NUMERICAL AND EXPERIMENTAL EVALUATION OF THE PRESSURE DROP IN A BLOOD FILTER

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Abstract: *Extracorporeal circulation is a procedure used in several clinical applications like cardiopulmonary bypass used in heart surgeries. During the surgery microembolies can be introduced into the patient arterial system. This embolies can be thrombus, bubbles air, platelets agregated, plastic particles or suture wires which can lead to cerebral damages. These damages resulted in the development of arterial blood filtres. Several studies about the flow effect over blood cells have been published in the last years in order to improve the project of pumps and heart valves. The flow analysis through blood filters e its effect over blood cells is a valuable tool for an adequate project of blood filters. In this work the flow in a blood filter was performed using computational fluid mechanics. Results are in good agreement with itlerature data.*

Keywords*: filter, blood, extracorporeal circulation, hemolysis, simulation.*

1. INTRODUCTION

The extracorporeal circulation is a proceeding where machines and equipment performs heart and lung functions, while these are impeded to work, due to the nature of surgical intervention. The pumping function is discharged by a mechanic pump, while oxygenators achieve gas exchange of the blood normally made by lung. The blood oxygenation and its pumping are achieved externally to the patient's body. The blood is drained from patient by gravity through the cannula inserted in the vena cava. This blood is accumulated in the (cardiotomy) recipient where particles bigger than 100 μm are removed by a polyester grid. Then, the blood is pushed by a pump being guided until the oxygenator where the gas exchange is made. The oxygenated blood passes through an arterial filter and then is infused into the patient´s arterial system.

The cardiopulmonary bypass is a diffused technique among the heart surgery crews. microembolies can be accidentally or inadvertently introduced in the arterial system of the patient. These microembolies can be thrombus, air bubbles, platelets agregated, plastic particles or suture wires. The observation of lung, cardiac and neurological dysfunction on immediate post-surgery and It´s correlation with the presence of embolus stimulated the development and use of filters. The primary objective of these filters is to lock all harmful material flowing through the extracorporeal circulation circuit, originated from equipment or surgical proceedings. A good arterial blood filter should operate within 6l/min outflow, offer short resistance to the blood flow, damage the blood cells as minimal as possible, present a short volume of filling and allow the air removal of the extracorporeal circuit.

It´s valid to say that the arterial filter is a extracorporeal dispositive which increase the security of extracorporeal circulation, avoiding the occurrence of embolisme. The introduction of the filter in the extracorporeal circulation circuit causes some disadvantage. The pressure gradient in the filter should be the minimum possible, providing low levels of hemolysis. The filter itself introduces a risk of (embolia aérea) due to the possibility of release, during the surgery, of retained air at the porous media, along the initial part of filling (Bergdahl e Björk, 1980). The filter can damage the blood cells due to the contact of these cells with artificial surfaces. The use of arterial filter in the removal of macro and micro embolus is controversial. Some people defends its use with the argument of brain damage reduction while others considerate its use questionable due the lack of conclusive clinical information (Kim *et al*., 2000). Although the focus of thrombus formation and the occurrence of (embolism) in oxygenators and arterial blood filters is known, there aren´t quantitative information about the rate, dimensions and quantity of embolus freed by oxygenators or filters. The tests made by Dewanjee *et al*. (1992) show the formation of thrombus and embolus after filter, due to the high concentration of substances freed by platelets and the presence of activated coagulation factors.

