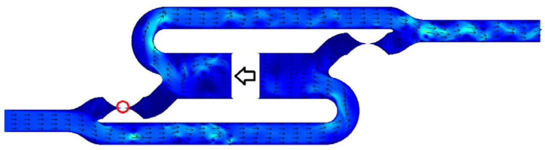
Figure 3: Average flow rate for each operating frequency (in red). The blue line indicates the ideal flow rate.



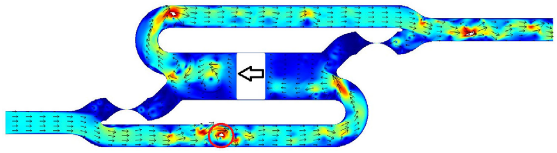
a) $f = 1.05$ Hz
Time in 0.476 s; $Vp = 0.13$ m/s and $Vmax = 1.69$ m/s



c) $f = 3.15$ Hz
Time in 0.476 s; $Vp = 0.39$ m/s and $Vmax = 2.2$ m/s



b) $f = 2.10$ Hz
Time in 0.358 s; $Vp = 0.27$ m/s and $Vmax = 1.72$ m/s



d) $f = 4.20$ Hz
Time in 0.358 s; $Vp = 0.52$ m/s and $Vmax = 3.4$ m/s

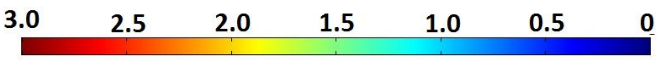


Figure 4: Velocity field module for the instant when the piston moves at the maximum speed to the left. The maximum velocity is indicated with a red circle, in the location of valve gaps for a) and b), and into the input branch for c) and d).

3.2 Pressure

Figure 5 shows the pressure distribution in LC, very close to the edge of the piston. It is observed that the mean pressure in each half-cycle is equal to the inlet pressure (0 Pa) or the outlet pressure (1.3×10^4 Pa). As frequency rises, the time derivative of pressure, raises too. In addition, the pressure peaks generated by the valves closing and opening are relatively higher because the closing and opening speed increases with frequency.

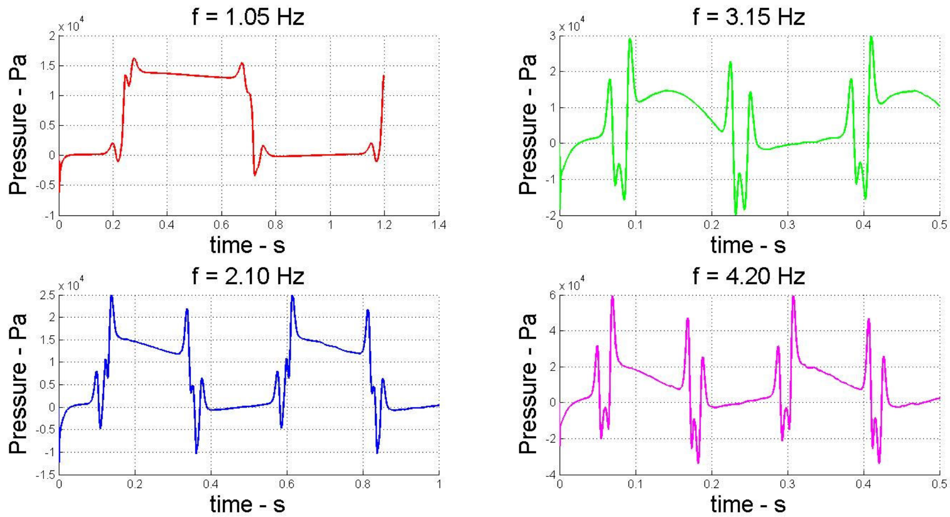


Figure 5: Pressure into the left chamber for each operating frequency.

3.3 Force and Power

The *Fx(t)* required to drive the fluid over time, is obtained by equation 4 and is presented in figure 6 for the frequencies analyzed. As the *f* increases, the time derivative of *Fx* is strongly modified. The instantaneous pumping power (*Pwr(t)*) is depicted in figure 7, where the average power for a cycle is indicated by a line for each frequency.

