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Fluid-structure interaction in a free end textile vascular prosthesis

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Abstract. Textile cardiovascular prostheses are tubular structures made of polyester filaments. They present particular mechanical properties linked to wavy form of their walls allowing them to stretch under pressure. Pulsatile blood flow was studied in a moving walls vascular prosthesis. First, an image processing device was used to measure prosthesis displacement under air pressure in an free end impregnated textile prosthesis. Then, fluid-structure interaction is simulated with a numerical computation code allowing to couple prosthesis walls motion with blood flow. Navier-Stokes equations governing fluid flow are numerically solved with N3S code based on finite elements method. The numerical process is based on the Arbitrary Lagrangian Eulerian (ALE) formulation allowing moving domains. The obtained results showed a particular distribution of blood flow velocities and shear stress near the graft walls. The flow velocity distribution near a prosthetic surface is strongly influenced by the crimping morphology and deformation. A local flow analysis is imperative to understanding pathologies implying hemodynamic factors and to optimize the prosthesis design.

PACS. 47.11.+j Computational methods in fluid dynamics – 87.19.Tt Rheology of body fluids

1 Introduction

Vascular diseases, usually associated to the consumption of rich carbohydrates food, are more and more frequent in the industrialized countries. Graft implantation is today a common surgical procedure in the management of patients having severe blood circulation difficulties. During the last forty years, the textile technology brought several solutions for vascular surgery with the introduction of woven and knitted prostheses made of polyester filaments. Textile grafts are not yet perfect because of the complexity of arterial biology and textile mechanics. Biocompatibility has been achieved but the problems of compliance and resistance to the blood flow remain. The first generation of cardiovascular textile prostheses made of hand-sewn woven structures demonstrated some difficulties after implantation because of a lack of compressional resistance and a tendency to kink. Today, crimped textile implants made of polyester share the market with those moulded in one single piece of PTFE. Compared to the flat shape of moulded grafts, the crimping of textile vascular prostheses showed several mechanical advantages. In fact, crimping (Fig. 1) was achieved by fixation of an "accordian" pleat deformation permitting the surgeon to control the longitudinal tension and improving the resistance to kinking. Besides, knitted and weaved structures showed better aptitude for sutures than moulded ones. For these reasons, textile implants are exclusively used in particular sites needing long grafts with satisfactory bending properties like femoral bypass at the knee level. In previous studies [2,3], we showed that flow velocity near a prosthetic surface is influenced by the morphology of crimping and correlations between flow nature and thrombosis progression was established.

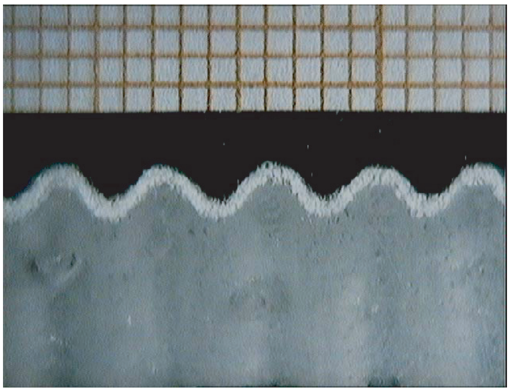

Fig. 1. Prosthesis crimping shape.

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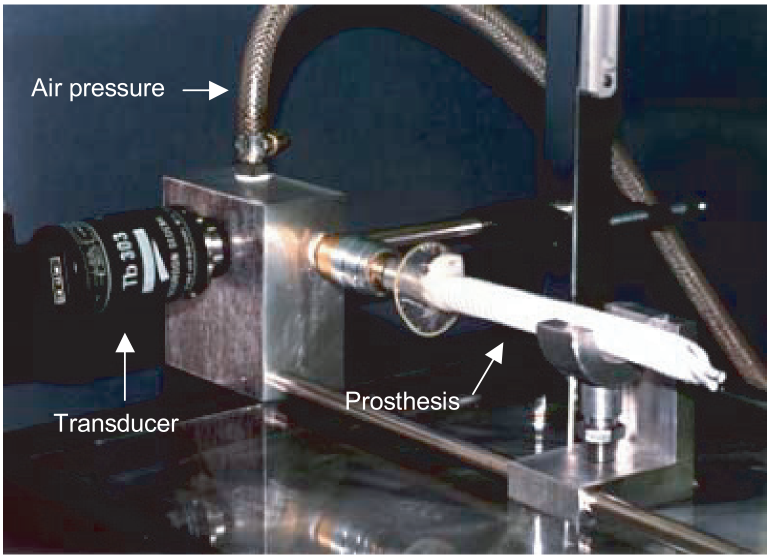

Fig. 2. Prosthesis deformation device.

