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Numerical Based Analysis of Hematocit Effect in Stented Coronary Artery

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Abstract: The problem of the non-Newtonian fluid flow in the complex geometries (pipes), whose domain of application is very varied, is a topic of interest to both physicists, industrialists and medical doctors. Various studies have been made ??on cardiovascular problems and more precisely the effect of intra-vascular prosthesis on the stat of arteries. Indeed, obstruction of the arteries, whether coronary or carotid known as stenosis, represents a great danger to the people who are affected. It modifies the hemodynamic and consequently generates a detrimental effect on nutrition and oxygenation of organs (heart, kidney and brain..). Among the medical intravascular interventions this stenosis is referred to as stenting. The stent is a flexible and deformable metal object used to straighten the obstructed artery and keep it open. The presence of such an object into the blood vessel disrupts the flow of blood. In this regards, this paper is a numerical study to show how the presence of the stent in an artery influences the blood flow. In this work, the blood flow is considered as a Newtonian and non-Newtonian fluid whose behavior is described by the mathematical Carreau's models. The hematocrit variation ratio is in the range of 30 to 55% which cover the rate on anemia and normal levels. The presence of the stent has a very low effect on speed and pressure, however it decreases the overall shear stresses which could favors and promote restenosis.

Key words: Stent · Artery · Numerical Simulation · Blood · Non-Newtonian

INTRODUCTION

Numerical simulations have the advantage of testing parameters that are difficult to measure directly *in vivo* or *in vitro*, such as shear stress gradients at the wall and the pressure. Blood is actually a non-Newtonian fluid; the apparent viscosity depends not only on the fluid properties, but also on the flow conditions. It consists of plasma containing red blood cells, white blood cells and platelets. Blood flow, its interaction with the arterial walls, stenosed arteries and their repair techniques have been the subject of various numerical and experimental studies [1].

L. Achab [2] has modelled the rheological behavior of the blood using Carreau, Cross and Casson's constitutive laws giving the variation of the shear stress as a function of shear rate to approximate the experimental curve. The results obtained using the Carreau's model is in good agreement with the experimental results. This shows that blood viscosity increases strongly with the hematocrit and undergoes a rapid decrease when the shear rate varies from 0 to 20, whatever the rate of hematocrit is and becomes nearly constant beyond this range.

Several studies have been conducted on intravascular prostheses, in order to solve the problems faced by surgeons, such as complications that occur after the operation and to offer solutions such as the optimization of stent geometry [3-7].

A number of researchers have been interested only in the structural mechanics i.e. the location of the stent and its response under the effect of balloon expansion [8-12]. Other researchers have attempted to explain the phenomenon of intimal proliferation under the effect of the variation of wall shear stress [13-14].

Mickaël Gay *et al.* [15] have used the finite element method to study the velocity profile and the Von Mises stress distribution in the stent during implantation. They have concluded that the obtained results will boost the

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development of new stent designs and deployment protocols of stent in order to minimize vascular injuries during its placement and reduce restenosis. Mickaël Gay *et al,* had focused their observations on other works. They concluded that coronary stents are widely used, because the rate of restenosis is reduced to 20 up to 30% compared to balloon angioplasty [16-18]. Benard *et al.* [19] have made an experimental study of the laminar flow of blood through an artery treated by a stent implantation. They observed that in a zone of weaker shear stress (SS) the endothelial stimulation favors the restenosis. The authors concluded that since the endothelial proliferation is highly influenced by the shear stresses (SS), their knowledge of their variation induced by the stent implementation can be important for optimizing the design of prosthesis.

George *et al.* [20] studied numerically the effect stent implanted in the aortic artery on the hemodynamic factors. They observed that a high blood pressure and a strong systolic slope of the pressure wave can lead to a high drag force on a graft-stent compared to the one generated from high blood viscosity. Balossino *et al*. [21] studied the effects of different models of stent after their expansion on the blood flow under pulsatile physiological conditions. They noticed that the maximum parietal shear stress (WSS) is higher on the stent than that on the artery wall.

The objective of the present study is a numerical investigation of the combined effects of hematocrit with the presence of a stent in an artery on the blood flow.

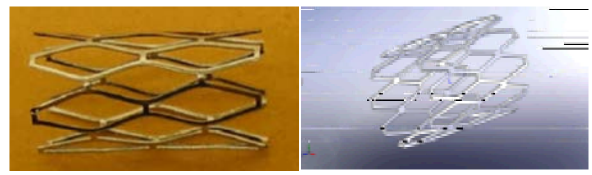
Numerical Simulation: This numerical study examines the effect of a stent in an artery on the parameters of the blood flow is considered as Newtonian and non-Newtonian. In the non-Newtonian blood behavior is described by the mathematical Carreau's model where the hematocrit is assumed to vary in the range from 30 to 55% covering rate of disease (anemia) and normal [3].

