Patient-specific simulation of the blood flow in a proximal femoral artery-graft junction

Y F Ivanova\textsuperscript{1,2}, L G Tikhomolova\textsuperscript{2}, A D Yukhnev\textsuperscript{1,2}, Y A Gataulin\textsuperscript{1,2}, E M Smirnov\textsuperscript{1,2}, R V Kalmikova\textsuperscript{2}, A N Morozov\textsuperscript{2}, A A Suprunovich\textsuperscript{2}, A A Vrabiy\textsuperscript{2} and G G Khubulava\textsuperscript{2}

\textsuperscript{1}Peter the Great St. Petersburg Polytechnic University, 29 Polytechnicheskaya Str., 195251, St. Petersburg, Russia  
\textsuperscript{2}Pavlov First St. Petersburg State Medical University, 6-8L'va Tolstogo Str., 197022, St. Petersburg, Russia

E-mail: radfn94@mail.ru

Abstract. The paper presents a computational procedure developed for patient-specified blood flow simulation in the proximal anastomosis of femoral-popliteal bypass. The calculations are based on a three-dimensional geometric model constructed from a CT angiography scan of the vascular bed in the anastomotic region and boundary conditions obtained by ultrasound Doppler measurements of the velocity in control sections of the bed. A numerical analysis of 3D pulsating flow structure in a particular model of the proximal anastomosis has been carried out. The model includes sections of the common femoral artery (CFA), the graft, the deep femoral artery and the superficial femoral artery. The Reynolds number based on the CFA inner diameter and the inlet bulk velocity is 880. The sizes and location of stagnant zones, areas with low values of the time-averaged wall shear stress modulus and with high values of the oscillatory shear index are determined. It is shown that the results of the prediction of areas with low values of the time-averaged shear stress modulus are in qualitative agreement with the data of local measurements of neointima thickness formed along the graft wall one year after the operation.

1. Introduction

In modern vascular surgery, the bypass grafting is one of the methods of blood flow recovery. In particular, in femoral-popliteal bypass, graft connects the common femoral artery (CFA) by a proximal anastomosis above and the popliteal artery (PA) by a distal anastomosis below the obstructed part of the artery.

After bypass surgery, it is important to monitor blood flow parameters recovery, as well as to assess the risks of postoperative complications, namely, the rapid growth of neointima in the area of the graft-artery junction (anastomosis). In postoperative diagnostics, along with clinical studies, numerical simulation can be used. It makes possible to describe a patient-specific three-dimensional blood flow structure in the anastomosis [1-3].
Patient-specific simulation allows evaluating hemodynamic parameters affecting neointimal growth. To date, the following parameters significant for this process have been identified: maximum wall shear stress, time-averaged wall shear stress (TAWSS), oscillatory shear index (OSI) [4-6]. It is possible to evaluate the rate of neointima postoperative overgrowth, using knowledge about the relationship of hemodynamic parameters with neointima growth in the anastomosis.

The objectives of this study are the development and pilot application of a patient-specific blood flow simulation in the area of the proximal synthetic graft/femoral artery anastomosis.

2. Methods

2.1. Geometric model

The patient-specific 3D model of the anastomosis considered in this work was constructed on the basis of angiography data obtained by multispiral computed tomography (GE Optima 660 with AWServer20 V.5 software package). Tomography was performed 12 months after the femoropopliteal bypass above the knee with the Ecoflone L8-80 linear vascular prosthesis. Four computer programs were used to construct the model. In the 3D Slicer code, medical images are viewed, divided into segments, and output to a source file with geometric data. In the VMTK (Vascular Modeling Toolkit) program, the ends of the 3D model are cut perpendicular to the centerline of the vessel and the facet body is formed. In Rhino 6, the faceted body is converted to a polysurface. Finally, Siemens NX 10 code is used to construct cross-sections of the vessels, at which the inlet and outlet boundary conditions for blood flow simulation are subsequently set.

