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3D COMPUTATIONAL ANALYSIS OF BLOOD FLOW IN CENTRAL VENOUS CATHETERS FOR PATIENTS SUBMITTED TO HEMODIALYSIS TREATMENT

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Abstract. *Hemodialysis is the main technique used for the treatment of chronic renal diseases. In the market, there are several models of equipment and vascular accesses with their peculiarities, being the central venous catheter (CVC) one of the most used. These are presented with great variability of models available for the application, making it necessary to identify the particularities of each model and the procedures used. One way to evaluate such differences is numerical simulation, a tool capable of performing a virtual analysis, non-invasive, of the physiological and mechanical processes. In this study, 3D numerical simulations of blood flow were run with the software ANSYS Fluent, using the blood as a non-Newtonian fluid. Through the simulations, the velocity and shear stress profiles were generated for three different types of CVCs with the intention of predicting potential geometries generators of cell lysis and platelet activation, the main consequences of mechanical injuries in the blood.*

Keywords: *Central venous catheter, hemodialysis, blood flow, shear stress*

1. INTRODUCTION

The hemodialysis procedure is widely used for the treatment of chronic kidney patients and, despite being invasive and subject to complications, it is often the only therapeutic option capable of prolonging and improving the quality of life of patients (L. A. Carvalho and Borges, 2010). It is a procedure in which blood filtration, usually performed by the kidneys, is done externally, via equipment. For this, a vascular access is inserted that will lead the patient's systemic circulation to an extracorporeal circulation equipment. In this equipment, the dialysis process occur: impurities and electrolytes are removed from the blood and then the dialyzed blood is returned to the patient via vascular access (Mitzner, S.R. et al., 2001). Among the main problems identified in this treatment, the main one is the difficulty in maintaining a functional vascular access in patients submitted to the technique in a prolonged way (Ash, 2007).

One of the most common vascular access options to perform the treatment is the Central Venous Catheter (CVC), which consists of implanting a catheter into a large caliber vein of the chest or upper arm (Guimarães et al., 2016). According to Carlotti (2012), the establishment in a certain access route occurs according to the preference of those who perform the procedure, observing certain circumstances such as: age; patient's condition; and type of therapy. In the case of hemodialysis, there is a preference for the jugular, subclavian and femoral veins for CVC implantation.

On the market, there are several options for CVC models available for application in the hemodialysis procedure. According to Clark et al. (2012), this device operates with high flow and, therefore, generates high shear rates, generating hemolysis, which is the breaking (lysis) of the red blood cells. Hemolysis is the main consequence of the mechanical stress generated by cardiopulmonary bypass, and can be identified through the presence of free hemoglobin in the plasma and a decrease in haptoglobin (protein responsible for capturing free hemoglobin in post-hemolysis blood to prevent the loss of iron by the body) during and after the procedure (Vieira Junior et al., 2013). Another serious consequence, according to Clark et al. (2015), is the platelet activation that leads to the formation of clots. Both are due to the increase in shear stress. Moreover, Paul et al. (2003) indicates the occurrence of lysis of red blood cells when there are submitted to shear stress above 30 Pa, while Clark et al. (2012) points out the platelet activation for shear stress higher than 10 Pa.

From a mechanical point of view, human blood is a suspension of particles - mainly white blood cells, erythrocytes, platelets and proteins - in plasma (liquid phase) and, therefore, has complex rheological characteristics. The hematocrit (portion of red blood cells that constitutes the blood) can be considered the main influence to the rheological behavior of the blood, since it makes up about 40% of blood volume in normal adults (Ameenuddin et al., 2019). In this way, the fluid viscoelasticity should not be neglected without prior analysis.

In this paper, a 3D numerical model was developed to evaluate how catheter geometries used in hemodialysis influence on hemolysis and platelet activation in patients undergoing treatment in steady state. The numerical model also considered the blood flow when the catheter is positioned in the subclavian, internal jugular, brachycephalic and upper vena cava veins. It is intended that, with this work, it can be verified how different catheter geometries affect blood. To characterize such effect, shear stress and strain rate were calculated to provide indicators on the choices to be made when using specific models to the detriment of others.