By this manner, the analysis of flow through the arterial blood filter and its effect over the blood cells constitute an essential tool to the development of arterial blood filters. The development of a large part of the filters available in the market have been made by experience which involves the evaluation f the product by the surgeons and. This development way is essentially a process of trial and error, making the process long, expensive and inefficient. Flow analysis techniques used in engineering can be a great support in the project of blood filters. The flow analysis have been done through experiments or by the use of numerical solutions. Andrade *et al*. (1997) made use of a flow visualization technique to improve the Project of a blood pump. Pinotti e Paone (1996) utilized a laser Doppler anemometry technique to study the flow inside a blood pump and estimate the damage occurred because of it. Castellini et al. (2004) made use of a Particle Image Velocimetry to evaluate the flow in a cardiac valve. Huebner (2003) utilized flow visualization technique through coloring injection, Laser Doppler anemometry and Particle Image Velocimetry to evaluate the flow of several blood filters produced in Brazil. Burgreen *et al.* (2004) and Untaroiu *et al.* (2005) made use of flow visualization techniques attached with numerical simulations to evaluate the flow in a ventricular assistance device. Gage *et al.* (2002) e Gartner *et al.* (2000) evaluated the flow in membrane oxygenators using computational fluid mechanic. It can be noticed that the use of numerical simulations attached with visualization and/or velocity measuring can be very helpful in the analysis of flow inside blood interacted displacements such as pumps, oxygenators and blood filters.

2. METHODOLOGY

The filter has an acrylic body and a polyester (filtering element) with porosity of 40 µm. The inlet and outlet connectors are made of reinforced acrylic of 9,5 mm diameter. The filter is built to adult patients and works with outflow within 6l/min. It has a line of air removal and recirculation to the venous blood and a shortcut line (bypass) which can be used in case of filter obstruction. The figure 1 shows components and the principle of working of a blood filter. The white arrow represents the "clean blood" and the blue one represents the "blood to be cleaned". The whole region before de filtering element is denoted as entrance channel while the exit channel involves all the region placed after the filtering element, as shown in figure 2(b).

