

A preliminary simulation for the development of an implantable pulsatile blood pump

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Abstract. A preliminary study of a new pulsatile pump that will work to a frequency greater than 1 Hz, is presented. The fluid-structure interaction between a Newtonian blood flow and a piston drive that moves with periodic speed is simulated. The mechanism is of double effect and has four valves, two at the input flow and two at the output flow; the valves are simulated with specified velocity of closing and reopening. The simulation is made with finite elements software named COMSOL Multiphysics 3.3 to resolve the flow in a preliminary planar configuration. The geometry is 2D to determine areas of high speeds and high shear stresses that can cause hemolysis and platelet aggregation. The opening and closing valves are modelled by solid structure interacting with flow, the rhythmic opening and closing are synchronized with the piston harmonic movement. The boundary conditions at the input and output areas are only normal traction with reference pressure. On the other hand, the fluid structure interactions are manifested due to the non-slip boundary conditions over the piston moving surfaces, moving valve contours and fix pump walls. The non-physiologic frequency pulsatile pump, from the viewpoint of fluid flow analysis, is predicted feasible and with characteristic of low hemolysis and low thrombogenesis, because the stress tension and resident time are smaller than the limit and the vortices are destroyed for the periodic flow.

Keywords: ventricular assist pump; blood flow; fluid structure interaction; finite element method; cardiac insufficiency

1. Introduction

The cardiovascular disease (CVD) is an important cause of morbidity and mortality in occidental world, and a significant public health problem in most of industrialized countries. Since 1900, the CVD has been the main reason of death in the United States of America (American Heart Association 2002) causing more than 950,000 deaths per year.

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The congestive cardiac insufficiency (CCI) is the most frequent kind of CVD. The USA statistics indicate that since the '90 almost 5 million of its inhabitants live with CCI (Bethesda 1996). For this reason, more than 40,000 million dollars are annually spent for the treatment of patients using different therapies developed in the last two decades. Even though, the CCI increased 145% between 1979 and 1999. The CCI is directly responsible for 30,000 to 40,000 deaths and, indirectly, of 250,000 other deaths each year. It is expected that this figures will increase as soon as more cardiac patients survive to the initial stages, increasing the development of the CCI terminal.

When the pharmacological and surgical therapies are not enough, the therapeutic alternative is cardiac substitution or the implant of an artificial heart. The cardiac substitution is an appropriate treatment for the advanced CCI. However, it has important limitations related to the selection of patients, ablation and organ destination and cost-benefit relation.

Each year, 25,000 patients with terminal CCI are subjected to cardiac substitution in USA, while other 4,000 are waiting. In 2001, 458 people died while they were waiting a donor. The cardiac substitution has a survival time that ranges between ten and twenty years associated to the continuous use of pharmacologic therapy, which is expensive and for lifelong. The candidates to transplantation are patients younger than 40.

With the aim to overcome these limitations, engineering and doctors have been working hard for four decades to develop systems able to guarantee mechanical circulatory support (MCS), temporal or permanently. Originally, these systems were created to give lifelong support assuming that other way of heart substitution wasn't possible. However, this objective was never reached. More recently, the temporal MCS has been shown as a viable option for patients with CCI that are waiting a cardiac substitution (Frazier *et al.* 2001) and for others that, although being candidates to substitution, need MCS for an established period of time (Rose *et al.* 2001).

Although recent clinical studies show, that in certain cases, the improvement of cardiac function allows the removal of the MCS system avoiding cardiac substitution (Mueller *et al.* 1999, Frazier *et al.* 2001), the development of an artificial heart, able to save lives of patients in critical stages of CCI and to allow them to return to normal activities, continues.

Nowadays and more than a decade ago, scientists are being developing axial flow blood pumps that should resolve the problem of size and they should make possible its implantation into patients with CCI. These devices are very simple, with a few moving parts, a small surface in contact with the blood and they also don't have valves, so they are not occlusive. Their easy insertion should simplify their removal. However, blood is moved at high velocities suffering high shear stresses in regions of narrow circulation channels that could cause hemolysis (red cells destruction) and platelet aggregation. On the other hand, the continuous flow isn't physiologic and for this reason, could develop other cardiovascular pathologies. Not to require valves in the system, makes that a mechanical failure be equivalent to a severe aortic insufficiency.

It is possible that the advantages of axial flow pumps, such as size and energy saving, will determine that its use might be generalized in clinical practice, and this fact allows to study the chronic consequences of a continuous flow (not pulsatile) in human body. The future of the research and development must be oriented to decrease the risks of thrombus, hemolysis associated to the artificial blood pumping and blood contact with non-human materials. A basic scheme of an implantable ventricular assist device is shown in Goldstein *et al.* 1998.

The pulsatile ventricular assist pumps are in general of big size and this fact generates difficulties for their implantation. However, if the devices are designed to supply a blood flow at a higher frequency than the physiologic case, they will lead more reduced size: should be totally

implantable, of low cost manufacturing, supply a flow like the physiologic one, be reliable and safer (Robbins *et al.* 2001).

