THREE-DIMENSIONAL NUMERICAL SIMULATIONS OF THE AORTIC FLOW IN PRESENCE OF A LEFT VENTRICLE ASSIST DEVICE

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ABSTRACT
Cardiovascular diseases are considered to be the main cause of death in developed countries. Due to limitations resulting *in-vivo* measurements of velocity, the analysis and evaluation of hemodynamic parameters by means of computational simulations become the only efficient solution, especially in pathological or assisted condition cases.

The aim of this study is to analyze, through a CFD model, the hemodynamic effects of a continuous flow Left Ventricle Assist Device (LVAD) evaluated in three different working conditions.

The proposed model provides a computational fluid dynamics analysis of LVAD inserted into the descending aorta in order to evaluate flow distribution at the most interesting aorta branches.

The results show the development of the flow in three different cases, showing that the LVAD investigated is able to regulate flow in the aorta.

Keywords: Left Ventricular Assist Device (LVAD), Aortic flow, CFD, Cardiovascular modeling.

1. INTRODUCTION

Cardiac failure has been defined as the epidemy of third millenium. The high number of patients requiring heart transplant for end stage cardiac failure shows that replacement therapy is not a real option for all patients. Therefore mechanical assistance is becoming a valid option to bridge patients to cardiac transplantation and to a destination therapy. In 2002, Dr Westaby and collegues (Westaby *et al.* 2002) implanted axial flow pump from the apex of the left ventricle to the descending aorta (Jarvik Heart Inc., New York). Low incidence of postoperative complications and totally implanted system allowed an increasing number of patients treated with Jarvik Heart. Some of them, reached follow-up above five years from the implantation. However, the main question in patients of Jarvik Heart is represented by the flow pattern to the brain with a source of retrograde flow.

Computational flow study of hemodynamics problems is extremely challenging, and the potential advantage of computational tools for improving health care is huge. More Computational Fluid Dynamics (CFD) studies about mechanical hearts, heart valves or assist devices have been performed in the last years. CFD is a valid alternative to experimental methods because it can produce flow field data in great detail. Computational analysis can also give helpful information to optimize mechanical devices design. Costs and times are lower than those required by an empirical approach (Kiris *et al.* 1998; Bazilevs *et al.* 2009).

The aim of this study is the evaluation of the pattern of flow in the sovra-aortic vessels and descending aorta in different condition of flow generated by LVAD implanted with an outflow in the descending aorta.

2. MATERIALS AND METHODS

LVADs are now part of routine devices at surgeons disposal for mechanical support in case of failing heart (Selzman *et al.* 2007). An LVAD is a life saving tool used when the natural heart is unable to provide sufficient blood flow. This device is a compact pump that is helpful for both bridge to transplantation and permanent mechanical circulatory support.

An implantable LVAD has to be small and efficient, generating 5 $l\ min^{-1}$ blood flow rate. Since the instrumentation for flow measurements is extremely difficult, it was helpful to analyze hemodynamic behavior by using computational techniques (Kiris *et al.* 1998).

LVAD produces two different hemodynamics effects:

- It causes an end-diastolic pressure decreasing in the left atrium;
- The systemic flow rate increases and, as a consequence, the systemic venous return and the right ventricle preload increase as well.

The device considered in this work is Jarvik 2000 Heart (Jarvik Heart Inc, New York, NY), a compact axial flow impeller pump implanted within the apex of the failing left ventricle that is positioned inside the descending thoracic aorta (Westaby *et al.* 2002).

The effect of an LVAD on hemodynamics is complex and demands a three-dimensional model of the aorta. For this study the geometrical model of the aortic arch has been constructed from Magnetic Resonance Images (MRI) and numerical simulations have been performed by means of commercial CFD software.

2.1. Geometric model

The geometry of the aorta is based on NURBS mesh reconstruction by using a commercial CAD software.

The geometric aortic model is generated as a smooth circular tube surface based on a given parametric geometry (Fung *et al.* 2008).

Blood vessels are tubular objects. Therefore, to reconstruct the aorta model the sweeping method has been used, as shown in Figure 1. This method is based on creating a path (Figure 1a), which constitutes the skeleton of the aorta. On the path, a template circle is translated and rotated to each cross-section in order to obtain the three-dimensional geometry (Figure 1b) (Zhang *et al.* 2007).