2. METHODOLOGY

The numerical simulation was performed using the ANSYS computational fluid dynamics (CFD) software. The governing equations solved were:

Mass conservation:

$$\frac{\partial \rho}{\partial t} + \vec{\nabla} \cdot (\rho \vec{u}) = 0 \tag{1}$$

where ρ corresponds to specific mass, t to time and $\vec{u}$ the velocity vector

Momentum Conservation:

$$\frac{\partial\, (\rho \vec{u})}{\partial t} + \vec{\nabla} \cdot (\rho \vec{u} \vec{u}^{t}) = \vec{\nabla} \cdot (\overline{\overline{\sigma}}) + \vec{B} \tag{2}$$

where $\vec{B}$ represents the body forces and $\overline{\overline{\sigma}}$ is stress tensor defined as:

$$\overline{\overline{\sigma}} = \overline{\overline{\tau}} - P\overline{\overline{I}} \tag{3}$$

Where $\overline{\overline{\tau}}$ is the deviatoric part of the tensor and P is the pressure.

The schemes employed in the simulation are shown in Tab 1 and Tab. 2.

Table 1 – Numerical Schemes used to solve the terms of the conservation equations.

Solution Methods	
Scheme	SIMPLE
Gradient	Least Squares Cell Bases
Pressure	Second Order
Momentum	Second Order Upwind

Table 2 – Relaxation coefficients.

Solution Controls	
Pressure	0.3
Density	1
Body Forces	1
Momentum	0.7

2.1 Geometry

The geometry of the circulatory system was morphologically based on Netter (2000) and dimensionally based on the mean diameter values cited by Marino (2015). The model proposed in Fig. 1a was then built in the Geometry module of the ANSYS software. The circulatory system geometry was considered to facilitate the evaluation of its influence on the catheter flow.

The catheters can differ in the position and geometry of the input and output lumens. These geometric variations lead to changes in flow and, consequently, changes in the shear rate, which can potentially lead to cell lysis and platelet activation. The study of catheters was done according to the dimensions shown in Tab. 3.

Table 3 - Characteristic dimensions of hemodialysis catheters. Adapted from: Marino, 2015

External Diameter[m]	Length [m]	Lumen	Lumen Effective Diameter [m]	Maximum flow [m³/s]
0.004	0.16	Inlet	0.002	6.58E-06
		Outlet	0.002	4.83E-06

Based on the dimensioning described by Ash (2008) and by Atherikul et al. (1998), the geometries expressed in Fig. 1b were defined. The insertion of the catheters flow direction is shown in Fig. 1c.

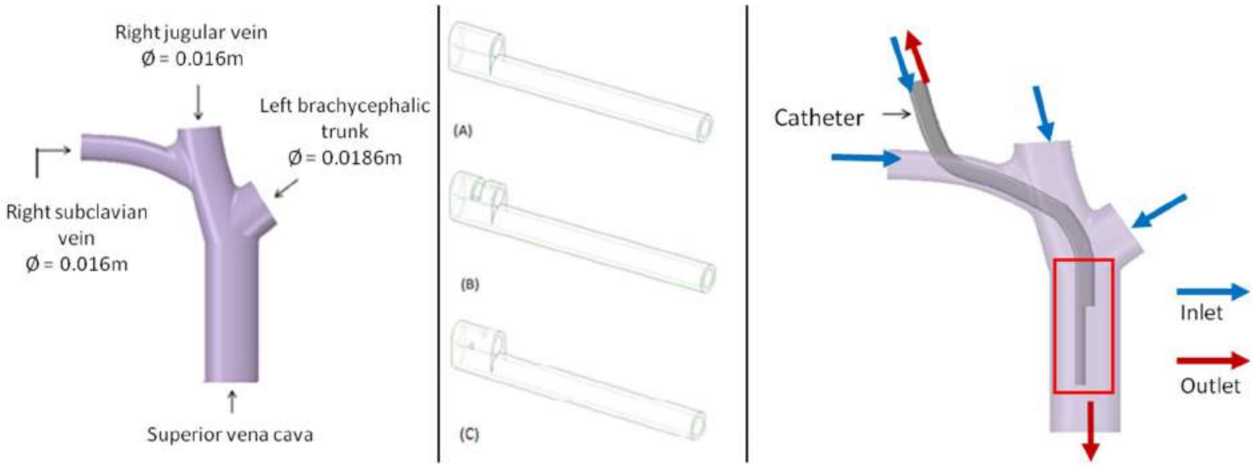

Figure 1: (a) Proposed numerical model. (b) Geometry of catheter tips. (c) Insertion of the catheters and flow direction (boundary condition).