Geometrical Configurations: The stent studied in this works (Fig.1), type Palmaz-Schatz stent micro-PS154 [5], a length of 10.6 mm, an outer diameter of 2.9 mm and a thickness of 0.1mm is placed against its deformation in the middle a coronary artery cylindrical of 60 mm in length.

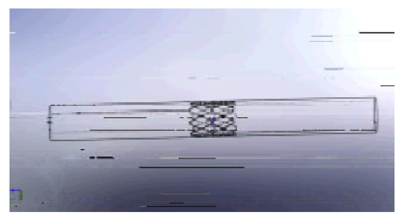
Boundary Condition: To numerically simulate blood flow in a stented artery, we imposed a maximum speed (0.25 m / s) at the inlet and static pressure of 14000 Pa to the output (Fig. 2).

Table 1: Blood properties

	Non-Newtonian Fluid (Carreau)		
Newtonian Fluid		*He* 30%	*He* 55%
ρ =1050 Kg/m^3	μ_o*(mpas)*	13.26	74.54
	μ_∞*(mpas)*	3.29	5.71
	β	1.25	0.946
	p	0.527	0.501
	Dis(%)	0.221	0.182
μ= 0,0035Pa.s	*Coef Teil*	0.0183	0.0714

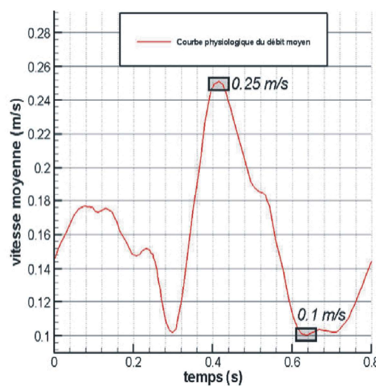


(a)

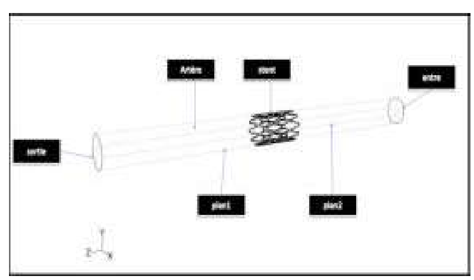


(b)

Fig. 1: a) Stent, b) artery



(a)



(b)

Fig. 2: a) Average physiological velocity in a coronary artery, b) domain boundary .

Mesh: Due to the complexity of the stent geometry considered an unstructured mesh is used. Therefore, the 3D domains were meshed with tetrahedral Hybrid

Fig. 3: Unstructured mesh (Tet/Hybrid).

elements. The mesh was refined near the stent (Fig.3). The commercial Fluent code is used for the finite volume analysis.

Mathematical Equations: The laminar flow of blood, considered incompressible, is governed by the Navier-stockes' equations and conservation of mass:

$$\rho\left(\frac{\partial v}{\partial t}+v.\nabla v\right)=-\nabla p+\mu\nabla^2 v \tag{1}$$

$$\nabla.v=0 \tag{2}$$

where v = velocity, p = pressure, μ = viscosity and ρ = density.

The blood properties used are given in Table [1]:

$$\tau=\left[\mu_\infty+\frac{\mu_o-\mu_\infty}{\left(1+\beta^2\dot{\gamma}^2\right)^2}\right]\dot{\gamma} \tag{3}$$

RESULTS AND INTERPRETATION

Numerical calculations are carried out in 3-D, but only the results corresponding to the vertical plan (Fig.2.b) are shown. In this plan, appear small square ribs of the same size.

The axial velocity is calculated upstream (Line 1), on rib (line 2) and downstream (line3) of the last rib of the vertical plane (Fig.4).