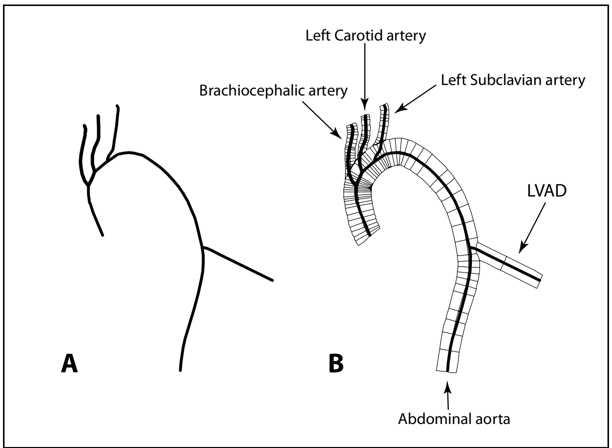

Figure 1: The aorta model reconstructed with skeleton-based sweeping method. Path (a); the sweeping method (b).

An LVAD branch was added to the model in the aorta descending part (Bazilevs *et al.* 2009).

The CAD geometry of the 3D model was used as input for the CFD simulations.

The simulations, preprocessing, meshing, solving, and post-processing were performed by using a commercial finite element analysis software (COMSOL 4.2a, COMSOL Inc, Stockholm, Sweden).

2.2. Governing equations

Blood flow in large arteries is typically well approximated as a Newtonian fluid (Fung *et al.* 2008). The fluid is assumed to be incompressible and homogeneous, with a Newtonian behavior (Shipkowitz *et al.*2000).

The blood flow in this model was predicted using the three-dimensional Navier-Stokes equations for incompressible flow (Equation (1), (2)).

$$\rho\frac{\partial \mathbf{u}}{\partial \mathbf{t}} + \rho(\mathbf{u}\cdot\nabla)\mathbf{u} = \nabla\cdot\left[-\mathbf{pI} + \mu\left(\nabla\mathbf{u} + (\nabla\mathbf{u})^{\mathbf{T}}\right)\right] + \mathbf{F} \tag{1}$$

$$\rho\nabla\cdot\mathbf{u} = 0 \tag{2}$$

where $\boldsymbol{\rho}$ is the fluid density, $\boldsymbol{u}$ = *(u, v, w)* is the fluid velocity, $\boldsymbol{p}$ is the pressure, $\boldsymbol{I}$ is the unit diagonal matrix, $\boldsymbol{\mu}$ is the kinematic viscosity, and $\boldsymbol{F}$ is the volume force.

Aortic wall was assumed to be rigid and therefore no-slip condition was applied at the aortic wall (Tse *et al.* 2011).

2.3. The mesh grid

On the basis of the geometry described above, the model is discretized with tetrahedral elements.

The final mesh, used in the whole flow simulation, consisted of 180,181elements and 135,822 degrees of freedom.

The system of linear algebraic equations were solved using the in-built direct sparse solver, PARDISO. A desktop computer with 2.53 GHz Intel Xeon 64, Quad Processor and 8.0 GB of RAM was used in all computations.

2.4. Blood flow parameters and boundary conditions

Time-varying flow rates boundary conditions are applied at the inlet branch in the fluid domain. Three different cases have been considered where the aortic inlet flow decreases with increasing flow rate in LVAD (Figure 2):

- **CASE 1 -** LVAD has a flow rate of 16.66 cm^3/s (1 $l\ min^{-1}$) and flow occurs through the aortic root.
- **CASE 2 -** LVAD is operating in the regime where over one half of the blood supplied to the aorta comes from the pump, with a flow rate of 33.33 cm^3/s (2 $l\ min^{-1}$).
- **CASE 3 -** LVAD is operating in the regime where nearly all the flow comes from the LVAD with a flow rate 66.66 cm^3/s (4 $l\ min^{-1}$), while the aotic valve is closed and the , flow is null.

The fluid is a blood substance with a density ρ = 1060 $kg*m^{-3}$ and dynamic viscosity μ = 0.0035 $Pa*s$ (Park *et al.* 2007).

The flow is limited to the flow model without considering the heat transfer and the temperature is constant at 38 °C.

The pulsatile flow rates waveform applied to the inlet aorta have been derived, analytically, from literature data (Kim *et al.* 2009, Olufsen *et al.* 2000).

