THE INFLUENCE OF THE ANGLE OF ANASTOMOSIS ON THE HEMODYNAMICS OF AN INFRAINGUINAL BYPASS

Introduction

The arterial reconstructions of the extremities represent nowadays one of the most important areas of vascular surgery. The results of the surgical and endovascular treatment are limited by two factors: the evolution of the degenerative illness and the subendothelial miointimal hyperplasia phenomenon (MIH) (1).

For the pathogenesis of these two processes and mostly in the case of MIH, aspects of flow through the bypass have a great importance (2-6). Among these, shear stress and the recirculation regions (vortexes) have a fundamental role. (7, 8).

Purpose

The main objective of this study is to enrich the knowledge about the relation between the blood flow at the anastomosis level, and the MIH occurrence. These results, obtained by the simulation of the flow through an infrainguinal bypass, are necessary for a future approach of the improvement of the geometrical form of a T-L anastomosis and by doing so, it leads to the improvement of hemodynamic aspects of the bypass.

Summary: The authors present a bidimensional computational model of the infrainguinal bypass. The study is the result of a multidisciplinary cooperation in a team formed by surgeons, Doppler echographists, mechanical engineer specialized in the fluid dynamics and computer engineer specialized in computational vascular modeling. The study compares the results from a short bypass made on the main femoral artery, and quantifies the influence of the distal angle of the bypass, on the shear stress. We consider that tight stenoses need small entrance angles of the bypass. These results, obtained by the simulation of the flow through an infrainguinal bypass, are necessary for a future approach of the improvement of the geometrical form of a T-L anastomosis and by doing so, to improve the hemodynamic aspects of the bypass.

Keywords: shear stress, infrainguinal bypass.

Materials and methods

The study is the result of a multidisciplinary cooperation in a team formed by surgeons, Doppler echographists, mechanical engineers specialized in the fluid dynamics and computer engineer specialized in computational vascular modeling. The surgeons role was to define the problems, the internist to measure the speed at the entrance, the fluid engineer modeled the phenomena, and the computer engineer programmed and computed the gathered data. The interpretation of the results was a team effort, and the surgeon validated the data.

Building the geometry of the model represented the first stage of the model construction (represented in the figure 1a). We considered a portion of the femoral artery (the inborn vessel) with a 6 mm internal width and a 0.5 mm wall thickness. We also considered that the artery is obliterated, presenting a circumpenential stenosis with an S degree of the flow section reduction. In parallel with the stenosed artery, we considered the existence of a bypass made of a central graft with the same clear width as the inborn vessel and two termino-latheral anastomoses, one proximal, within to 2 cm upstream of the stenosis, and the other distal, situated at approx. 2 cm...
downstream of the stenosis. The reason for choosing a shorter model is the uselessness of the too long intermediate rectilinear pipelines, given that the focus is on the study of the phenomena in the anastomosis zones.

The model geometry was described in a system of XOY orthogonal axes, having the origin in the axe of the inborn artery, in the same line as the symmetry of the stenosis. The stenosis was geometrically described by a central linear portion and two lateral sigmoid portions (fig. 1b). The model was given the possibility to change the form of the stenosed artery, through the modification of the report between the length of the central portion and one of the lateral portions of the stenosis. The same sigmoid profile was attributed to the two anastomosis (fig. 1c). Another geometrical parameter relevant in the present study was the angle of the anastomosis (fig. 1C).

Considering the aim of the simulation, that of investigating, among other things, the effect of the wall elasticity module in the different portions of the pipeline, the frontiers area was generated separately, so that the modification of this constituent parameter from a region to another of the model became possible. In all, there were identified 18 like regions, their frontiers being named in accordance to figure 1a.

The next stages consisted in the realization of the mathematical model and then of the numerical one. The problem that had to be solved was, by definition, one of fluid-structure interaction, with two main aspects: one of hydrodynamics and the other of elasticity, being set some conditions of continuity of the tensions and deformations at the level of the separation surface, between the moving fluid and the solid vascular wall.

The laminar motion of the blood, considered as a Newtonian incompressible fluid [6], through the area with the 2D geometry, was described through the two equations which govern the flow process. These are: the continuity equation (expressing the principle of the mass preservation, applied in the fluid dynamics) and the Navier-Stokes equation (which represent the impulse preservation equations). In the context of a plan parallel
geometry, the problem of elasticity becomes one of elastic wall deformation, produced by planar solicitation (a force exerted by the fluid movement) and it is reduced to solving an equation of mechanic equilibrium. The solution to both problems, reduced to solving systems formed by equations with partial derivatives, can’t be found analytically, so it imposed the use of a numerical method.

The used numerical method was The Finite Element Method (FEM). The FEM principle consists in the partition of the analysed field in finite elements, usually triangular ones, and the approximation of the solution of a partial derivatives equation (the Navier-Stokes equation in the problem of hydrodynamics, respectively the equation of equilibrium of the forces in the problem of elasticity) at the elemental level, through inferior order polynomial interpolation (9-12).

In the figure 2 is illustrated the net of discretization of the field area analysed in the present article. This net consists of 5525 triangular finite elements, each with six knots of interpolation.

The coupling of the two problems was solved with the help of an algorithm implemented in software Fastflo 3., based on the ALE method (Arbitrary Lagrangean Eulerian) (13).

The direct result of the integration of the mathematical model equations is represented by the velocity fields, the pressures and the shear stress. The whole cardiac circle was temporally discretized with the 0.01 seconds step, so that for each of the 80 steps was obtained a spatial casting of the nameable variables.