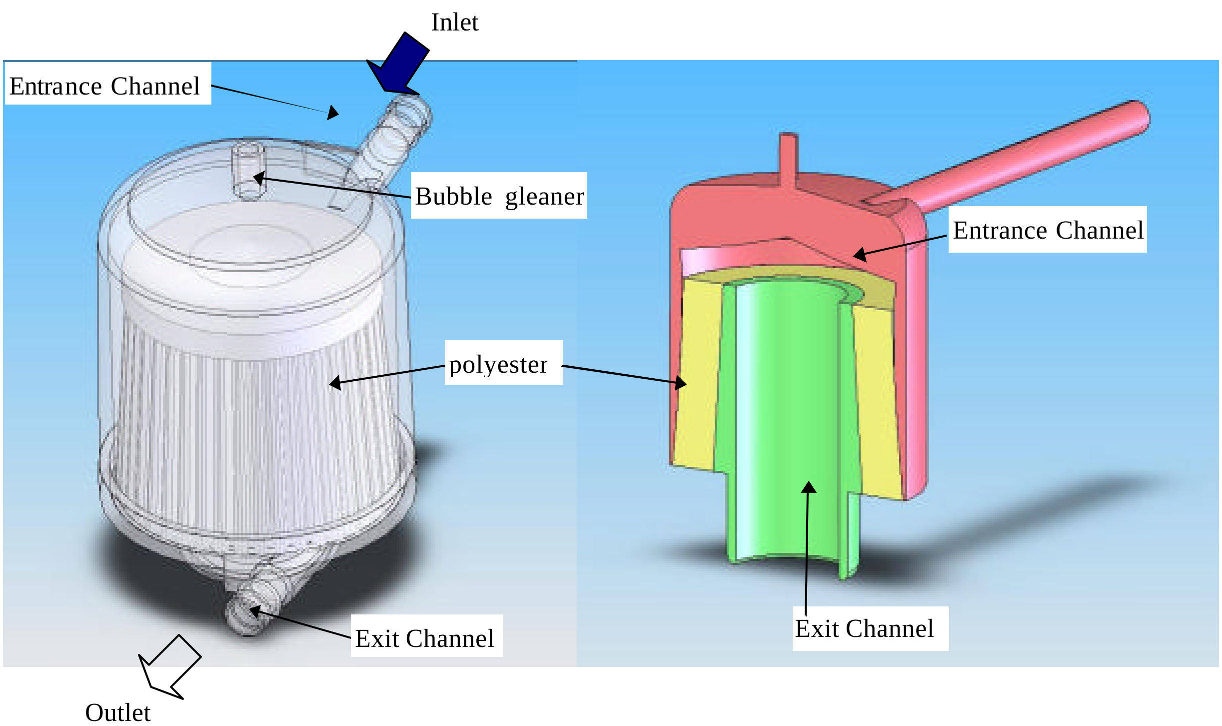

Figure 1. (a) Filter overview. (b) Plane along the centerline

The flow can be represented by the mass conservation, momentum and a turbulence model. Turbulent flow fields evaluated with the Navier-Stokes equations can be time averaged in space or time [Chung, 2002]. When this procedure is adopted the equations describing the mean flow have averaged products of velocities fluctuations. Is this work the model considered is averaged in time and the fluid is considered incompressible.

The mass conservation equation is written as follows

$$\nabla.\vec{u} = 0 \tag{1}$$

The momentum equation is given by

$$r\vec{u}.\nabla\vec{u} = -\nabla p + \nabla.\left[(m + m_T)\left(\nabla\vec{u} + \nabla\vec{u}^T\right)\right] \tag{2}$$

Where ρ represents the specific mass, p pressure, u velocity vector, μ viscosity and μ_T turbulent viscosity.

The turbulence model adopted was K–OMEGA due to its higher numerical stability and accuracy on the flow rate in study. The Transport equations for the turbulent kinetic energy and the specific dissipation are given by

$$\frac{\partial k}{\partial t}+U_j\frac{\partial k}{\partial x_j}=t_{ij}\frac{\partial U_i}{\partial x_j}-b^{*}kw+\frac{\partial}{\partial x_j}\left[\left(n+s^{*}n_t\right)\frac{\partial k}{\partial x_j}\right] \tag{3}$$

$$\frac{\partial w}{\partial t}+U_j\frac{\partial w}{\partial x_j}=a\,\frac{w}{k}\,t_{ij}\frac{\partial U_i}{\partial x_j}-bw^{2}+\frac{\partial}{\partial x_j}\left[\left(n+sn_t\right)\frac{\partial w}{\partial x_j}\right] \tag{4}$$

The Kinematic Eddy Viscosity is given by

$$n_t=\frac{k}{w} \tag{5}$$

The Closure Coefficients and Auxiliary Relations are as follows

$$s=\frac{5}{9},\quad b=\frac{3}{40},\quad b^{*}=\frac{9}{100},\quad s=\frac{1}{2},\quad s^{*}=\frac{1}{2},\quad e=b^{*}wk \tag{6}$$

The effect of the filtering element was simulated using Darcy law (Gage *et al.* 2002). The pressure drop is included in the source term of the momentm equation along the radial direction.

$$?\,P=\frac{\mu}{K_p}\cdot V \tag{7}$$

where P is the mean static pressure, μ is the dynamic viscosity , K_p is the permeability of the porous media e V is the superficial velocity.

The boudary conditions are:

- At the filter inlet the mass flow rate, k and w were prescribed;
- At the walls the no-slip condition was applied.
- At the outlet the flow was considered developed and the pressure was set to zero.

Simulations were performed considering a working fluid with transport properties similar those of the blood during extracorporeal circulation. The simulations were performed using a flow rate of 4,5 l/min and the results compared with experimental data from Huebner (2003).

The SOLIDWORKS™ software was used to generate a simplified virtual model of this filter. The filtering element) was removed and replaced by a simplified geometry (hollow frustum of cone). This simplification was necessary due to the complexity of the filtering element. Three different volumes were extracted: the region where the blood flows before the filtration, the simplified filtering element and the region where the filtered blood flows. A small extension was added to the entrance and exit connectors, attempting to approximate the real conditions of the flow. The volumes were exported to the ANSYS™ Workbench which generated the meshes of each part of domain. The software utilized to the numerical simulation was the CFX™ V11.0. The CFX™ is an integrated system of simulation. It is constituted by three modules: the first one is the pre-processing, the second processing module, and the last one is the post-processing. The pre-processing module is responsible for the definition of all physics involved in the simulation, as the sort of fluid or boundary conditions definition. The processing module, known as solver, calculates matrices referring to each equation. Further, there is the analysis of results made by the post processing module. The figure 2 shows an overview of the mesh at the surface of domain.