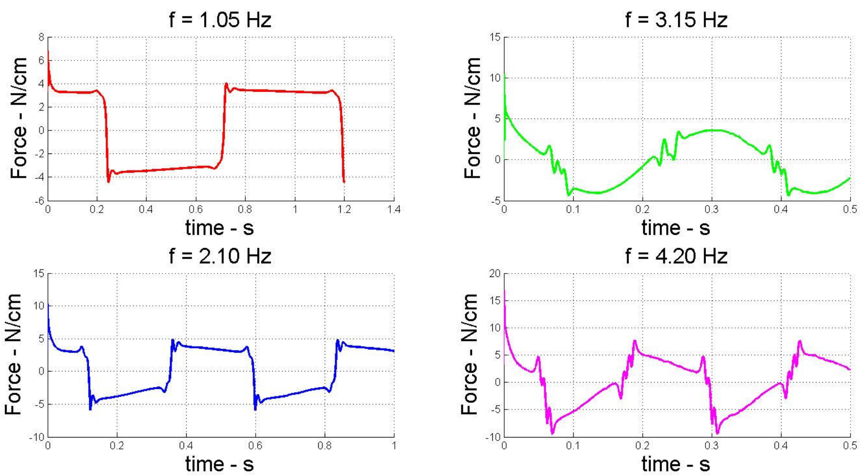


Figure 6: Force on fluid for each operating frequency.

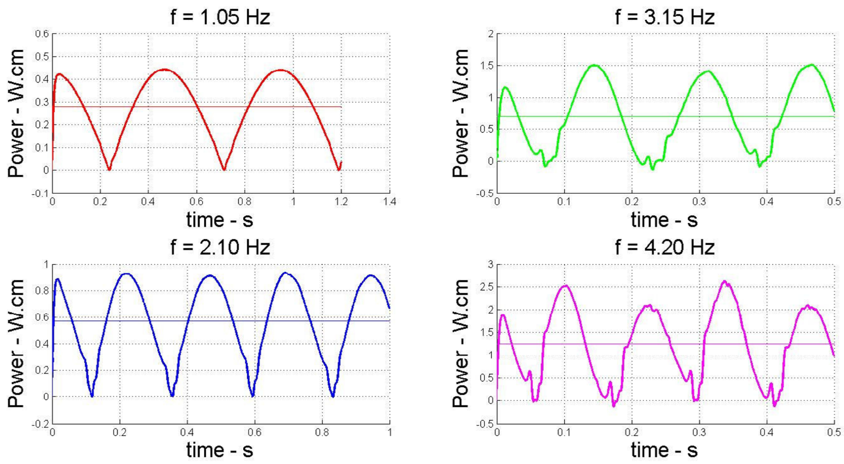


Figure 7: Power on fluid for each operating frequency.

3.4 Platelet paths and PAS mean

The platelet trajectories in the blood flow are of the kind of those show in Fig. 8, the trajectories in dashed line correspond to PLs for the DAV operating at 2.1 Hz released at LC (group 1) and released at RC (group 2) for red and blue color respectively. While the trajectories in continuous line correspond to PLs for the DAV operating at 4.2 Hz, released at LC (group 3) and released at RC (group 4) for red and blue color respectively.