In literature, the deformation of the natural artery have been widely investigated. Most studies [1,5,9], considered that arterial tissue is a perfectly elastic material. They showed that pressure inside natural grafts induced only an augmentation of the diameter of the elastic natural artery and no longitudinal stretching was observed. Meanwhile, studies on mechanical behavior of textile vascular prostheses were extremely rare. Different kinds of crimping, depending on their circular or helicoidal shape, width and depth were proposed. However, the impact of crimping on textile graft mechanics and therefore the flow nature inside the moving wall prosthesis has not yet been precisely defined.

We proposed to study the deformation and fluid flow in a free end textile graft commonly used in vascular surgery. First, we investigated experimentally the prosthesis deformation under different air pressure levels. Then, we carried a numerical simulation of pulsatile blood flow in the vascular prosthesis using N3S computer code (released by the Hydrodynamic Laboratory of EDF, France). Graft displacements and physical parameters of blood (relative density and viscosity) are taken into account.

2 Materials and method

The knowledge of the prosthesis wall shape is imperative when defining the exact geometry of the flow model. In order to examine the prosthesis wall reaction to the pressure generated by the blood circulation, we have submitted the prosthesis to different air pressure levels corresponding to the physiological pressures exerted on the artery wall during the cardiac cycle. Since the prosthesis should be airtight, we have tested an impregnated knitted prosthesis of 12 mm of diameter and 100 mm length (Microvel, Meadox Medicals, Inc).

According to Westerhof et al. [8] and Womersley [9] works, the arterial pressure varies between 49 and 137 mm of Hg in a 12 mm diameter artery. The experimental device (Fig. 2) is composed of a compressed air source and a pressure transducer situated upstream the prosthesis. The outlet was occluded with a polyester resin and the inlet was fixed to the air pressure device. The textile graft was subject to different air pressures levels and its profile was projected on a screen allowing thus a more precise measure of prosthesis displacement. The projected image was filmed then digitized with an image processing software. The prosthesis wall velocity equations were obtained by deriving the displacement equations and then introduced in the N3S computer code.

The numerical simulation of blood flow in the moving wall vascular prosthesis was carried out by using N3S computer code that solves Navier-Stokes equations describing fluid flow thanks to the finite element method. The used N3S algorithm was based on an Arbitrary Lagrangian Eulerian (ALE) method taking into account the deformation of the flow domain during simulation. Flow is studied in pulsatile conditions and physical parameters of blood (relative density and viscosity) are taken into account. The equations solved were Navier-Stokes equations with the following equation of continuity:

$$\begin{cases} \operatorname{div}\overrightarrow{V} = 0 \\ \rho\left[\frac{\partial\overrightarrow{V}}{\partial t} + \left[(\overrightarrow{V} - \overrightarrow{V_m}).\overline{\overline{\mathrm{grad}}}\right]\ \overrightarrow{V}\right] = -\overrightarrow{\mathrm{grad}\,p} + \operatorname{div}(\mu\ \overline{\overline{\mathrm{grad}\overrightarrow{V}}}) \end{cases} \tag{1}$$

where ρ is the density, $\overrightarrow{V}$ is the two-dimensional velocity vector, p is the pressure, μ is the viscosity and $\overrightarrow{V_m}$ is the wall prosthesis velocity.

The geometry construction and the initial meshing were made with the SIMAIL software (released by SIMULOG, France). The mesh density was more important near the wall than by the axis of the prosthesis in order to give greater importance to the calculations accuracy near the crimping area (Fig. 3). Since the flow domain is axisymmetric, the simulation involves half of the prosthesis. At