The axial velocity of the blood changes were shown in Figures 5, 6 and 7 which is considered Newtonian and non-Newtonian with different hematocrit values, respectively along the lines 1, 2 and 3. It is noticed that the axial velocity profile of the blood and assumed Newtonian with a hematocrit of 30% are identical and that the speed increases with the hematocrit in the central area of the artery and decreased in the vicinity of wall. In the area near the wall, at the radial abscissa given the velocity

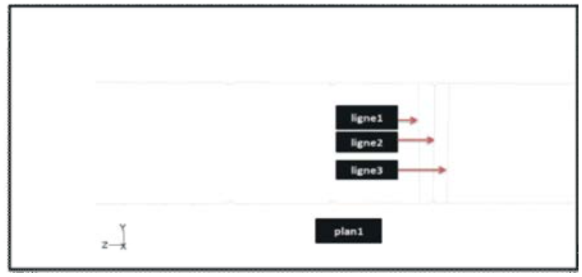


Fig. 4: Plan1 (with 3 ribs of the same size (square))

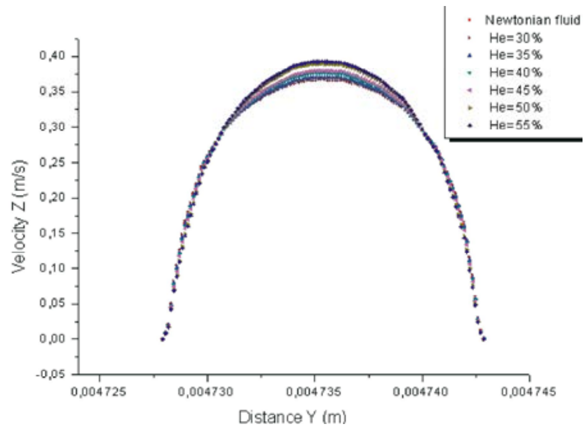


Fig. 5: Evolutions of the axial velocity along the vertical line 1 for different rate of hematocrit

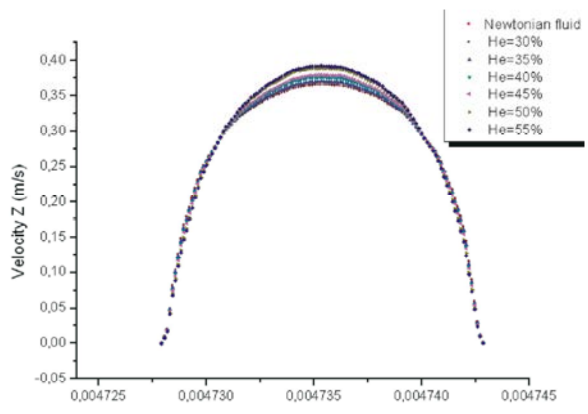


Fig. 6: Evolutions of the axial velocity along the vertical line 2 for different rate of hematocrit

gradient must decrease with increasing of hematocrit rate, as the blood viscosity increases with the latter which leads to the slowdown in fluid layers between each other. In the central area, radial abscissa fixed, the shear rate increases with the hematocrit rate which makes the blood more fluid.

Figure 8 represented the pressure distribution in the direction of the flow. It is observed a drop in pressure along the artery with the presence of small fluctuations at

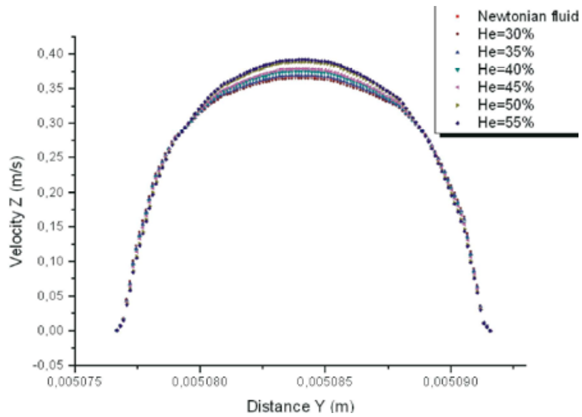


Fig. 7: Evolutions of the axial velocity along the vertical line 3 for different rate of hematocrit

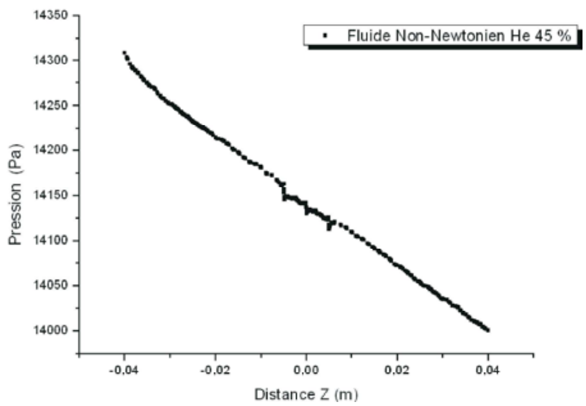


Fig. 8: Longitudinal evolutions of pressure for normal subject (He 45%).