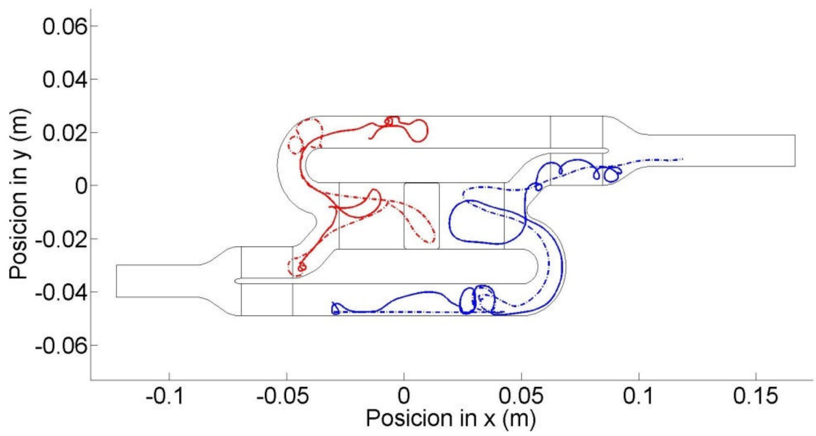


Figure 8: Some PLs paths correspond to examples of: group 1 (2.1 Hz) released at LC (red dashed line), group 2 (2.1 Hz) released at RC (blue dashed line), group 3 (4.2 Hz) released at LC (red solid line) and group 4 (4.2 Hz) released at RC (blue solid line). The *y* dimension has been shifted to better comprehension.

The calculated shear stress as indicated in equation 6 varies over time along of a PL path. Figure 9 shows the variation of the shear stress as a function of time, for the same trajectories presented in figure 8. It may be observed that the highest stress reaches 4.1 Pa for the trajectory of group 4. When evaluated the *PASn* by equation 7 in these trajectories, the functions that are depicted in figure 10 are obtained.

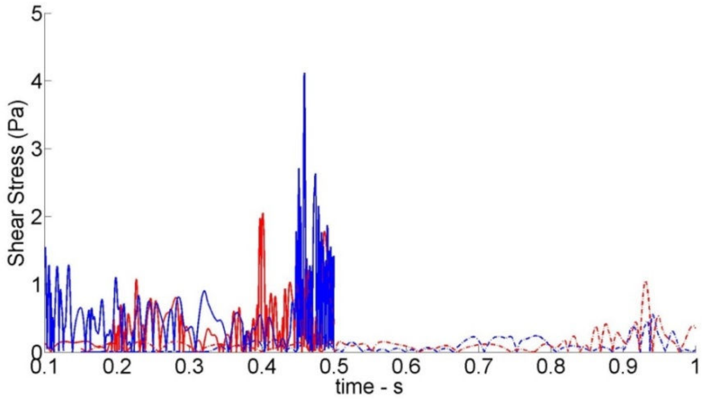


Figure 9: Shears stress for the PL paths corresponding to examples of figure 8.

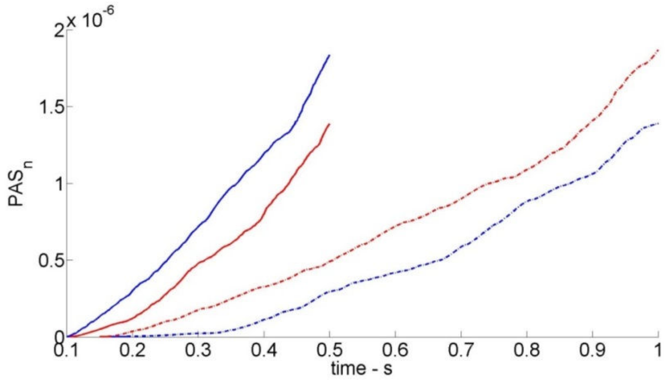


Figure 10: PAS_n for each path of the figure 8. Group 1: red dashed line, group 2: blue dashed line, group 3:red solid line and group 4: blue solid line.

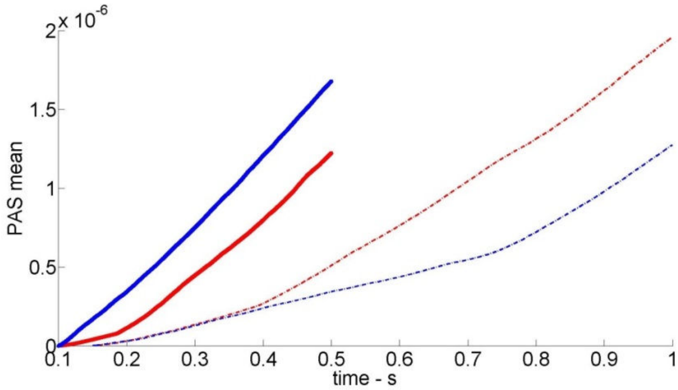


Figure 11: *PASmean* for the four groups. Group 1 (2.1 Hz) released at LC (red dashed line), group 2 (2.1 Hz) released at RC (blue dashed line), group 3 (4.2 Hz) released at LC (red solid line) and group 4 (4.2 Hz) released at RC (blue solid line).