The shapes and functioning principles of implantable ventricular assist pumps (IVAP) have been changing and evolving along the time to achieve designs that allow less invasive implantation. In this sense, heart analogous systems are the starting point, i.e. pulsatile, becoming the first generation of this kind of life support devices. Then centrifugal pumps were developed and nowadays axial pumps are developed and produced with very small rotors that could impulse 5 l/min flow rate with rotational velocities up to 8,000 rpm in certain cases. The second generation of centrifugal and axial flow pumps has the rotors supported by bearings; these are likely to wear failures and therefore bad functioning or device stopping. The IVAPs of third generation are centrifugal and axial with rotors supported totally by magnetic levitation. This kind of support avoids direct contact between components and minimizes wear; it is clear that the need of electromagnets adds complexity and raises the energy consumption (Bluestein *et al.* 2010).

In this work the fluid-structure interaction (FSI) between Newtonian blood flow and a driving piston that moves with a known periodical speed within a chamber, is analyzed. The mechanism is of double effect and has four valves; two for suction and two for impulsion (Robbins *et al.* 2001, Malchesky 2009). Through simulation by means of commercial software based on Finite Element Method, the flow is resolved in a preliminary planar geometry to determine regions of high velocities and high shear stresses that could promote hemolysis and platelets aggregation. The closing and opening of the valves are modelled by boundary conditions of non-slip over moving solid surfaces at prescribed velocities. Finally, the force at which the piston must be driven to achieve the pumping for a non-physiologic frequency (more than 1Hz) and the pumping power, are computed. It is noticed that this geometry has the third direction non defined, and, for that reason several quantities will be expressed per unit width. Also, the impeller must be named driver, but in the subsequent section we will refer to it as a piston not to lose the reference to finally 3D design of this pump.

2. Physical model

The pulsatile IVAPs require being of double effect so as to reduce their size and avoid regions where the blood might remain trapped with the risk of coagulation and infections. A double effect mechanism has been used by IVAPs based on the principle of deformable chambers, where a magnetically driven piston compresses one of the chambers while distends the other so as to prepare it for the next hemicycle of impulsion (Robbins *et al.* 2001).

The problem analyzed in this paper, is inspired in a pump design where a double effect piston actuates in direct contact with blood into two rigid chambers of cylindrical geometry. This piston will levitate magnetically without contact with solid pieces of the chambers, and will be driven alternatively by a linear electric motor. As the piston is of double effect, it will be impelling blood in one of the chambers while in the other it will be suctioning. Therefore, in each moment, there will be two valves open and two valves closed.

Because the flow into the chambers is tridimensional (3D), its simulation requires a great computational effort. Therefore, in this work a preliminary analysis of the induced flow by the mechanism in a simplify geometry, is made with the aim to view the flow in the two chamber with low computational cost. Fig. 1 shows the scheme of a pulsatile IVAP of double effect in a geometry that is non-defined in the third dimension (by which, the geometry is planar or 2D). Four

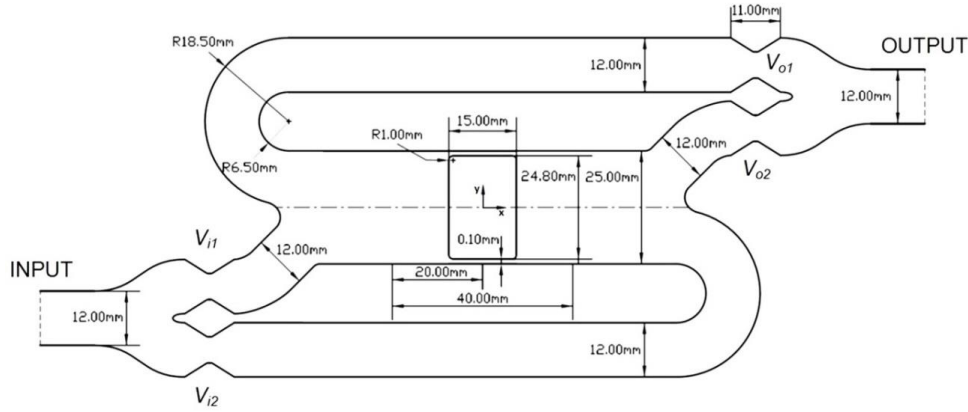


Fig. 1 Pulsatile double effect device geometry in a planar configuration for its simplified analysis

valves can be observed with the clearance between the piston and wall chambers (out of scale for best comprehension). An alternative flow from the impeller chamber to the suction chamber into the clearance, promotes low friction and energy saving.

The dimensions established in Fig. 1 belong to a device that will work in a higher frequency than the physiologic case, i.e. more than 1 Hz. This is the only possibility for a pulsatile impeller mechanism to be of implantable size; for this reason the frequency might be between 3 to 6 Hz. The valves of Fig. 1 open and close with a prescribed velocity with the same frequency of the entire device, because of this, valves should be electromechanical and their development is a future research direction.

3. Model equations

The preliminary analysis presented in this work, has the simulation of FSI between an alternative impelled fluid –blood- and the impeller structure, where there is a piston of known velocity and valves, acting with synchrony so as to produce the impulsion in a chamber and suction in the other. The model is a classical moving boundary problem, with time depending boundary conditions in a planar geometry.