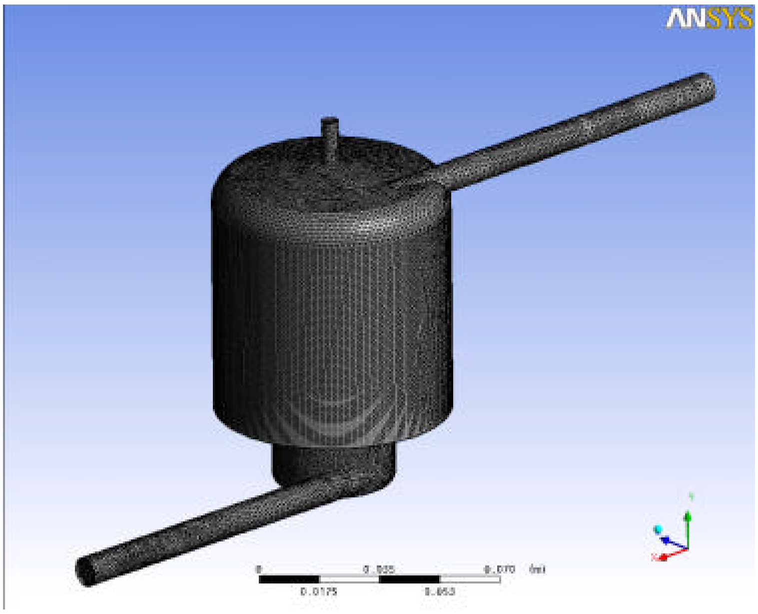

Figure 2. Overview of discretized domain

Figure 3 shows details of the mesh. At the left side, it can be seen the detail of the attachment of inlet connector with the filter body. At the right side, it is possible to notice the fine grid near the wall surrounding the exit connector. This fine grid was applied in the generation of every solid surface of domain, making the pressure drop more accurate.

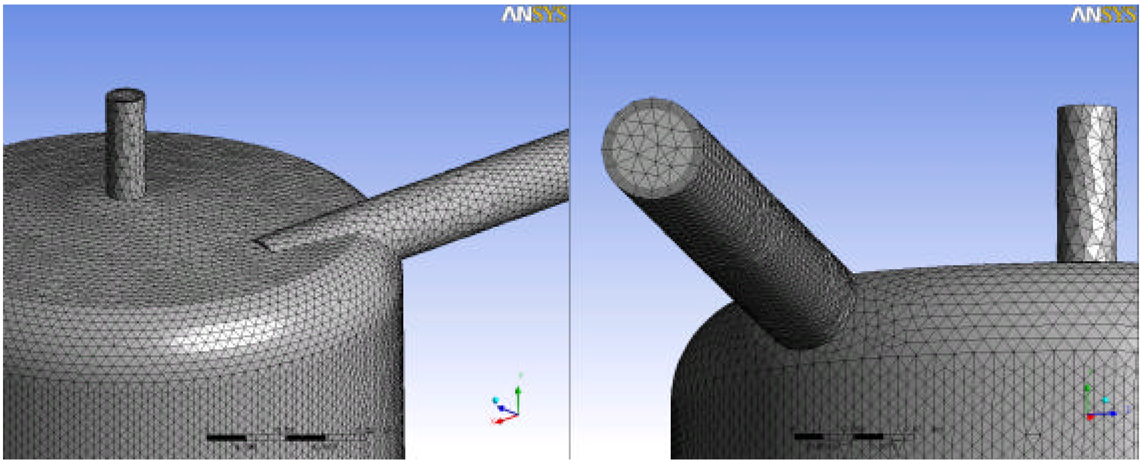

Figure 3. Mesh details

A convergence criterion of 10^{-3} was adopted to the root mean square of the residual (rms) considering all domain. The simulations utilized a domain of 440.000 nodes.