The waveforms of flow rate have been obtained according to the equation (3):

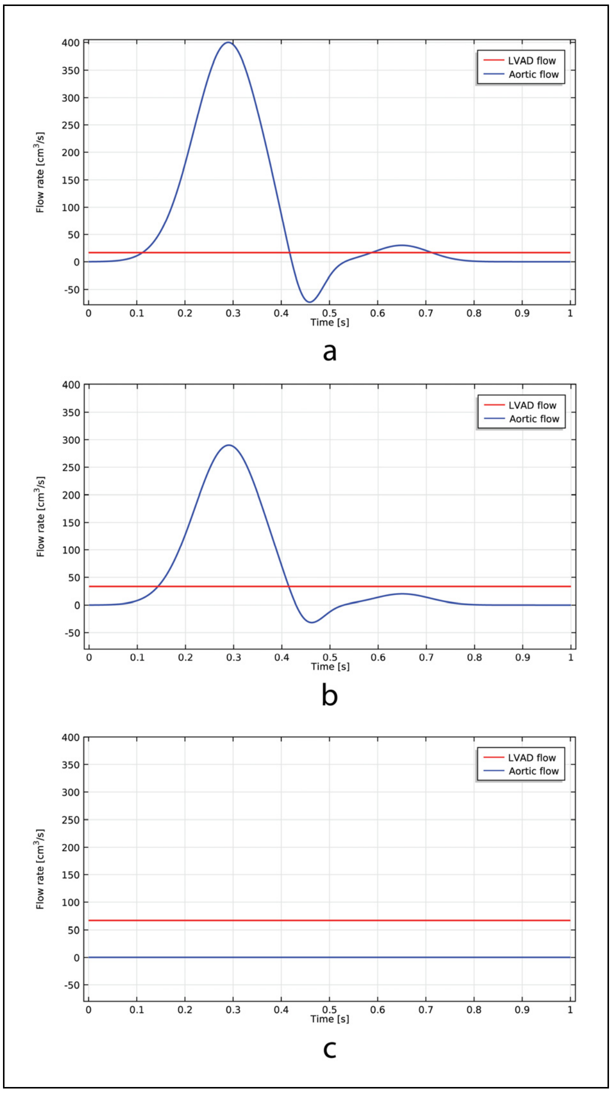

Figure 2: Inlet flow waveforms of the aorta (blue line) and of the LVAD (red line) in presence of an LVAD flow rate of 16.66 cm^3/s (a) 33.33 cm^3/s (b) and 66.66 cm^3/s (c).

$$\sum_{i=1}^{n} \boldsymbol{a}_i \cdot \boldsymbol{e}^{-\boldsymbol{b}_i(\boldsymbol{c}_i-\boldsymbol{t})^2}, \quad \boldsymbol{n}=3 \tag{3}$$

where a_i, b_i and c_i are constant, and t is the time variable. The course of the function is shown in figure 2.

For the fluid domain, the boundary conditions at the outlet branches are set as shown in Figure 3 (Bazilevs *et al.* 2010). The boundary conditions are described in detail in the following:

$$P_n = C_r \cdot q + p_0 \tag{4}$$

where C_r is a constant resistance, q is the volumetric flow rate, and p_0 is the physiological pressure level.

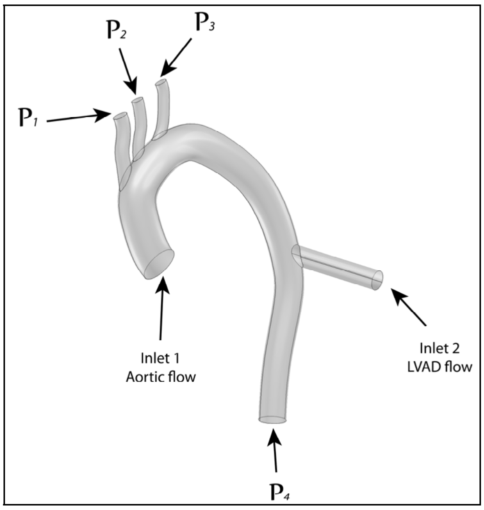

Figure 3: CAD Model and boundary conditions for the fluid domain.

In the performed simulations, only two cardiac cycles have been considered. The output was taken at the second cycle in order to damp the unstable effect of initial cycle (Tse *et al.* 2011).

Based on the steady-state second cycle, in the subsequent paragraph numerical analysis will be discussed.

3. NUMERICAL SIMULATIONS AND RESULTS

3.1. Computational results

Numerical results obtained for the descending aortic distal anastomosis are in agreement with clinical observations and findings for this configuration.