Given the nature of the preliminary work, the fluid will be supposed as incompressible and Newtonian in laminar flow. For this geometry it is possible to compute an indicative Reynolds number in terms of blood density and viscosity, flow rate, frequency and a generic 1 cm width. The value is around 2,800 in the limit between laminar and transition flow that doesn't justify a turbulent model. In CFD, turbulent models are recommended from Reynolds number in the order of 10^4 due to the strong approximation and supposition about the dynamics of flow, mainly near walls. Also, when blood flows with flow rate of 50 ml/min or more, the approximation of Newtonian fluid is valid. The flow model will be built by the Navier-Stokes equation (Eq. (1)) and continuity equation (Eq. (2)), while the FSI will be expressed by the boundary condition at the moving boundary.

$$\rho \left(\frac{\partial \mathbf{u}}{\partial t} + \mathbf{u} \cdot \nabla \mathbf{u} \right) = \rho \mathbf{g} - \nabla p + \mu \nabla^2 \mathbf{u} \quad (1)$$

$$\nabla \cdot \mathbf{u} = 0 \quad (2)$$

ρ and μ are the density and viscosity of the fluid, p is the pressure, \mathbf{u} the velocity, t the time and \mathbf{g} the gravitational acceleration.

Fig. 1 shows pulsatile IVAP at zero position ($x=0$). The piston moves with harmonic time dependent velocity: while it impels volume in chamber 1 through V_{o1} valve, it suctions blood in chamber 2 through the V_{i2} input valve. The piston movement constitutes the more important interaction with the fluid because by this, it is defined the velocity fluid in contact with piston boundaries. The chambers are simultaneously active and in each hemicycle they work alternatively impelling and suctioning blood. To make it possible, the input valve and exit valve, must work from “totally open” mode to “totally close” mode or vice versa. In this preliminary analysis, the valves pass from the close estate to open state and vice versa, at very small time intervals.

The boundary conditions for the input and exit flow will be of normal traction with pressure reference: zero pressure at the input flow and pressure reference as a healthy heart (of 100 mmHg order) at the exit flow.

On the other hand, the boundary condition to fix and movable pump's walls will be of non-slip.

$$\begin{aligned} \mathbf{u} &= 0 \quad (\text{fix walls}) \\ \mathbf{u} &= u_w \mathbf{i} \quad (\text{piston wall}) \end{aligned} \quad (3)$$

If the piston movement is assumed as harmonic, the position of the piston center must be expressed as

$$x = A \cos(2\pi f t) \quad (4)$$

and therefore its velocity will be

$$u_w = -2\pi f A \sin(2\pi f t) \quad (5)$$

where A is the piston movement amplitude and f its frequency.

To start computation, the initial conditions were the following:

For fluid flow

$$\mathbf{u}_0 = 0 \quad (6)$$

For the piston position and velocity

$$\begin{cases} x = A \\ u = 0 \end{cases} \quad (7)$$

Additionally, the valves are other moving walls where the non-slip condition is established in terms of their velocities. $v_{wall\ up}$ is the velocity of the valve's upper wall and it is defined as

$$v_{wall\ up} = \begin{cases} Y_{valve\ ef} * fI(t) \text{ si } \cos(2\pi f t) > 0 \\ -Y_{valve\ ef} * fI(t) \text{ si } \cos(2\pi f t) \leq 0 \end{cases} \quad (8)$$

Where
$$fI(t) = \frac{60e^{-60 \sin(2\pi f t)}}{(1 + e^{-60 \sin(2\pi f t)})^2}$$

$v_{wall\ bot}$ is the velocity of the valve's bottom wall and is defined as

$$v_{wallbot} = -v_{wallup} \quad (9)$$

For best comprehension, Fig. 2 shows the valves parts in open and close states:

Fig. 3 shows the position graphs of the piston and valves in a specific period. It is seen that the piston starts its movement from the extreme x position ($x=A$) and the short time in which the valves commute.

The upper and bottom valve parts are supposed to oscillate around a mean position, to achieve both open and close states. The oscillation amplitude is $Y_{valve_{ef}}$ and this movement is appropriate for a smooth mesh deformation. To allow the computation avoiding node superposition and element collapse, $Y_{valve_{ef}}$ is slightly lower than a quarter of the conduit height. In this work $Y_{valve_{ef}}=2.877$ mm. Also, this is the reason of very thin clearance between the valve extremities when completely closed.