2.2 Blood rheological modeling

Rheological modeling is of fundamental importance for computational evaluation. For that, there are several models available in the literature according to the application. For this study, the Carreau model described by Lamothe et al., 2018, was chosen to represent the fluid, taking into account the non-Newtonian characterization of the flow in the venous system:

$$\mu = \mu_\infty + (\mu_0 - \mu_\infty)[1 + (\lambda\dot{\gamma})^2]^{\frac{(n-1)}{2}} \tag{4}$$

with μ_∞ = 0.0035 Pa.s, μ_0 = 0.056 Pa.s, λ = 3.313s, and n = 0.3568, where μ is the viscosity of the fluid, dependent on the shear rate $\dot{\gamma}$, μ_∞ the viscosity at high shear rates, μ_0 the viscosity at low shear rates, λ is a time constant that represents the transition range in which the viscosity stops being constant and begins decreasing, and is the flow rate (or power law index).

This rheological selection was made because, at low velocities (venous circulation), this model is capable of providing the non-Newtonian effect on flow and, at high velocities occurring in catheters (extracorporeal circulation), it is close to the Newtonian model commonly used in studies (Karimi et al., 2014).

2.3 Mesh

An unstructured mesh was produced using the Mesh module of the ANSYS software, with elements of a linear order from the initial size of 0.0004 m, which represents 10% of the diameter of the catheter tips under study. The mesh convergence was verified according to the methodology suggested by Roache (1994) and the specifications selected to generate the meshes are shown in Tab. 4.

Table 4 – Mesh specifications

Mesh	
Body Sizing	
Type	Element Size
Element Size	3 e-4 m
Size Function	Uniform
Behavior	Soft
Growth Rate	1.20

2.4 Bondary conditions

Non-slip condition was adopted on the walls, velocity profile at the inlet of the veins and the catheter, and percentage of flow at the outlets. The velocity entry profiles were constructed by developing the average velocities (described by Marino, 2015 for patients at rest) in tubes long enough to obtain a developed the flow, observing the entrance length for each case (subclavian, jugular, brachycephalic and catheter). Subsequently, the profiles established by the flow were exported as the before mentioned boundary condition of normal entry velocity to the surface for their respective geometric counterparts.

3. RESULTS AND DISCUSSION

3.1 Mesh Quality

The graph in Fig. 2 shows the maximum velocity results found in a velocity test completely developed in a long tube with dimensions of the subclavian vein, as well as Richardson's approximation.

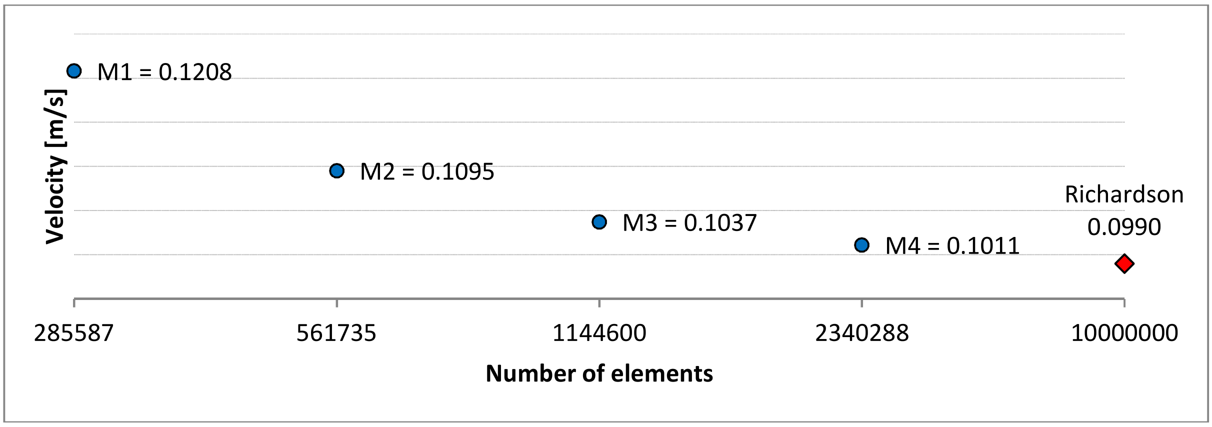

Figure 2 - Velocities by number of mesh elements and Richardson approximation.