3. RESULTS

Figure 4 shows flow streamlines inside the filter and velocity vectors generated on a horizontal plane crossing the entrance connector at the middle.

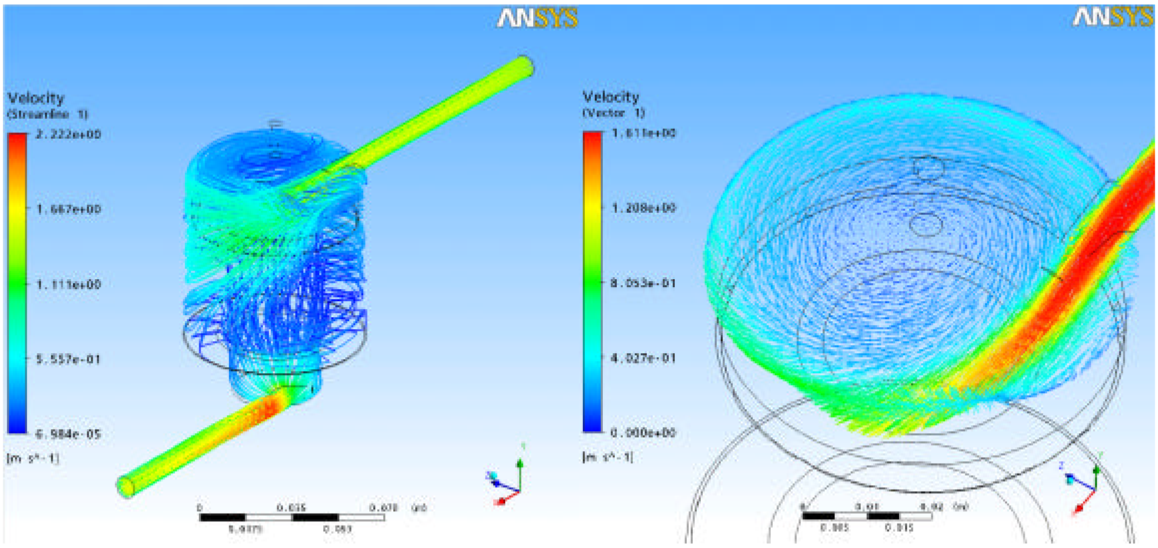

Figure 4. Streamlines along filter and velocity vectors on a horizontal plane

The flow passes through the entrance connector at the filters upper part, hitting the opposite wall. At this point the main flow moves downward, clockwise. It is noticeable a flow division at the nearest region of the exit connector where a part of it moves clockwise. These results follow the pattern obtained on visualization tests with coloring injection made by Huebner [2003], shown on figure 5.

Figure 5. Top view showing the flow direction and the division at the exit connector neighborhood (Huebner, 2003)

Figure 6 shows velocity vectors on an orthogonal plane (denoted N-S) and a plane parallel to the entrance connector (denoted E-O), both at the center line of filter´s body. At N-S plane, it can be seen a presence of a jet coming from the entrance connector and a flow pattern moving clockwise inside the filter. At E-O plane, on the entrance connector´s opposite wall, it is noticed a flow direction change and a splitting at the inner bottom.