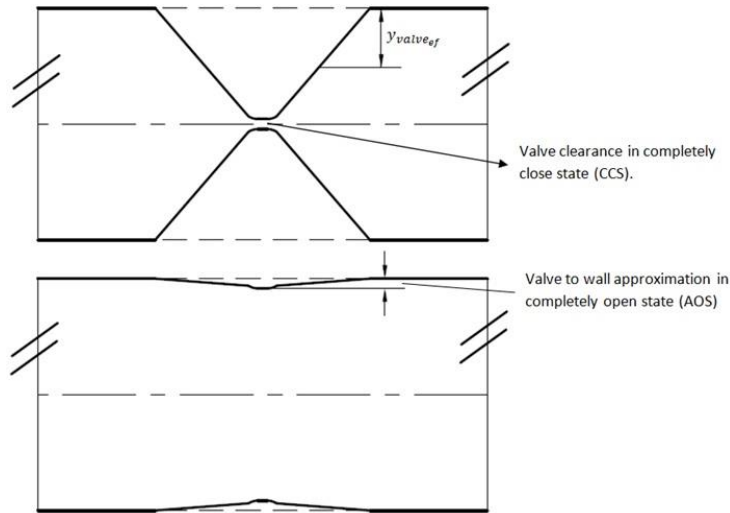


Fig. 2 Scheme of the completely close and open valve states

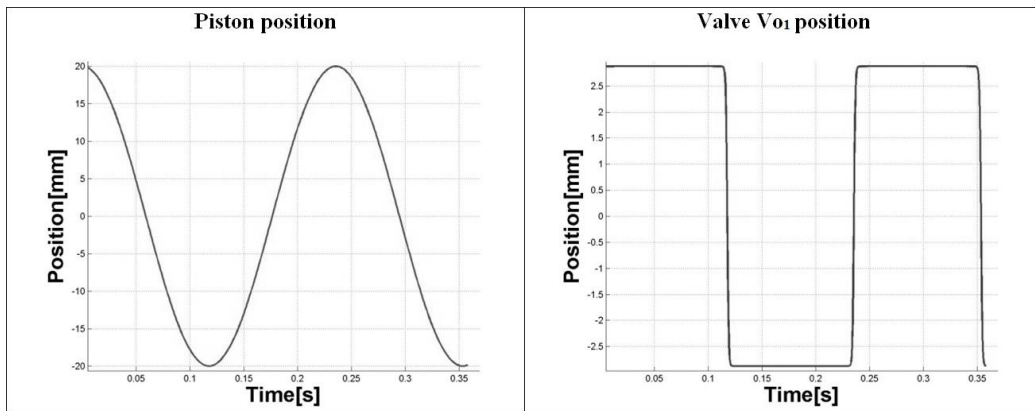


Fig. 3 Piston and valve V_{o1} positions in a specific period of time

4. Finite elements simulation

Despite the complexity of the problem, it might be resolved by means of a standard commercial software. This is the case of COMSOL Multiphysics version 3.3 that has been used in the present work. Although the software has the capacities to simulate 3D flows, in this preliminary work, it is decided to do 2D simulations in a planar geometry, with the aim of obtaining a general knowledge of the performance for a pulsatile impeller device as indicated in the Fig. 1. The low costs of computing time allow faster feedback of results tending to achieve optimum parameters and geometry for the final design. The case analyzed corresponds to a device with: $f=4.25$ Hz, maximum piston amplitude of $A=20$ mm, suction pressure $p_{suc}=0$ Pa, head pressure $p_{head}=15$ kPa (111 mm Hg) and fluid values parameters: density (ρ) of 1.03×10^3 kg/m³ and viscosity (μ) of 3.50×10^{-3} Pa s.

We use the Fluid Mechanics module that has the capacity of managing moving boundaries and time dependent boundary conditions. A non-structured mesh with triangles of P2-P3 kind (second and third order polynomials) is used. The mesh elements are 69,296 and the total degree of freedom is 604,531. This tessellation was chosen because more refined meshes don't produce best results.

Fig. 4 shows the domain tessellation resulting from the use of COMSOL mesh generation tools. In these tools the following maximum sizes of the element (in meters) are selected for different regions of domain: 0.00005 m for clearance zones, 0.0002 m for the valve extremity and 0.00055 m for other zones.

In the refinement process, a balance between precision and computing time is searched. Refinements were made at the clearance of piston and housing zone. The problem analyzed has zones of domain where the formulation could be treated as multi-scale, however, the software is able to resolve simultaneously the flows in all the regions of the domain.

The model is resolved for the u , v and p variables by means of a monolithic scheme, with a temporal integrator of variable order and an adaptive time step with 0.001 s as a maximum value. To resolve the algebraic system equations in the Newton loop, the solver UMFPACK of COMSOL is used.

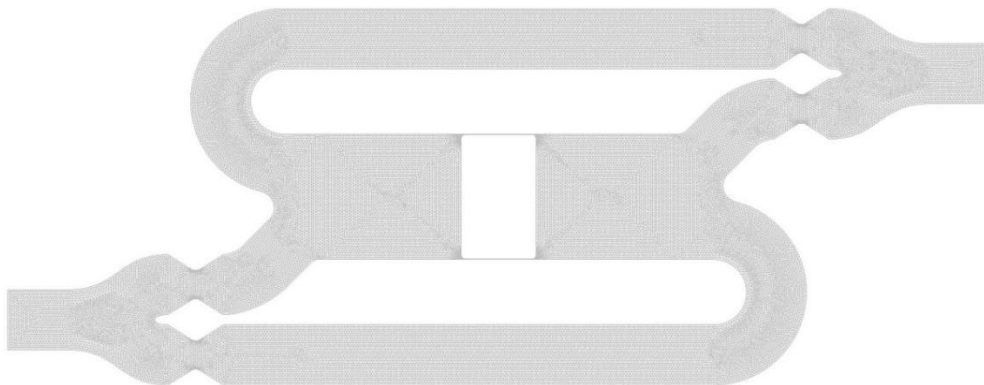


Fig. 4 Mesh used for the simulation, with triangular elements P2-P3. The darker zones correspond to zones of automatic refinement

For a standard execution of 1.73 cycles, a computing time of 263 minutes (4 h 23 min) is necessary in a personal computer with the following characteristics: 4 GB of RAM memory, 1 GB of GRAM memory, CPU AMD Phenom II X4 of 3.4 GHz and a hard disk drive of 1 TB.