In Fig. 2 it is possible to observe target velocity obtained by each tested mesh. An error of approximately 10% was estimated according to Richardson's extrapolation regarding the mesh M2 that proved convergence of the numerical method. The error obtained was considered satisfactory for the work proposal. Finally, the M3 mesh was selected in order to minimize computational costs, showing a numerical discrepancy of 4,7 % to Richardson's extrapolation.

3.2 Velocity field

The velocity fields on the circulatory system geometry shows the influence of each type of catheter tip to the local blood flow. The central plane of the proposed geometric model was chosen to evaluate the results. Figure 3a presents an overview of the velocity field of the proposed model for catheter A and Figure 3b shows the velocity profile on the scale of the blood flow at the circulatory for the same catheter A. Because there are no geometric variations in the catheter outlet, the velocity field is similar in all models, and just one geometry is brought.

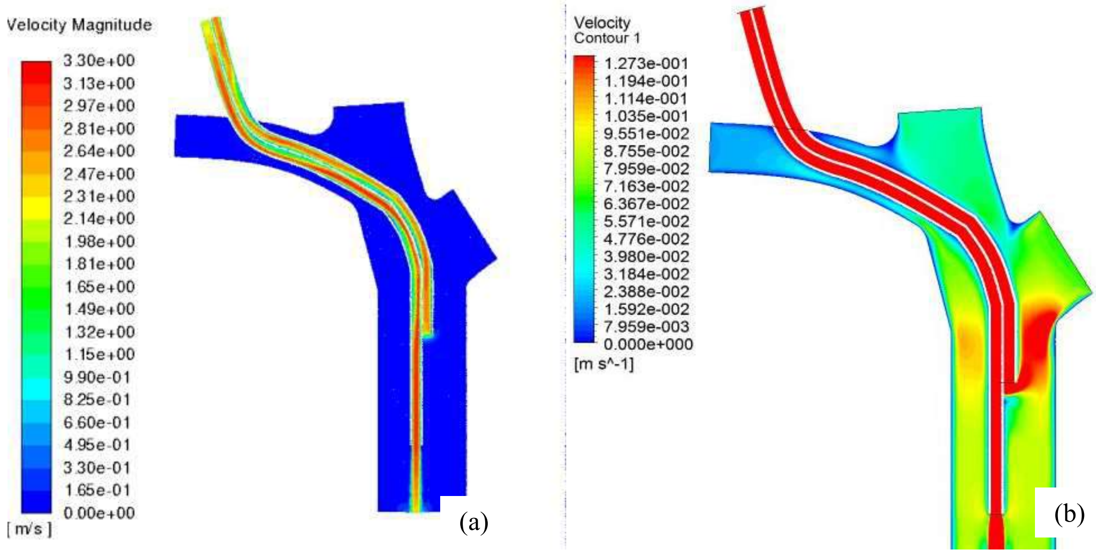

Figure 3 - (a) Velocity Field. (b) Velocity in the vessels.

From Fig. 3a we observe that the catheter flow does not interfere substantially with the circulatory system flow. This can also be viewed in another scale, as shows Fig. 3b, that presents a velocity distribution in the vessels 10 times lower than in the catheter.

Since the main problem resides on the catheter tips, especially on the entrance lumen, Fig. 4a, b and c, bring detail of the velocity field at the entrance of each catheter, at the tip of each chosen model, where the peculiarities of each geometry can be noticed.