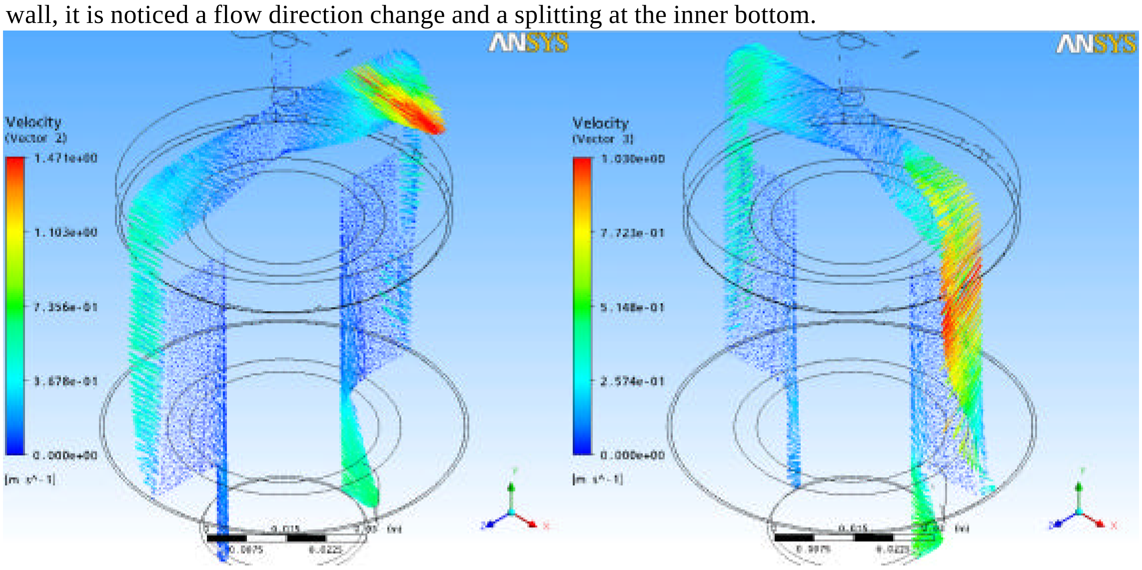

Figure 6. Velocity vectors on planes N-S and E-O

Figure 7 shows a detailed view of the velocity vectors on N-S plane and flow visualization at the same region.

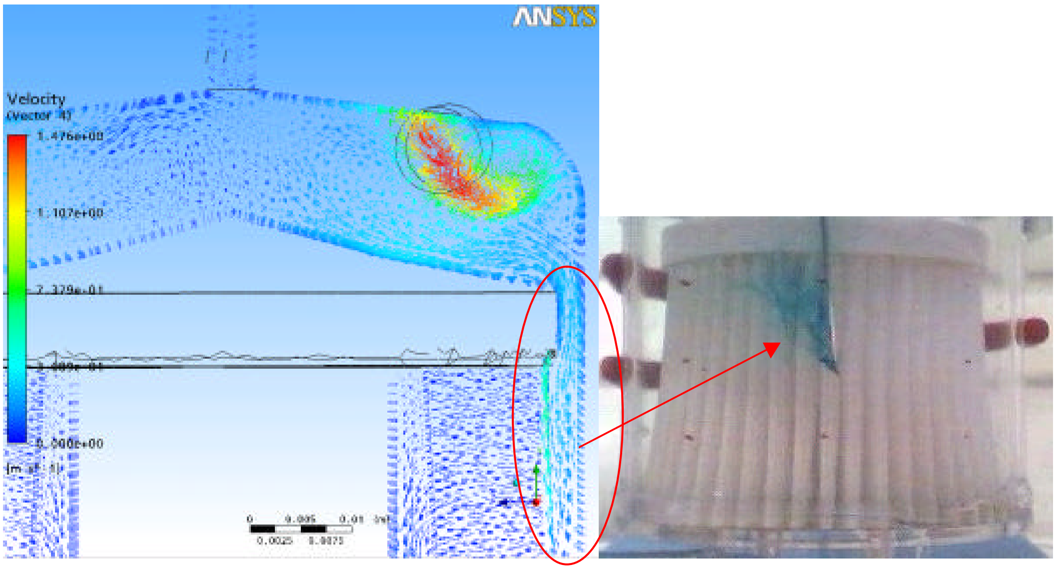

Figure 7. Flow details on N-S plane.