Figure 11 shows the all *PASmeans* obtained. Groups 1 and 2 have practically the same slope of growth at the end of curves, which is 2.85×10^{-6} 1/s, while group 3 has a slope of 3.65×10^{-6} 1/s and group 4, 3.35×10^{-6} 1/s. These high slopes for group 3 and 4, imply a greater possibility of blood damage.

4 DISCUSSION

Fig. 3 shows that the VAD output flow rate is proportional to the frequency and is smaller than theoretical flow rate prediction. This occurs as a consequence of differential flow rate induced for the valve closing and reopening and losses for the piston-chamber wall clearance. Some unexpected results are obtained in the simulation of blood flow, those are the maximum velocities that appear inside the flow when driven frequencies are high (see fig. 4 c and d). This fact may be explained because the *Re* number has values of 900, 1900, 2800 to 3700 at the input and output conduits, for the four simulated frequencies. Thus, the laminar model in these cases of high frequencies may be inappropriate due to possible local turbulence, this may be resolved with more refined discretization and best computational resources.

As the piston movement f increases, higher velocities are developed within the flow, which leads to an increase in the shear stress on blood involving an increasing risk of blood damage. This fact can be verified by comparing the shear stresses for the simulation corresponding to $f = 2.10$ Hz (groups 1 and 2) versus the simulation corresponding to $f = 4.20$ Hz (groups 3 and 4), for practically the same number of cycles. Figure 9 shows the shear stresses are higher for groups 3 and 4, and it happens with faster variations in time (more abrupt peaks), in certain way, this is explained by the tortuous trajectories followed by particles (see figure 8).

In Figure 2, it can be seen that the output flow rate as the frequency rises, has instantaneous higher peaks (positives and negatives) due to the valve closing and reopening, that derive in pumping efficiency loss and increasing possibility of blood damage. On the other hand, figure 5 shows that pressure peaks increase more than twice; the positive peak generated by the Vo closure, goes from 2.5×10^4 Pa for $f = 2.10$ Hz to 6.0×10^4 Pa for $f = 4.20$ Hz, while the negative peak generated by the Vi opening goes from -1.0×10^4 Pa for $f = 2.10$ Hz to -3.0×10^4 Pa for $f = 4.20$ Hz. This high reduction of suction pressure may promote blood damage by flow cavitation, as is discussed in a previous work [1, 2, 6].

When the *PASmean* results, from 2.85×10^{-6} 1/s to 3.35×10^{-6} 1/s for group 4, are compared, all of these values are lower than the results obtained by Morbiducci et al. [10] (6.0×10^{-6} 1/s) by a simulation of 350 ms period of time for a mechanical cardiac valve (MHV). Therefore, even in the case of elevated frequency, VAD could cause less blood damage than an MHV.

It is important to note that predictions in this work, must be compared with predictions of future research. The need of a 3D realistic geometry arises, so as to make the model better for active valves and the piston driven without contact. A higher computational cost must be assumed.

5 CONCLUSION

A ventricular assist device consisting in a driven piston without contact and four active valves, pumping at four different frequencies: 1.05, 2.10, 3.15 and 4.20 Hz, considering a simplified plane 2D geometry and laminar flow conditions,was simulated. Global variables as output flow rate, force and power applied to the flow have practically linear dependence behavior with the driven frequency parameter.

The results obtained show that, the emerging blood damage due to flow conditions, as velocities, pressures and acting shear stress, are far from the risk values. In this sense, the mean platelet activation state prediction, could be of the same order of magnitude, and smaller than which evaluated over mechanical heart valves and other pulsatile devices.

It is important to emphasize that predictions must be corroborated by future researches, considering a 3D realistic geometry, and a flow model adapted to conditions of transition or turbulent flow, for high frequency pumping. However, the predictions communicated in this paper may be taken as preliminary to conduct new computational simulation or experimental analysis.

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