This study was undertaken to assess the effect of blood flow in the outlets branches of the aorta in patients with LVAD. Referring to figure 3, the following outlets branches of the aorta have been considered: (P1) brachiocephalic artery, (P2) left carotid artery, (P3) left subclavian artery. Also, the outflow of the abdominal aorta (P4) was considered.

The flow field in these arteries is more difficult to study and depends on vessel geometry.

The results are a numerical comparison of the hemodynamic response in the cardiovascular system that describes the values of the flow-rate output of the arteries in the three configurations considered.

In figure 4, the values relating to the outlet flows are shown.

In figure 4a, blood flow through the aortic branches during the first work conditions has been illustrated. In correspondence of the systolic peak, an increasing flow in the abdominal aorta and in the brachiocephalic artery can be seen. Inside the carotid and subclavian arteries the same peak is shifted in time.

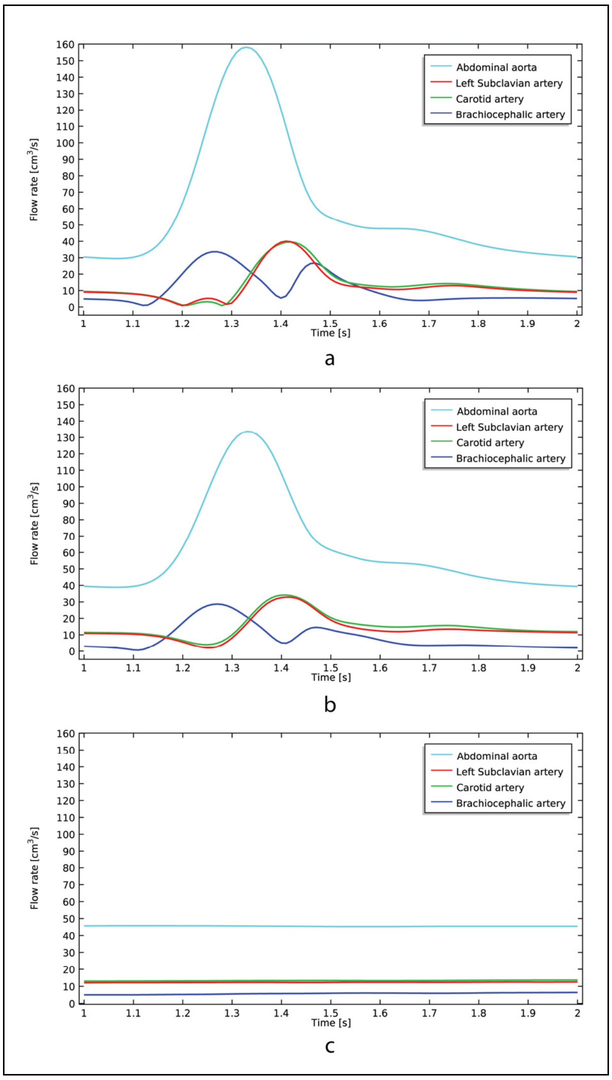

Figure 4: Flow rate distribution among outflow branches during a heart cycle in presence of an LVAD flow rate of 16.66 cm^3/s (a) 33.33 cm^3/s (b) and 66.66 cm^3/s (c).

Figure 4b describes the outlet values of the flows assuming the second condition. These values are slightly different from those reported in figure 4a. It is possible to note a decrease of the systolic peak in all the four ducts; during the diastolic phase, only a little increase of the flow in the subclavian artery, carotid and thoracic aorta appears.

Figure 4c describes the outlet values of the flows considering the third condition, in which the flow comes almost entirely from LVAD. In this case, the output values keep constant values.

In Table 1 the mean outlet flow rate values of a cardiac cycle are reported, by considering the various aortic districts. Values are expressed in liters/minute, for each of the three cases analyzed. These values are referred to the second cardiac cycle.

Table 1: Flow rate through the peripheral vessels. The values are expressed with *l/min* dimensions.