On N-S plane it can be seen an ascendant flux on the closest region of the entrance connector. This phenomenon is clearly noticed on the coloring visualization at the same plane. At the central plane, next to the bubble trap, it is seen the presence of a recirculating zone favoring the bubbles capture. On N-S plane it can be observed a jet coming from the filter´s entrance connector.

Figure 8 shows pressure variation along the plane localized on the filter´s upper region. The plane was positioned along coordinate Y taking the entrance connector´s line as center basis. The pressure gradient shows a maximum value near the entrance and the space of each color indicates a constant variation. Therefore, the flow is fully developed in agree with the experimental data introduced by Huebner [2003].

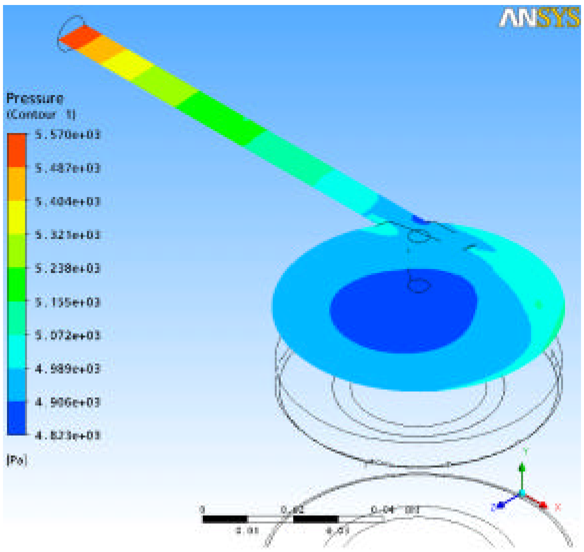

Figure 8. Pressure field on upper plane.

Figure 9 shows the pressure variation along planes N-S and E-O, respectively. On both planes it is observed that the pressure is greater at the region near the wall and it decreases at filter´s inner region. The presence of the filtering element can be easily seen by the uniform pressure variation at its region.

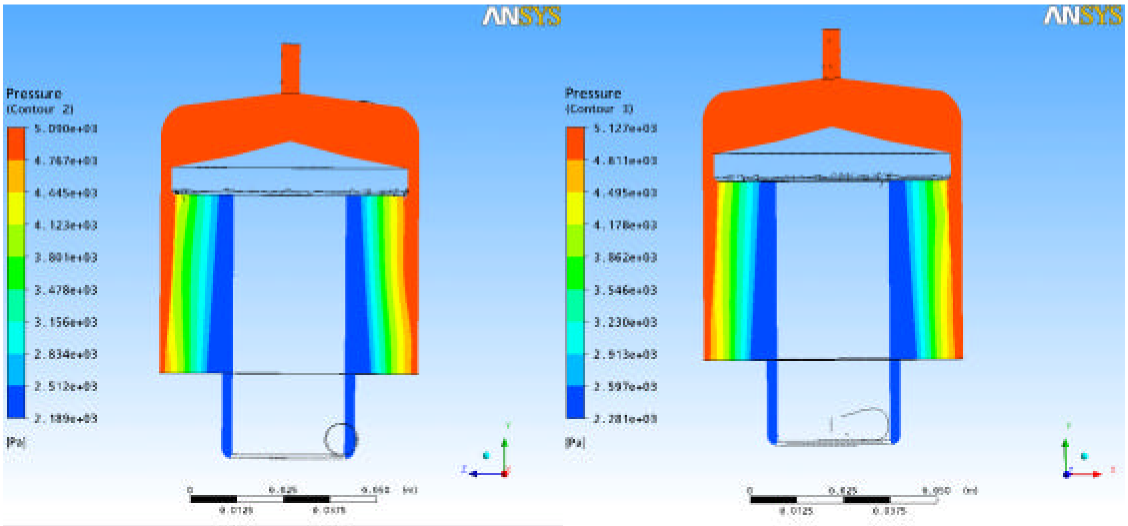

Figure 9. Pressure fields on planes N-S and E-O, respectively

Figure 10 shows kinetic energy distribution and dissipation on upper plane. The turbulent kinetic energy maximum value occurs at the region where the blast coming from the entrance hits the filter´s wall, forcing a flow´s change of direction. The dissipation has a maximum value at the entrance connector´s neighborhood where the flow is exposed to an expansion.