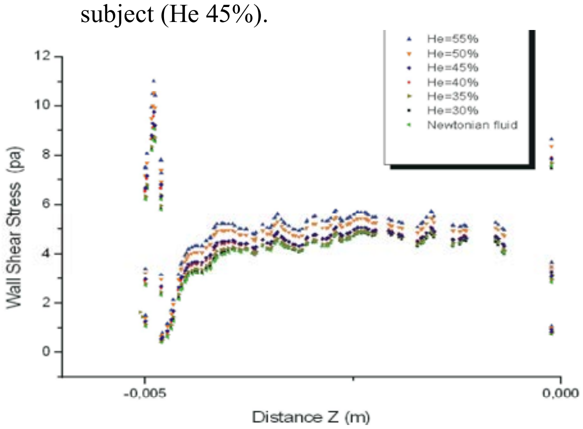


Fig. 9: The longitudinal evolutions of wall shear stress, in the vertical plan, between two consecutive ribs for different hematocrit.

the stented part. In general, one may say that the implantation of the stent does not have excessive influence on the pressure. This is an advantage for prosthesis.

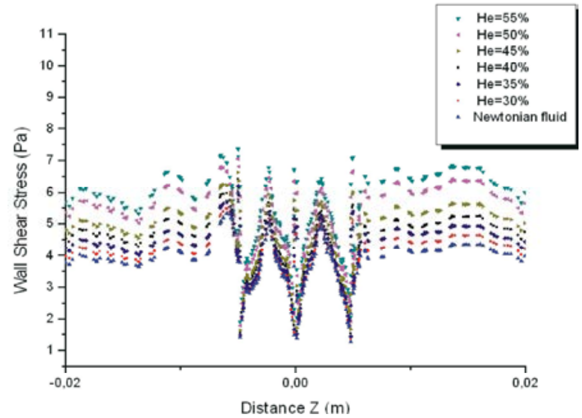


Fig. 10: Evolutions of the wall shear stress, in the vertical plan, along the entire artery

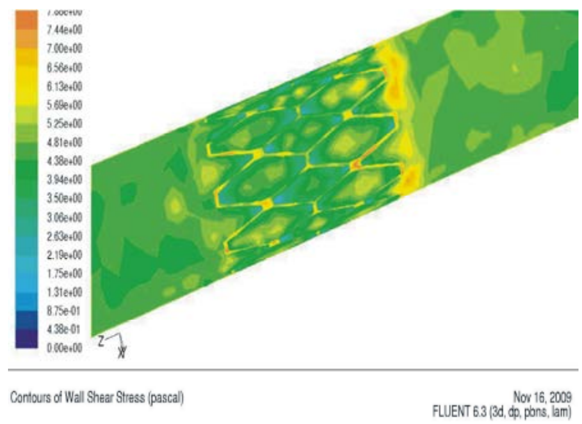


Fig. 11: Distribution of the wall shear stress

Figure 9 showed the evolution of the wall shear stress (WSS) between two consecutive ribs in the vertical plane, for different values ??of hematocrit.

The WSS increased with the hematocrit which explains the increase in the degree of blood deceleration with hematocrit in the wall vicinity. On the ribs, the WSS undergoes a peak amplitude much greater than elsewhere followed by a minimum. This peak is due to the impact of the flow on the ribs. In the space between the ribs, the WSS remains essentially constant for a given hematocrit.

Figure 10 showed the evolution of the wall shear stress in the vertical plane along the entire stented artery for the same hematocrit values. Along the stent, the WSS is fluctuating while it is generally constant and of the same value in the non-stented part. A high hematocrit values, the average value of the WSS in the stented portion is slightly lower than that of the smooth parts. The opposite phenomenon occurred for low values ??of hematocrit and Newtonian fluid.

In order to see the effect of the stent on the distribution of the WSS over the whole wall, a 3D color distribution is given in Figure 11. We noted that the WSS is greater on the metal part constituting the stent and just in its upstream and lower immediately upstream and downstream of the stent mesh conjunctions where the flow is lifted off. Within the mesh, the distribution of the WSS is similar to those in the non-stented parts of the artery.

CONCLUSION

The purpose of this study is to characterize the main features of blood flow through a stented artery considering the Newtonian and non-Newtonian behavior of blood increasingly marked by an increase in hematocrit.

The stent does not have a significant influence on the speed of blood flow due to its low thickness which is advantageous for the surgery but disturbs the distribution of wall stress without leading to an average value too different from those non stented parts of the artery.

The increase in hematocrit decreases the axial velocity of the blood in the vicinity of the arterial wall and increases it in the central part and makes increase the wall shear stress in general. In fact, more the non-Newtonian behavior of blood is marked (high hematocrit) more the blood is slowed down near the wall and accelerated further.

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