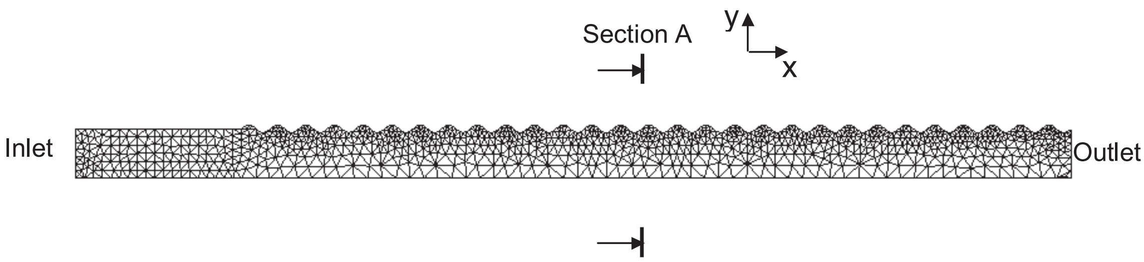

Fig. 3. Typical axisymmetric mesh for pulsatile computations.

the axis of symmetry, the normal velocity gradient is zero in addition to a zero cross flow:

$$\frac{\partial u}{\partial y} = 0, \quad v = 0. \tag{2}$$

This assumption brings a considerable gain in calculation time. A flat-wall piping added to the entrance permits the pulsatile velocity profile to be established when the prosthesis is reached. The blood was assumed to be Newtonian, homogeneous and incompressible.

For boundary conditions, we have used no slip at the wall, as well as a Poiseuille profile for the upstream flow that varies in a parabolic fashion and a free flow at the outlet:

$$u = 0 \quad \text{and} \quad v = 0 \qquad \text{at} \quad y = R \tag{3}$$

$$u = u_{max}\left(1 - \frac{y^2}{R^2}\right) \text{and}\, v = 0 \quad \text{at} \quad x = 0 \tag{4}$$

R is the prosthesis radius, u_{max} is deduced from the flow rate data of Figure 4 divided by the section of the graft.

3 Results

The prosthesis displacements were measured for different pressure levels between 0 and 90 mm Hg. The image processing provides the coordinates of all points belonging to the graft boundaries. The physiological pressure in the textile graft was given by the flow and pressure curve during the physiological cycle (Fig. 4). The temporal flow variation was adapted from Westerhof et al. [8] and Womersley [9], who described the evolution of pressure and flow in any artery section during the cardiac pulsation.

Under air pressure the free end vascular prosthesis stretched and kept the same longitudinal axis. The prosthesis remained straight and its average diameter was unchanged. The graft elongation induces only a flattening of the crimping whose width was increased and height was decreased. Table 1 shows the textile graft dimensions at different air pressure levels. Air pressure induce an elongation of the graft and the mean graft radius remains constant during the cardiac cycle. Only crimping height and width change with time. The analysis of these datum showed also that the graft length varied but not proportionally to the air pressure. The main length variation is

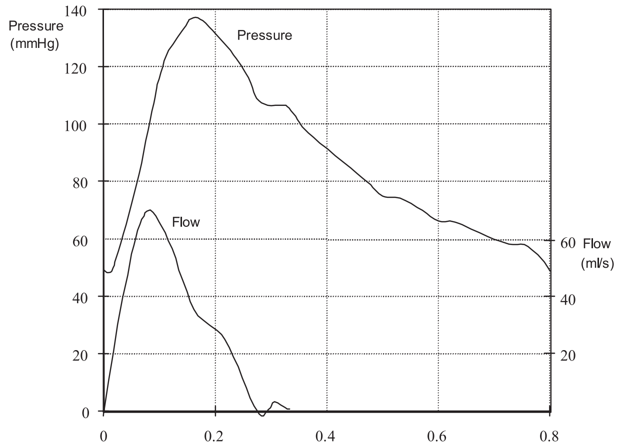

Fig. 4. Flow and pressure variation during the cardiac cycle.

obtained at low pressure levels. From 30 mm Hg the prosthesis length varies very slowly. The variation of crimping height (H) and crimping width (W) with pressure was described using the following mathematical equations:

$$H(\text{mm}) = 1.242e^{-0.0038p} \tag{5}$$

$$W(\text{mm}) = 0.2694\text{Ln}(p) + 2.3266. \tag{6}$$

The pressure profile varying with time was determined by using Fourrier series. H and W varying then with time were derived in order to obtain velocity components at the graft edges and written in a FORTRAN program giving the boundary conditions concerning the wall displacements.

The isovelocity cartography and the axial velocity profiles at different steps of the cardiac cycle are shown in Figures 5 and 6. The Poiseuille profile characterizing the graft inlet is not maintained along the cycle. A transitory perturbation phenomena associated with the flow domain inertia was observed.