To ensure the unicity of the solutions of the partial derivatives equations systems, there were enforced boundary conditions. One among these refers to the variation form, during a cardiac circle, of the axial average fluid velocity at the entrance of the bypass. This velocity was measured through Doppler echography and then mathematically described through the polynomial interpolation of the obtained values, just as it is seen in the figure 3.

In order to find out the way in which the á angle of the anastomosis influences the occurrence of the miointimal hyperplasia, the program was run at a 95% stenosis and at different angles, respectively 20, 45 and 60 degrees, following the portions vicinal to the walls in which appear zones of fluid vortexes and the duration in which these persists.

In all simulations, the physical parameters blood values taken into consideration were: the density \( \rho = 1100 \) and the dynamic viscosity \( \mu = 0.004 \text{ Pa·s} \). For all the walls that limit the area, it was necessary to choose an elasticity bigger than in reality, which means that the walls are much rigid, because in the absence of the circumperitale tensions (two-dimensional models) the deformations would be very large.

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**Fig. 2.**

**Fig. 3.** The condition on the input frontier( f12) represented by the axial moderate speed up-stream of the proximal anastomosis.
Results

We obtained and compared 240 images.

Figure 5 presents the $\tau$ spatial distribution $|\nabla \tau|$ (the shear stress value in a moment of time) and the distribution of $\frac{\partial \tau}{\partial t}$ (the shear stress gradient, which contains the longitudinal and transversal variations) from the moment in which (the temporal variation of the shear stress) is maximum. It also represents only the distal anastomosis.

Table 1. The total action time of the vortex zones on various points of the distal anastomosis

<table>
<thead>
<tr>
<th></th>
<th>$S\ 95%_{\alpha} 20$</th>
<th>$S\ 95%_{\alpha} 45$</th>
<th>$S\ 95%_{\alpha} 60$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Toe</td>
<td>4 ms</td>
<td>18 ms</td>
<td>16 ms</td>
</tr>
<tr>
<td>Distal from toe</td>
<td>16 ms</td>
<td>1 ms</td>
<td>7 ms</td>
</tr>
<tr>
<td>Heel</td>
<td>10 ms</td>
<td>1 ms</td>
<td>1 ms</td>
</tr>
<tr>
<td>Bed</td>
<td>26 ms</td>
<td>30 ms</td>
<td>21 ms</td>
</tr>
<tr>
<td>Total distal</td>
<td>26 ms</td>
<td>30 ms</td>
<td>98 ms</td>
</tr>
<tr>
<td>Proximal anastomosis</td>
<td>16 ms</td>
<td>30 ms</td>
<td>30 ms</td>
</tr>
</tbody>
</table>

Fig. 5. The $\tau$ and $|\nabla \tau|$, spacial distribution from the moment in which $\frac{\partial \tau}{\partial t}$ is maximum.
Discussions

It is unanimously accepted that the spatial and temporal gradients of the shear stress which characterise the vortexes, as well as the low shear stress level, lead to miointimal hyperplasia.

The temporal gradient of the shear stress was defined as increase or decrease of the shear stress value in the same point of the vessel wall, throughout very short periods of times. The spatial longitudinal gradient of shear stress was defined as variation of the shear stress in the same moment between neighbouring points of the wall.

According to the table, at least in the case of tight stenoses, it is preferable to have small connection angles, which means that the distal anastomosis will be exposed to much smaller variations of the shear stress, given the fact that vortexes induce large modifications of the temporal and spatial shear stress gradients. Vortexes which affect in the same moment different places were responsible, in the case of the distal anastomosis, for the longer period of time than the duration of the cardiac cycle obtained at the 60 degrees angle. The table shows that the increase of the bypass entering angle in the case of the distal anastomosis lead to the increase of the time in which the toe is submissive to the vortexes, and a smaller angle exposes mainly the heel.

Figure 5 reveals the big variation of the shear stress at the level of the distal anastomosis (the -0.02 and -0.004 portion, representing the entrance of the graft in the inborn artery), both in the case of the toe and distal from it, as well as at the level of the opposite wall to the toe, in the moment when the temporal variation of the shear stress is maximum. Also, it is noticed that the heel and the bed of the anastomosis is submissive to low shear at the same moment of time. The vortexes that appeared in the graft were not taken into consideration because the sudden change of direction between the areas f10 to f9 and f13 to f14 generates vortexes which represent artefacts (these angles would correspond to important bends of the graft).

The limits of the study are represented by the bydimensional modeling, by the rigidity of the walls, by the fact that the same angle was used both in proximal and distal anastomosis, by the sudden change of direction between the areas f10 to f9, f13 to f14, f9 to f8, f13 to f14 and f14 to f15 (which is not in accordance with reality), and last, but not least, by the authors initial experience in this complex problem.

The used approximates, in the same time possible sources of errors, enforce the switch to a tridimensional model. This switch involves the use of powerful and expensive computational stations, but it will open the path of virtual surgery, when the surgical treatment will be simulated before surgery for each individual, the current study being the first step in the path of this desideratum.

Conclusions

1. Although the study has numerous limits, which enforce caution in the interpretation of the results, it offers a new view for a bypass, respectively a hemodynamical one.

2. At least in the case of tight stenoses, small entrance angles of the bypass are preferable, from the viewpoint of the studied factors.

3. The increase of the bypass entrance angles into the distal anastomosis prolongs the time in which the toe is affected by vortexes, and a small angle exposes mainly the heel.

4. The heel is submitted to low shear stress zones when the temporal variation of the shear stress is maximum.

5. A multidisciplinary approach is enforced, in which the surgeon has both the role to raise the questions and validate the results, through comparing the simulations results with experimental animal studies and professional experience.

6. The switch to a tridimensional model shall offer a detailed image, with a less degree of error regarding the hemodynamics of the bypass, opening the path to the virtual surgery.
References:
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