Flow rate [*l/min*]			
	Case 1	Case 2	Case 3
Brachiocephalic artery	0.57	0.42	0.30
Carotid artery	0.69	0.75	0.67
Subclavian artery	0.64	0.68	0.61
Abdominal aorta	3.08	3.13	2.40

Figure 5 shows the velocity vector plot of the flow lines for all the cases considered during the maximum value of the systolic phase ($t=1.3$ s). Figures 5a and 5b illustrate how the flow in the aorta is controlled by the pulsed input signal and maintains a laminar profile,

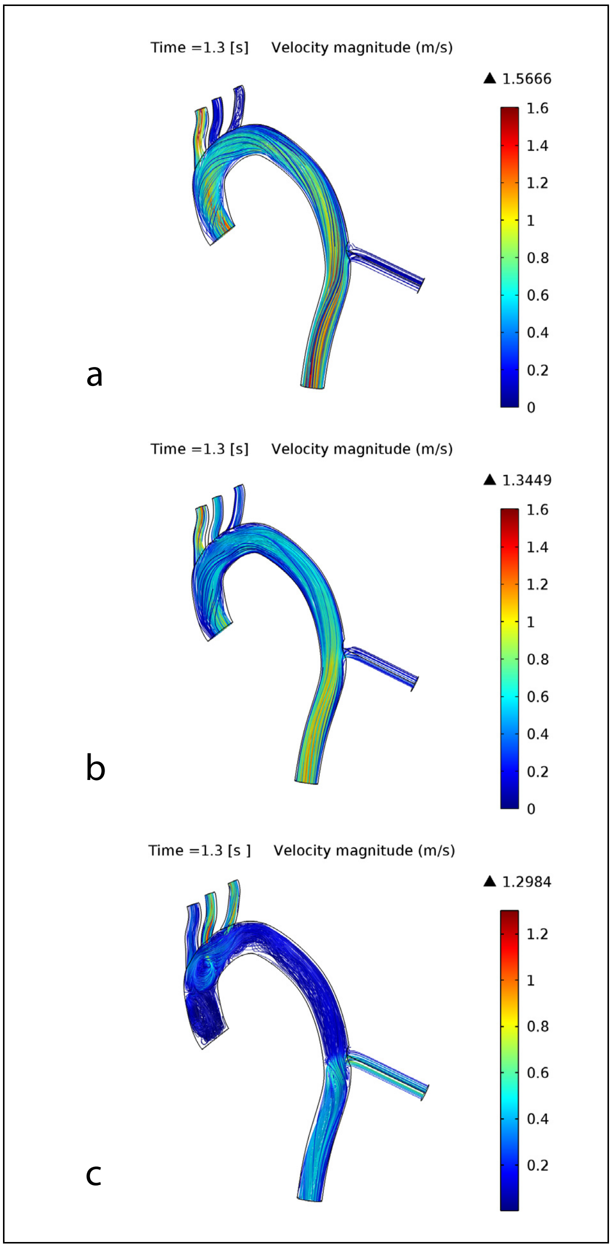

Figure 5: Velocity magnitude streamlines at systolic peak in presence of an LVAD flow rate of 16.66 cm^3/s (a) 33.33 cm^3/s (b) and 66.66 cm^3/s (c).

while in Figure 5c it can be seen that the blood, reaching entirely from the LVAD, creates vortices in the aortic arch and in the closeness of the brachiocephalic artery; these vortices are due to the closure of the aortic valve.