As it is seen, the software obtained convergences only for 1.73 cycles, it is thought, that the reason is the error multiplication because the tessellation is insufficient; a more refined discretization should implicate best computational resources. It is noted that the first semi cycle corresponds to a transient phenomenon because the variables are initialized with zeros.

4.1 Equations to post-process the results for temporal mean flow rate, force and power

To calculate the temporal mean flow rate, is used the expression

$$q_m = \frac{1}{N * T} \int_{t_i}^{t_i + N * T} q(t) dt \quad (10)$$

N is the number of cycles simulated for the pump. If we consider that the first semi cycle correspond to a transient phenomenon of the pump functioning, then, this semi cycle will not be consider for the mean flow rate calculus. Therefore, the t_i value is set as $T/2$ and $N=1$.

Also, the piston must move exerting a force over the fluid to overcome the pressure that is established by the body in the arterial circuit. It is evident that this force is the same force exerted by the fluid over the piston but in an opposite direction. The first one may be calculated by integration of the stress vector over the piston area. In this case, due to geometrical assumptions, the integration is made along piston perimeter. Certainly, the force will depend on time, will be cyclic and will be generated by electromagnetic interaction between piston and a linear motor. Therefore, this result is crucial as an input to develop it. The instantaneous force per unit width of transversal direction is computed from the expression (11)

$$\mathbf{F}(t) = \oint \mathbf{t}(\mathbf{n}, t) dl \quad (11)$$

in which the integral is made around the perimeter of the piston in each instant of the work cycle. Where

$$\mathbf{t}(\mathbf{n}, t) = t_x(\mathbf{n}, t) \mathbf{i} + t_y(\mathbf{n}, t) \mathbf{j} \quad (12)$$

Finally, the work per unit time is calculated from the expression (13)

$$P(t) = \mathbf{F} \cdot \mathbf{u} \quad (13)$$

and its mean temporal value, from the following expression

$$P_m = \frac{1}{N * T} \int_{t_i}^{t_i + N * T} P(t) dt \quad (14)$$

5. Results and discussions

Because of the planar geometry -non defined in the third direction-, the results of flow rate,

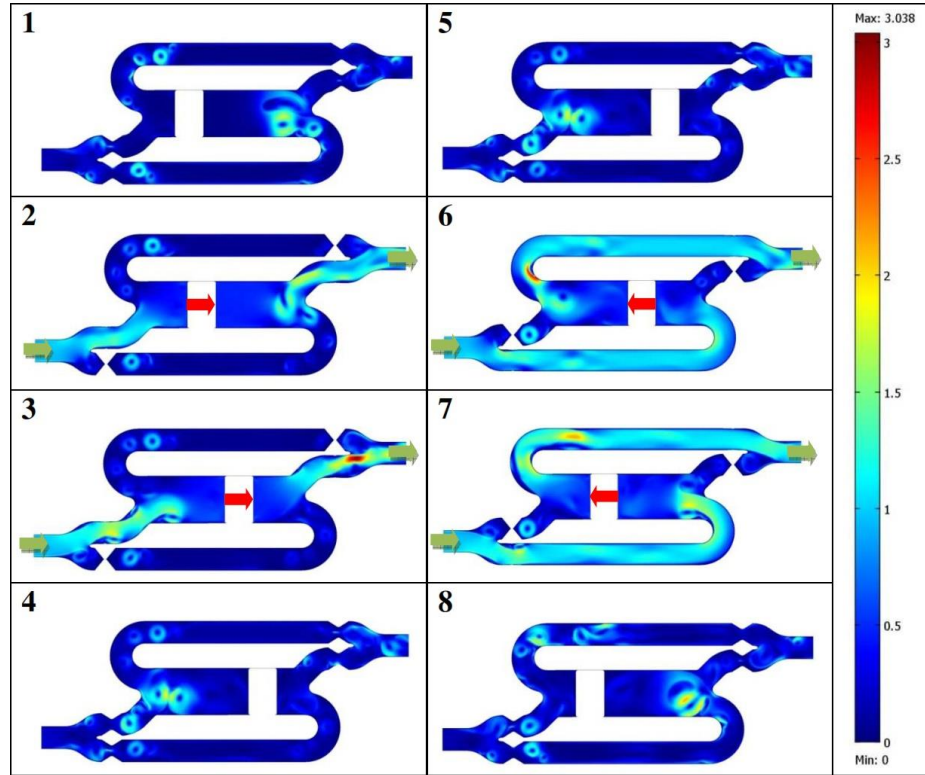


Fig. 5 Color scheme to show vector velocity module variations at eight instants for an entire cycle. The velocities are given in m/s. The maximum velocities are around third of the maximum velocities within the rotational devices

force and power, are expressed by 1cm unit width. Namely, [$\text{cm}^3/(\text{s cm})$], [N/cm] and [W/cm] respectively.