When flow starts ($t = 0.05$ s to $t = 0.1$ s), the axial velocity profile, typically Poiseuille at the entry, is deformed and a plate profile is obtained in the other sections. This period often called "starting flow" characterizes the velocity distribution before the jet arrival. Crimping crests show low velocities due to the creation of local recirculation zones. Figure 7 shows these recirculation zones situated in the crimping crests at $t = 0.1$ s. When the jet reaches the prosthesis sections ($t = 0.1$ s to $t = 0.25$ s), most velocity profiles show a parabolic shape. At maximal

Table 1. Prosthesis dimensions at different air pressure levels.

Pressure (mm Hg)	Graft length (mm)	Elongation (%)	Crimping height: H (mm)	Crimping width: W (mm)
0	100	0	1.31	2.97
30	112	12	1.07	3.32
45	116	16	0.99	3.43
60	118	18	0.91	3.51
75	120	20	0.85	3.56
90	121	21	0.79	3.62

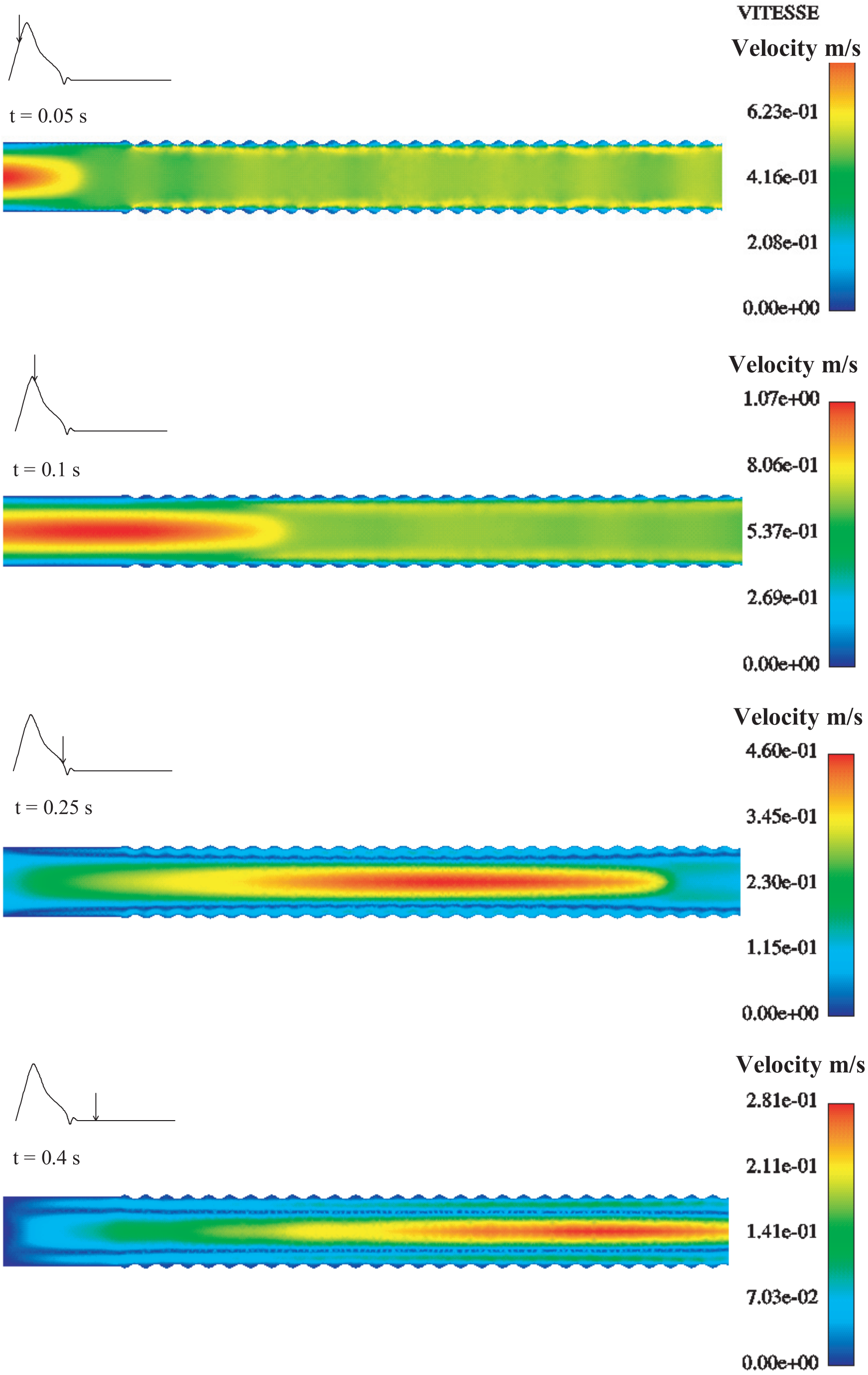

Fig. 5. Evolution of isovelocity cartography during the cardiac pulse.

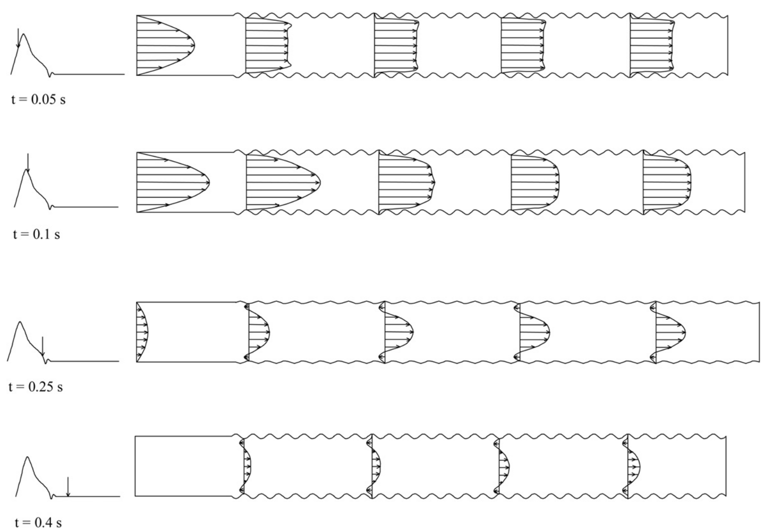

Fig. 6. Evolution of axial velocity profiles during the cardiac pulse.

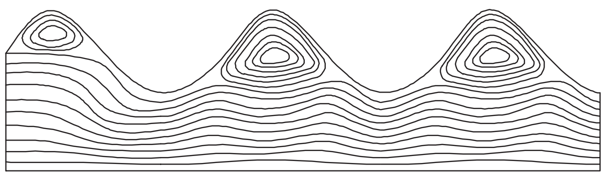

Fig. 7. Streamlines contours near crimping at $t = 0.1$ s.

pressure level ($t = 0.25$ s) crimping was flattened and no recirculation zone was observed because of the low crimping height.