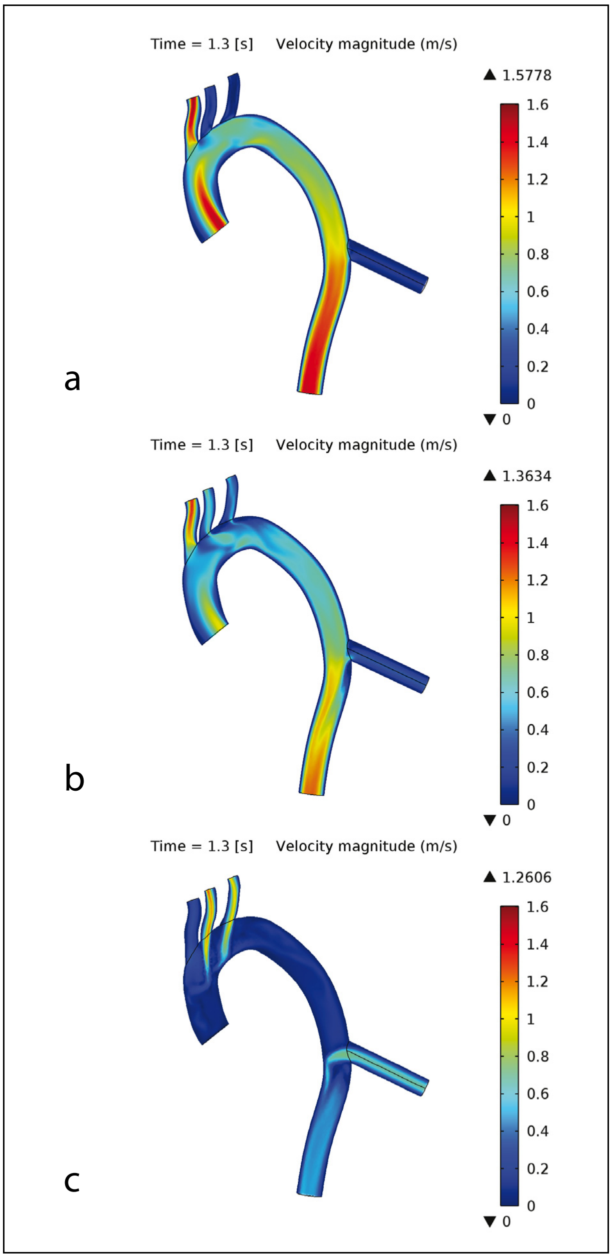

Figure 6: Instantaneous distributions of velocity in the cross-sections, during the systolic peak in presence of an LVAD flow rate of 16.66 cm^3/s (a) 33.33 cm^3/s (b) and 66.66 cm^3/s (c).

Figure 6 shows instantaneous distributions of velocity in the cross-sections within the aorta during the maximum systolic peak.

In Figure 6a it is possible to observe a parabolic evolution of the flow into the aorta, with velocity value greater than in the other two cases (figure 6b-c). Therefore, in the aorta velocity decreases with the increasing of the flow rate in the LVAD.

The corresponding values of average velocity during a cardiac cycle for the three cases are listed in Table 2.

Table 2: Average velocity in output through the peripheral vessels during a cardiac cycle. The values are reported in *m/s*.

Velocity Magnitude [*m/s*]			
	Case 1	Case 2	Case 3
Brachiocephalic artery	0.28	0.21	0.13
Carotid artery	0.50	0.49	0.45
Subclavian artery	0.46	0.46	0.41
Abdominal aorta	0.38	0.36	0.27

DISCUSSION

Nowadays, it is plausible that the onset of disease is also closely connected to the alteration of the fluid structures of the cardiovascular system. In order to study and interpret quantitatively these relationships, a certain degree of control over the fluid-dynamic variables, both versus time and versus space is essential. Unfortunately, conventional methods of investigation, that are based on theoretical or *in vitro* models, have proved inadequate or, only partially, effective in supporting this type of analysis. In this context, an alternative approach is provided by models of CFD.

In recent years, many researches have been oriented to the mathematical representation of vascular LVAD, to the description of the model of blood rheology, to the complex multi-layer structure of the vascular tissue and to possible diseases, with the chemical and mechanical interactions among blood and vessel walls.

The employ of computational models can accurately reproduce the conditions of circulatory and ventricular origin and evaluate the related effects of ventricular assistance device under different operating conditions.

These applications are a useful test bench for numerical optimization methods available for studying in order to ensure cardiologists and cardiac surgeons the necessary support for the study and prediction of the arterial system behavior in physiological and pathological conditions in the presence of LVAD.

The goal of this paper has been to characterize the dynamics of the blood flow, in the presence of LVAD implanted by a simple suture, in the section of the descending aorta. In this way, it was possible to demonstrate the effectiveness of this device in the regulation of the flow inside the conduit.

By means of transient simulations, it has been possible to verify three different operating conditions, each characterized by a particular continuous flow input to the LVAD. In spite of this variation, a certain equivalence of velocity profiles and outlet flow rate quite balanced in the various branches was noted.

Based on these considerations, it has been possible to describe this phenomenon by means of a quantitative model able to predict the behavior of the aortic blood flow in the section, in the presence of such a device, in order to provide a useful tool in the clinical field.

Appropriate simulations, indeed, allow pre-assessments from both the diagnosis and the achievable therapies points of view, since they guarantee correct pre-surgery estimates in relation to the choice of the VAD to be implanted and a valuable observation about the artery reactions when exposed to any continuous and pulsatile flows.

CONCLUSION AND FUTURE WORK

A future interesting development could include the analysis of the blood flow based on the placement of the cannula of the VAD, by also considering other surgical techniques, such as the apical positioning.

Furthermore, the model will be considered also in conditions of wall deformation and elasticity in order to characterize, in a suitable way, the behavior of such systems.

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