5.1 Velocities

Fig. 5 shows color schemes of the vector velocity module variations in different regions of the device, for eight instants of an entire cycle, obtained as a post process of velocity results. The green arrows indicate the input and output flow and the red arrows indicate the direction of the piston movement at the corresponding instant in the picture. The cycle analyzed covers the second and third hemicycles, between the 0.117 s and 0.361 s instants.

The picture 1 corresponds to the maximum position of the piston, where it is stopped in its movement to the left direction, instantaneously there aren't input and output flows because the Vo_1 and Vi_2 valves are moving from totally open to totally closed, while the other two valves (Vo_2 and Vi_1) are doing the opposite. The flows in each chamber are decelerated and, due to their inertia, strong vortices are produced mainly in the chamber where the piston is suctioned to this instant (the right chamber or chamber 2). Although the vortices are harmful for blood, because they must benefit the thrombus formation and platelet aggregation, the non-stationary characteristic of the device reduces the life of vortices to very short intervals, counteracting the mentioned risk.

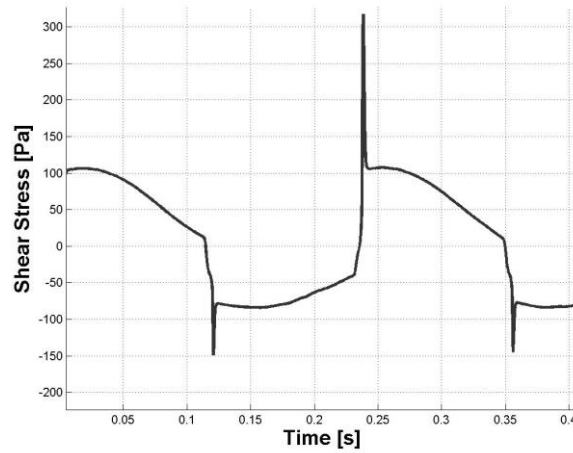


Fig. 6 Maximum shear stress values along 1,73 cycles computed

Each consecutive cell correspond to an increment of 39 millisecond, except cells 4 and 5 that are the same, the cell 4 indicates the final position of the piston movement to the right and the cell 5 indicates the initial movement to the left.

The vortices must be seen clearly, but many of them, are recirculations that move at relatively low velocities. For example, the chain vortices that appear at the instants from cells 1 to 4, are recirculation movements in the suction phase of chamber 1, when the curve branch of output is closed and the input jet in Vi_1 valve generates a weak suction in it. For the instants corresponding to cells 1 and 4 (also 5 and 8), there is a slight reflux because the valves are commuting from open state to close state and vice versa.

The maximum values of velocities into the device are lower than 3.04 m/s, which is an important fact, if these values are compared with the 8.00 m/s that blood has in axial ventricular assist devices. Higher velocities mean higher shear stress near stationary pieces of the device, therefore, it is hopefully that the configuration analyzed in this work will be less harmful for the blood.

5.2 Shear stress and pressure

An important magnitude to take into account when blood is driven, is the shear stress. This is because the red cells, mainly, have a resistance limit of 150 Pa only for 100 s (Grigioni 2004, Leverett 1972, Sallam 1984). If stress or resident time are higher of these values, the red cells break; the phenomenon is named hemolysis and it is produced generally in artificial devices for blood impulsion.

It is crucial to do a design that minimizes the risk of hemolysis; in this case, pulsatile devices appear as a good mechanism. The reasons are: non-stationary flow and the possibility that the stress values will be really low in comparison with rotordynamic impulsion devices.

Fig. 6 shows the shear stress in the gap between piston and chamber walls where the maximum values in the entire cycle can be seen. Despite huge peaks, the maximum shear stress is of 112 Pa - lower than the stress resistance limit for red cells- when the piston is moved to the left, suctioning

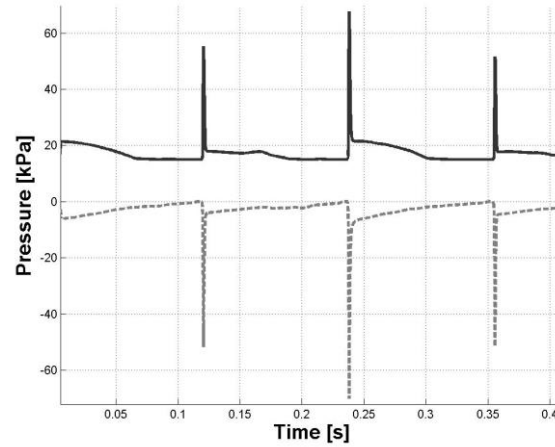


Fig. 7 Maximum and minimum pressure values along 1,73 cycles computed

by the input curve branch and driving by the output curve branch too. The time residence of this stress is very short in comparison with the limit, therefore, it is concluded that the red cells, mainly, might not have any risk of breaking up.