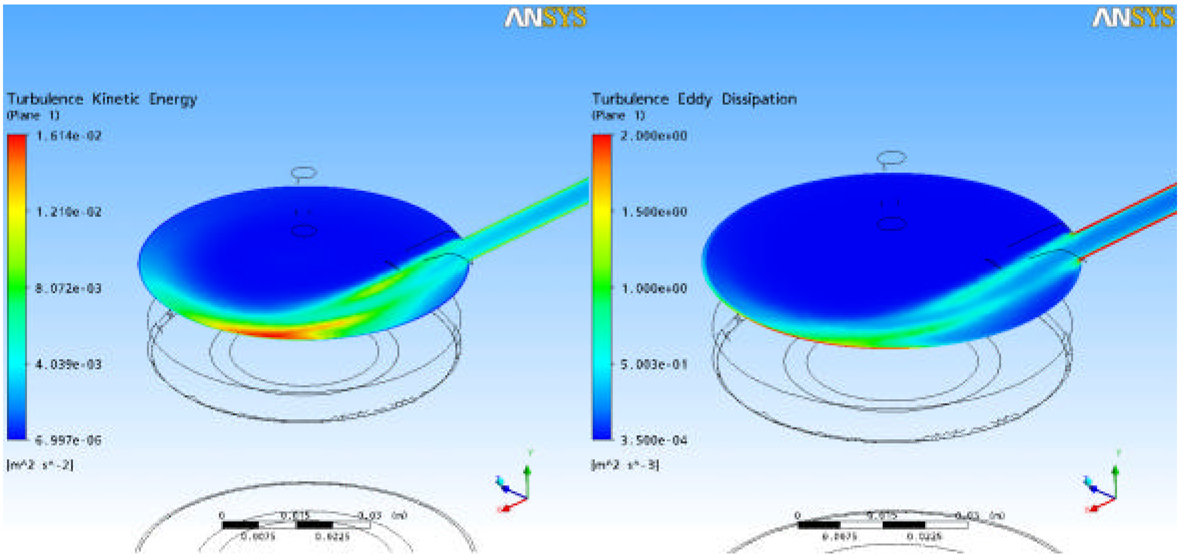

Figure 10. Turbulent kinetic energy and dissipation, on top plane.

Similar results were obtained with an outflow of 1,5 and 3,0 l/min. Table 2 shows values of pressure drop obtained numerically and the percent variation compared with the results introduced by Huebner[2003].

Table 1. Numeric and experimental values of pressure drop

Outflow (l/min)	Numeric(Pa)	Experimental (Pa)	Difference percent
1,5	1081	1039,65	3,98
3,0	2778	2777,75	0,01
4,5	5556	5443,54	2,07

The values shown indicates a maximum difference of 3,98% between numerical and experimental results. These results can be improved with a better correction of coefficients used on Darcy´s Law to model the pressure drop at the porous media. The results introduced along this work indicates that despite the difference between the numerical and experimental pressure drop the behavior of field flow follows the pattern obtained by Huebner [2003].

4. CONCLUSION

The present work presents results obtained from a numeric simulation of flow field in a arterial blood filter used in extracorporeal circulation surgeries. The turbulence model utilized was the K-**?** of two equations due to its robustness and convergence facility. The results obtained presents good agreement with data found on literature and the differences observed can be reduced by the adjust of Darcy law´s constant, utilized to simulate pressure drop on the filtering element which consists of a polyester grid.

5. ACKNOWLEDGEMENTS

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