The 3D model constructed is shown in Fig. 1. Internal diameter of the common femoral artery (CFA) at the entrance to the model is 4.5 mm, deep femoral artery (DFA) – 3.4 mm, superficial femoral artery (SFA) – 1.6 mm, graft – 8.1 mm.

![Figure 1. Patient-specific model geometry of the femoral artery proximal anastomosis region and examples of the computational grid cross-sections](image-url)
2.2. Mathematical model and computational aspects

A pulsating flow of an incompressible viscous fluid with density of 1000 kg/m$^3$ and viscosity of 0.004 Pa·s, which is typical for blood flow in large vessels, was considered. The elasticity of the vessel walls was not taken into account.

As in the previous authors work [7], devoted to modeling the flow in a simplified distal anastomosis model, the present patient-specific simulation was carried out using the ANSYS CFX fluid dynamics package. The package provides the possibility of numerical integration of non-stationary three-dimensional Navier-Stokes equations using the finite volume method. No turbulence models were used in the present computations.

According to the finite volume method, the computational domain is divided into sufficiently small elements (control volumes), for each of which the conservation equations in integral form are applied. Then the approximation of the integrals by quadrature formulas, interpolation and finite difference schemes are used. In the ANSYS CFX, control volumes are automatically built around the nodes of the computational grid.

For the numerical simulation of blood flow in the considered anastomosis model, an unstructured computational grid was constructed in the ANSYS Meshing program (Fig. 1). The number of mesh elements is about 3 million. The mesh contains a layer of prismatic elements that are condensed towards the wall with a coefficient of 1.4. When performing the simulation, the physical-time advancement was carried out with a second-order scheme, the time step was 0.01 s. For convective fluxes calculations, the upwind second-order scheme option was chosen.

To set boundary conditions, patient-specific data of clinical ultrasound measurements of the pulsating flow rate (Mindrey 7) were used (Fig. 2). Time-averaged flow rate at the inlet is 200 ml/min. 50% of the time-average inlet flow rate (100 ml/min) is set at the graft outlet. 10% of the time-average inlet flow rate (20 ml/min) is set at the SFA outlet. The cycle time, $T$, is 0.82 s. At the instance of maximum flow, the bulk velocity, $V_b$, at the model inlet (CFA) is 1.6 m/s. Velocity profiles were assumed to be flat. At the outlet of the DFA, a reduced pressure of zero was prescribed. The no-slip condition was set on the walls.

![Image](image.jpg)

**Figure 2.** Patient-specific bulk velocity waves used to set boundary conditions at the inlet (CFA) and outlets (graft and SFA)

The most important parameter of the simulated pulsating flow is the maximum inlet Reynolds number, which is evaluated with the inner diameter of CFA at the inlet and the bulk velocity at the instance of maximum flow. For the considered model of anastomosis, the maximum Reynolds number is 1800. Pulsating blood flow is also characterized by the Womersley number. The value of this parameter, built on the basis of the CFA inlet radius and the above given cycle time, is 3.1.
Methodological calculations have shown that in case of starting from zero velocity fields, the periodic solution is practically established in two cycles. The below presented distributions of the time-averaged wall shear stress (TAWSS) and the (dimensionless) oscillatory shear index, defined as

$$OSI = \frac{1}{2} \left( 1 - \int_0^T \bar{\tau}_w \, dt\right) \left/ \int_0^T |\tau_w| \, dt \right)$$

were obtained by averaging the actual data over the third cycle.

3. Results
When analyzing the calculated flow field in the artery/graft junction region, a stagnant zone is found, the length of which at the instance of maximum flow is about 20 mm (see the picture of streamlines in Fig. 3). It is known that the presence of stagnant zones is closely related to the presence of areas of small values of TAWSS and increased values of OSI. These features of the flow structure in the anastomosis area promote development of postoperative subintimal hyperplasia.