The maximum and minimum pressure values are shown in Fig. 7. The maximum values are produced near to the driven piston face and in input branches when the valves are closed, i.e. when the flow is practically stagnant there. The minimum pressure values, i.e. depressions, appear when blood is suctioned in chamber 2 through the curve branch and at the center of vortices.

The huge peaks in pressure and shear stress at the instants in which valves open and close, emerge because the pumping effect introduced by the valve walls movement, vary the chamber volume. It is noted that peaks depend on the assumed valve shape and size. Peaks will be small for narrowest valves or in presence of a compliant material in input and exit branches, converting pressure work in elastic energy deformation as the aorta do in human body.

5.3 Flow rates

Fig. 8 shows the output flow for the 1.73 cycles analyzed. The dashed line curve belongs to the output flow through the exit valve Vo_1 , and the solid one belongs to the exit valve Vo_2 . The flow is pulsatile and the mean flow rate (see Eq. (10)) is $82.23 \text{ cm}^2/\text{s}$ (i.e. for each centimeter in the transversal direction). For a piston of 1 cm width, the mean flow rate will be of 4.933 l/min, i.e., the flow rate of heart to satisfy the physiologic demand of basal activity, or minimum flow rate to do daily most basic activities. This shows that the size and frequency of the device are appropriated to impulse the required flow for a patient with advanced CCI.

Because of the fluid incompressibility, the input and output flow rates of the pump should be equal. The comparisons between them indicate that the relative error is of 10^{-5} order, considering it, as a good indicator of the correct resolution of the equation.

The clearances under and over the piston allow a reflux between the chambers when the piston impulses in a chamber and suction in the other one. This backflow depends on the time and is related to the piston velocity and pressure difference between chambers. When the pressure

difference induces a flow that is higher than the lateral surface piston drags, the flow will be backflow at these instants. Fig. 9 shows the flow rates fluctuations at the top clearance (black) and bottom clearance (grey) for the 1.73 cycles analyzed. It is seen clearly that, in general, there is a flow towards one chamber to another (positive or negative flow rate) assuring “washing” (removal and replacement) of the fluid remaining in this part of the device. In the case of blood, this is very important to avoid coagulation and necrosis.

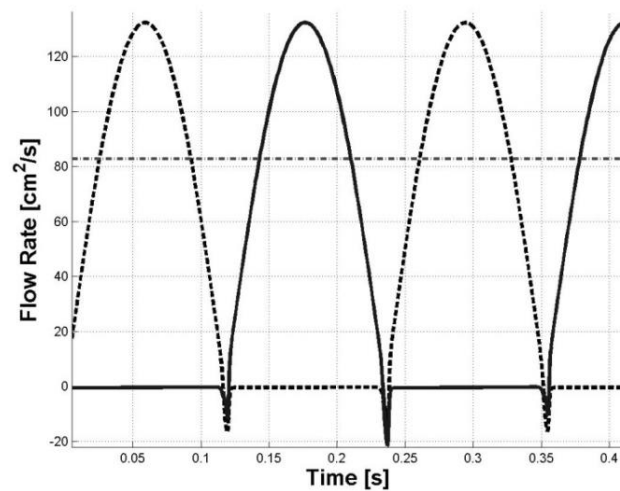


Fig. 8 Output pulsatile flow rate for the analyzed device. The temporal mean flow rate will be of 5 l/min if the piston have 1 cm width, i.e. the physiologic flow rate for basal methabolism

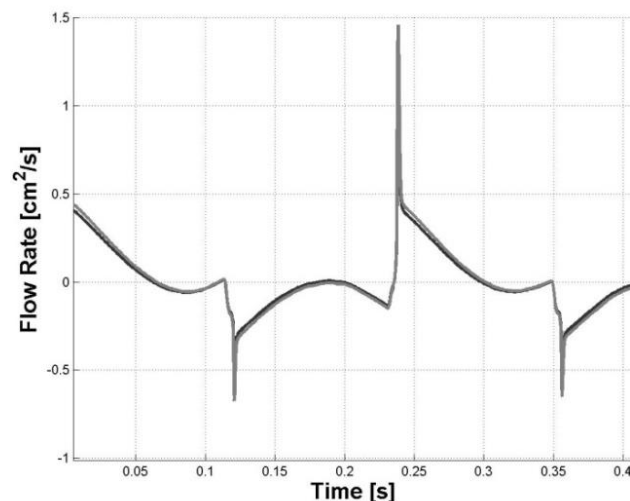


Fig. 9 Flow rates into clearances between piston and housing for the 1.73 cycles analyzed. The black curve corresponds to the top clearance and the grey corresponds to the below clearance. The non-zero flow rate revels the washing and renewal of fluid into the clearance