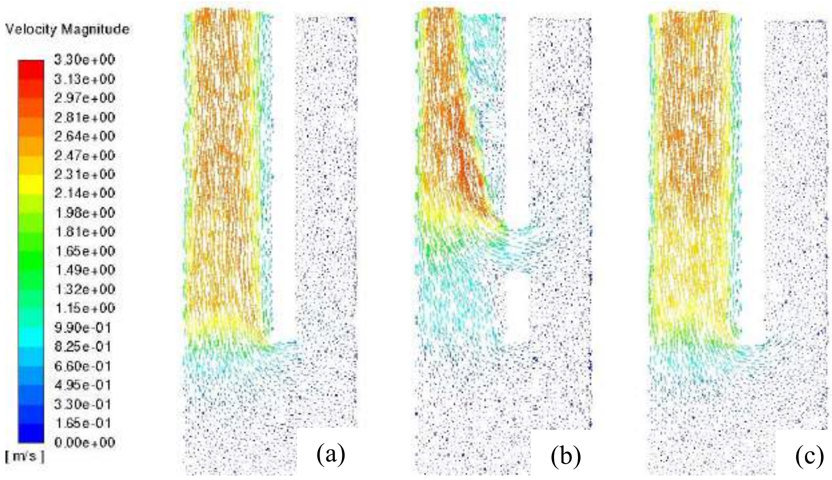

Figure 4 – Velocity vectors fields at the entrance of the catheters: (a) catheter A, (b) catheter B , (c) catheter C.

The maximum velocities achieved in each type of catheter are close, being 3.11 m/s, 3.09 m/s and 3.27 m/s for catheter models A, B and C, respectively. It is noticed in type A and C catheters that the flow is presented in a similar way, as a suction region, exhibiting smoothness of the velocity field in the initial section of entry of model C, however, assuming greater velocity after the lateral entrance openings. The type B catheter also presents low inlet velocity but exhibits high values of velocity next to the lateral opening, which, in turn, conducts the flow close to the catheter wall until the characteristic flow profile is established.

3.3 Shear Stress Field

To calculate the shear stress, three equations were inserted into ANSYS software to be processed during the simulation, namely: (i) the "strain rate" function (contained in the CFD package library); (ii) the Carreau model for calculating the dynamic viscosity, Eq (4); (iii) the function that defines the estimated shear stress τ_e as indicated by Clark et al. (2012).

$$\tau_e = \dot{\gamma}_e \, \mu_e \tag{5}$$

Figure 5, it is possible to observe the shear stress field at the exit of the catheters, which, as identified, were similar among all the tested geometries.

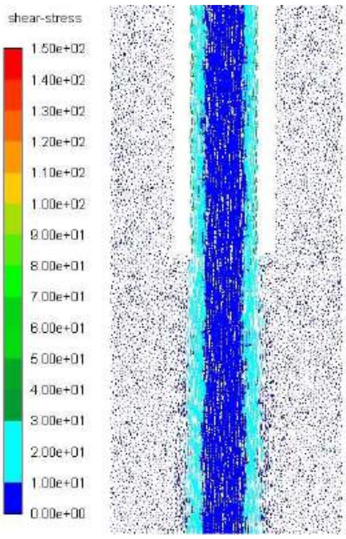

Figure 5 - Shear stress profile presented at the catheter outlets.

It is observed that, even with high discharge velocities, there are no high shear stresses at the exit of the catheters ($\tau_e < 30$ Pa). Figures 6 a, b and c it is possible to see the shear stress field at the entrance of each catheter model.

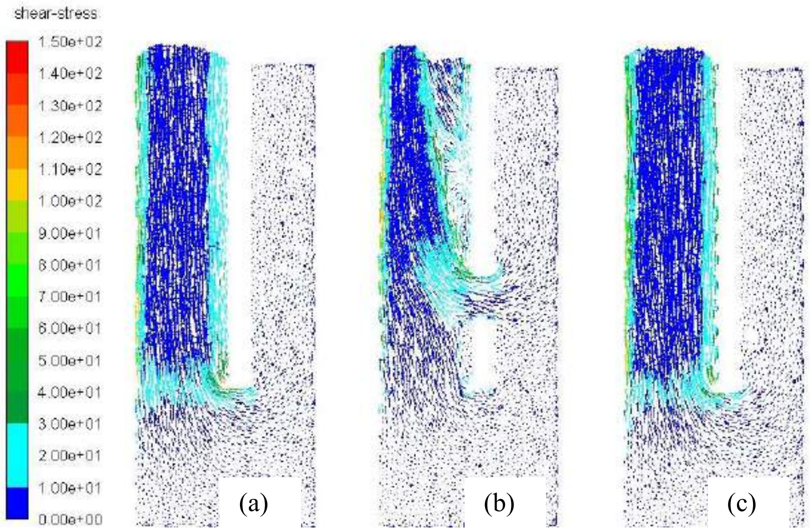

Figura 6 – Shear stress: (a) catheter A, (b) catheter B e (c) catheter C.