From the instant $t = 0.4$ s, the velocity vectors near the wall rotate 180° which brings these vectors perpendicularly to the wall. This rotation places them in the opposite direction of flow, which explains some negative speeds observed near the wall. This phenomenon starting at the diastole phase occurs when forces of inertia become inferior to viscosity forces. The friction generated by the fluid viscosity is opposed to particles motion and sweeps them in the opposite direction of the flow.

In order to visualize how fluid was sheared, we wrote a FORTRAN program run by N3S that calculated shear stress as:

$$\tau = -\mu \left(\frac{\partial u}{\partial y} + \frac{\partial v}{\partial x} \right). \tag{7}$$

This parameter was examined in a section A situated at the prosthesis middle at maximum flow ($t = 0.1$ s) corresponding to the end of the systole phase. Figure 8 shows that shear stress at the prosthesis axis at $t = 0.1$ s is very low because of the plate shape of the velocity profile, but the wall shear stress rises sharply when approaching the prosthesis wall and decreases rapidly to a near zero value inside the crimping zone.

4 Discussion

The experimental study of prosthesis wall displacement under an internal pressure revealed an original mechanical

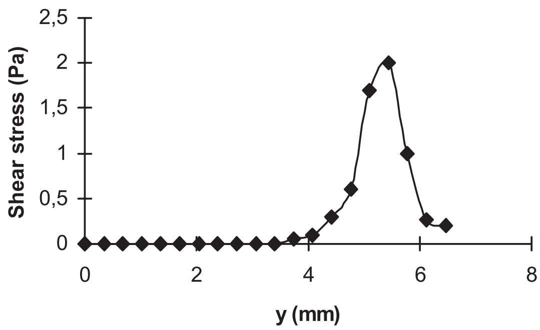

Fig. 8. Shear stress distribution near the prosthesis wall at $t = 0.1$ s.

behavior of the textile structure. The prosthesis reaction to pressure does not agree with theoretical models applied to biological arteries available in the literature [1,4,5].