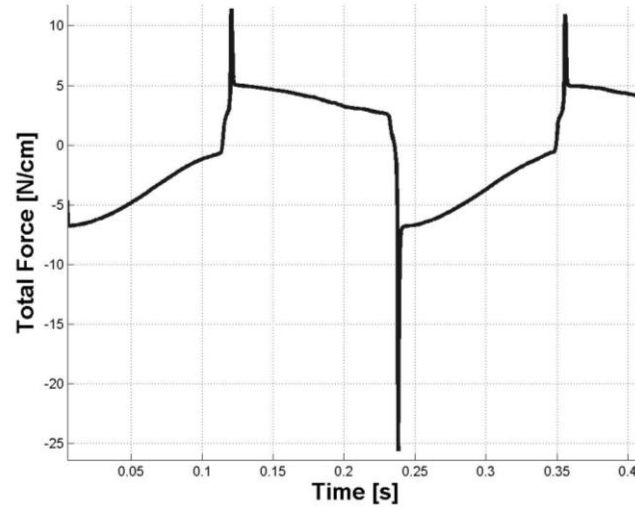


Fig. 10 Exerted force by piston over fluid in x direction for the 1.73 simulated cycles. The piston starts its movement at $x=A$

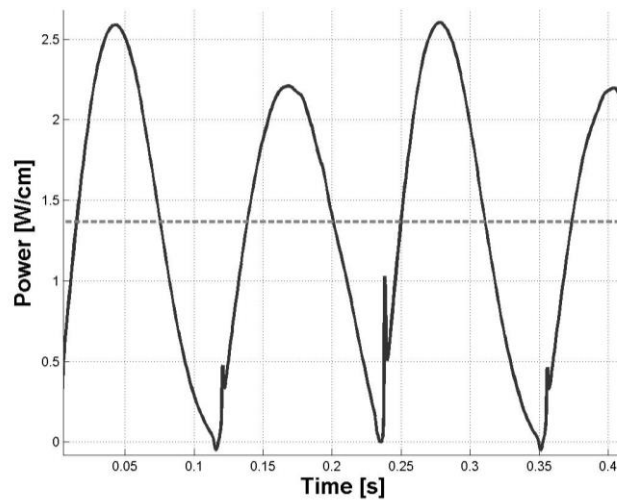


Fig. 11 Work per unit time made by the piston over the fluid. The difference in the successive semi cycles is a consequence of the higher energy demanded by the input and output curve branches. The 1.37 W/cm temporal mean value is represented by a dashed line

5.4 Force and power

Fig. 10 shows the time variations of force exerted by piston over the fluid in x direction, assuming that piston starts its movement in $x=A$. It can be seen the higher gradients when the valves commute and the chambers change from drivers to preloads and vice versa. If the piston is supposed to be 1 cm width, the maximum force could be limited to an intensity of 25.61 N.

Other effect that emerges from the device asymmetry is a net force in transversal direction y ,

that is two order lower than the force in x direction. It could cause a pitching effect or periodic piston misaligned. By the reason of its small magnitude, this transversal force is not analyzed in this work.

On the other hand, the work per unit time make by the piston over the fluid, that will be a portion of power supplied from the motor (other important portion of power is required to accelerate and decelerate the piston), is obtained by the force and velocity product in the axial direction. Fig. 11 shows the work per time unit for the 1.73 simulated cycles, it can be seen that there is a higher need of work in the first and third semi cycles because of the input and output curve branches.

Analogously to the temporal mean flow rate calculus, $t_i=T/2$ and $N=1$. This mean temporal value is 1.37 W/cm.

6. Conclusions

In this work, it is shown a preliminary computational analysis for the development of a new implantable ventricular assist device, to be used in patients with terminal cardiac insufficiency. The device is inspired on a pulsatile mechanism of double effect, which impulses blood to a frequency higher than the physiological one; the element for pumping is a piston that doesn't have contact with any part or wall of the device and can be driven by means of an electromagnetic linear motor. The device is not symmetric and has two branches for suction and two branches for impulsion; the piston defines two chambers where alternatively produce suction and impulsion. Each branch for suction has a valve that opens and closes the area pass, working in the transversal direction of the flow. The same happens in the output branches. The simulation is made assuming incompressible Newtonian laminar flow in a planar configuration to allow a 2D model. The model equations are resolved by means of the software COMSOL Multiphysics version 3.3, executed in a computer of standard architecture.

The results are encouraging because they show a fluid-structure interaction that could cause lower stresses than the resistance limit of blood particles, mainly the red cells. Therefore, it is thought that the device could produce low taxes of hemolysis. Also, the values of driven forces and power make predictable their supply by means of electromagnetic interactions.

The vortices that take place in different regions of the two chambers are harmful for the blood. However, the low velocities involved in these vortices and their short life due to the pulsatility of the mechanism, make and destroy them minimizing the risk of associated thrombosis.

From this preliminary analysis arises future works that might be resumed as:

1) Planar geometry (2D):

- Analyze how to optimize the piston size and shape as well as the clearance size between piston and housing.
- Analyze optimum size, shape and opening and closing time of valves, using appropriate mesh adaptation.
- Particular analysis of the flow in regions where Newtonian approximation is doubtful.

2) Spatial geometry (3D)

- Design a realistic configuration based on cylindrical shapes for the chambers, piston and input/output conduits.
- Computational analysis of turbulent flow into the realistic shape of the device, for higher frequency than the present work (Ashrafizaadeh 2009, Ismail 2008).

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