In all cases, the values reached for shear stress exceed the limits for the occurrence of hemolysis (above 30 Pa) and platelet activation (above 10 Pa). The flow region subjected to this tension, however, varies according to geometries. The maximum shear stresses in the flow of the catheters were approximately 137 Pa in catheter A and 152 Pa for catheters B and C, maintaining the same order of magnitude for the three cases. High shear stress is perceived at the catheter walls (boundary layer). However, in Fig. 6b, it is possible to observe a larger area close to the opening, which has a high shear stress (≈ 122 Pa) due to the flow generated by the lateral opening.

Figure 7 shows the same shear stress field in a perpendicular cut to highlight the influence of the lateral openings.

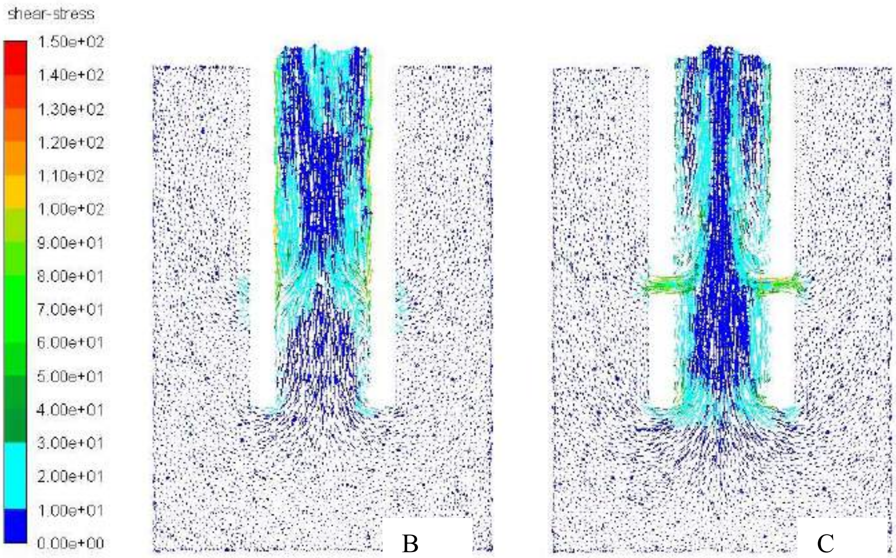

Figure 7 - Details of the shear rate in the lateral openings of the catheters B and C.

From Fig. 7 it is possible to notice that the model C also offers a considerable region of high stresses (≈ 80 Pa) at the side entrances.

4. CONCLUSION

In this paper, a three-dimensional model of blood flow in the thoracic region was generated, in order to computationally test the influence of the geometry of commercial catheters for hemodialysis on blood flow. This study had the objective to identify potentially harmful geometries for patients undergoing treatment.

Based on the analysis of the results, it can be concluded that:

• the blood flow on the circulatory system brought in the present study is not influenced by the presence of the catheter;

• velocity fields at the exit lumen of each catheter are not influenced by the flow at their entrance lumen;

• velocity and shear stress fields at the entrance lumen affected by different catheter geometries, as expected, and can show significant variation;

• models with lateral entrances can develop flow regions with increased velocity and shear stress.

Another relevant factor observed is that the greater shear stress in catheters with lateral openings can lead to greater platelet activation which, in turn, can generate clots and cause the obstruction of these orifices or even the catheter lumen. However, for in vivo use, vascular access and catheter positioning in the circulatory system should also be considered. Based on these considerations, the use of catheters with a greater number of entry holes may become relevant. It is then up to the health professionals to evaluate and select the catheters to be used in each patient, giving preference to the catheter tips that generate less shear stress when possible, avoiding tips with lateral openings, as they are less advisable, from a mechanical point of view.

For future works, the group intends to further evaluate the type B catheter, exploring the geometry regarding the lateral opening region. Mesh refinement should be further pursued to ensure mesh growth rate below 6% and guarantee excellent quality results for the specific region of interest. Another point of interest is to evaluate the time of blood permanence, that is, the time that the blood cells are exposed to the shear stress in a certain place. The permanence time and coupled with the information of the local shear stress would make possible to calculate the hemolysis indexes and platelet activation, values that are directly explored by the medical community.

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6. RESPONSIBILITY NOTICE

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