These models described a transverse deformation of the artery under a pulsatile flow.

Ballyk et al. [1] have studied the circumferential compliance of a femoral artery and a Dacron textile graft but without considering the crimped shape of the graft wall. He defined the compliance as the percentage change diameter per kPa pressure. He noticed femoral artery had a higher compliance than the textile graft and concluded that this compliance mismatch can induce the graft implantation failure.

The specific mechanical behavior of textile vascular prosthesis that can be interpreted in terms of longitudinal and circumferential compliance is attributed to crimping which confers to the textile structure the aptitude to stretch because of the crimping flattening.

In a previous study [3], we measured the influence of crimping on fluid flow the kinetics by using Laser Doppler Anemometry. The prosthesis was opened longitudinally and placed inside a perfectly transparent flow circuit. We measured fluid velocity profiles near crimped and flat graft walls. The obtained results showed that flow was highly braked near the prosthesis crimped wall. In this case, the shear stresses near the graft wall were higher than those obtained in the case of a flat wall graft. Theses results agree with those obtained with numerical simulation.

The analysis of velocity and shear stress distribution shows that blood is highly sheared near the prosthesis wall. When we consider the the reference case of Poiseuille flow in a cylindrical rigid duct at the peak of flow rate ($t = 0.1$ s), shear stress value can be calculated as:

$$\tau_w = \frac{4\mu q}{\pi R^3}. \tag{8}$$

Blood viscosity at 37 °C is 3.1 mPa.s and flow value deduced from Figure 4 is 66 ml/s. Taking a value of 6 mm for the graft radius, the Poiseuille calculation yield $\tau_w = 1.2$ Pa.

Figure 8 shows that shear stress in the moving wall graft at peak of flow rate is about 2.2 Pa. The crimped shape and the displacement of the graft walls seems to contribute to the high shear stress obtained near the textile graft wall.

From all the hemodynamic factors, wall shear rate is one of the most frequently cited parameters affecting the pathology of arterial walls [6,7]. The low shear stresses zones observed inside the prosthesis crimping could have an influence on the thrombogenicity of the textile graft and could favor the adherence of platelets to the graft wall. The low particle velocities observed in the recirculation zones explain the obtained low shear stress. The hemolysis phenomenon characterized by red cells destruction under high shear rate could happen near the internal wall.

The classical experimental tools, generally used to measure fluid flow, such Laser Doppler Anemometry are not suitable to the case of the textile prosthesis. These tools need transparent surfaces. The opacity of the textile surface, which is due to the manufacturing process does not allow this type of measurements. After knitting or weaving, the prosthesis undergo a humid heat-treatment in order to reduce the porosity of the knitted or woven structure, then the graft is impregnated with an organic substance as collagen or albumin in order to reach an optimal water-tightness. In our previous study [3], the prosthesis was opened longitudinally and then velocity profiles were measured by using Laser Doppler. It was the only way to reach flow field with laser beams. Other experimental tools such as the ultrasonic technique require the introduction of probes inside the prosthesis and provoke some disturbances inside the flow field. The interest of the numerical simulation, in the case of the textile graft, is the possibility to study fluid flow in domains having very low diameter that experimental tools cannot solve and the description of blood flow inside very tiny zones having the dimension of crimping.

5 Conclusion

In the present work, we presented a numerical simulation of the fluid-structure interaction of pulsatile blood flow in a crimped textile vascular prosthesis. The ALE formulation used by the N3S code coupled textile prosthesis deformation with blood flow dynamics. The graft displacement showed a longitudinal stretching and a crimping flattening. The analysis of velocity and shear stress distribution showed that blood is highly sheared near the prosthesis wall. Crimping presented recirculation zones characterized by low shear stress.

Quantitative and qualitative studies of flow field could bring answer to some pathologies linked to hemodynamic factors in prosthesis. The low wall shear stress observed in the crimping area can explain some problems that could affect the prosthesis, such as particles deposit and wall cell development.

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