Zones of low shear stresses (less than 1 Pa) are observed in the artery/graft junction region, as well as along the internal wall of the graft (Fig. 4). As for OSI, its “dangerous” high values (up to 0.5) are observed both directly in the suture area and in separate places along the graft (Fig. 5). To obtain integral estimates, the TAWSS and OSI values were averaged over a section of the artery and graft walls (over a length of 20 mm) in the area of the artery/graft junction. This area is the most dangerous in terms of the subintimal hyperplasia risk. The estimate obtained for TAWSS is 3.1 Pa, that is outside the range associated with an increased risk of subintimal hyperplasia (TAWSS<1 Pa). The estimate obtained for OSI, which is about 0.2, is 40% of the maximum of critical value (OSI=0.5).
The calculated local values of TAWSS were compared with the data of ultrasonic measurements of the neointima thickness (Fig. 6). Measurements were carried out from the "heel" of the anastomosis (the downstream edge of the suture) along the graft wall at a length of up to 25 mm along the side closest to the SFA. The measured neointima thickness distribution generally correlates with the calculated data for TAWSS and OSI. At lower shear stresses, a larger thickness of the neointima is observed and, conversely, at higher shear stresses, the thickness of the neointima is smaller. For OSI: at higher oscillatory shear index, a larger thickness of the neointima is observed and, conversely, at lower OSI, the thickness of the neointima is smaller.

![Graph showing changes in neointima thickness, OSI, and TAWSS](image)

**Figure 6.** Changes in the neointima thickness, NI; OSI and the mean modulus of TAWSS along the internal wall of the graft (one year after surgery)

**4. Conclusions**

A technique has been developed for patient-specified blood flow simulation in the proximal anastomosis of femoral-popliteal bypass. The calculations are based on a three-dimensional geometric model constructed from a CT angiography scan of the vascular bed in the anastomotic region and boundary conditions obtained by ultrasound Doppler measurements of the velocity in the control sections of the bed. A numerical analysis of the flow structure in a particular model of the proximal anastomosis, has been carried out. The sizes and location of stagnant zones, areas with low values of the time-averaged wall shear stress modulus and with high values of the oscillatory shear index have been determined. It has been shown that the results of the prediction of areas with low values of the time-averaged shear stress modulus are in qualitative agreement with the data of local measurements of neointima thickness formed along the graft wall one year after the operation.

**Acknowledgements**

The study was carried out in the framework of Project No. 20-65-47018 supported by the Russian Science Foundation.

**References**

[1] Marsden A L 2014 Optimization in Cardiovascular Modeling *Annual Review of Fluid Mechanics* **46**(1) 519–46

[2] Diaz-Zuecarini V, Agu O, Tomaso G Di and Pichardo-Almarza C 2014 Towards personalised management of atherosclerosis via computational models in vascular clinics: technology based on patient-specific simulation approach*Health Technol Lett* **1** 13–8

[3] Randles A, Frakes D H and Leopold J A 2017 Computational Fluid Dynamics and Additive Manufacturing to Diagnose and Treat Cardiovascular Disease *Trends in Biotechnology* **35**(11) 1049–61
[4] Haruguchi H and Teraoka S J 2003 Intimal hyperplasia and hemodynamic factors in arterial bypass and arteriovenous grafts: a review Artif Organs 6(4) 227-35

[5] McGah P M, Leotta D F, Beach KW, Riley JJ and Aliseda A 2011 A longitudinal study of remodeling in a revised peripheral artery bypass graft using 3D ultrasound imaging and computational hemodynamics J Biomech Eng 133 41008 1-10

[6] Donadoni F, Pichardo-Almarza C, Bartlett M, Dardik A, Homer-Vannisinkam S and Diaz-Zuccarini V 2017 Patient-Specific, Multi-Scale Modeling of Neointimal Hyperplasia in Vein Grafts Front. Physiol. 8 20

[7] Ivanova Y F, Yukhnev A D, Gataulin Y A, Smirnov E M, Vrabiy A A and Vavilov V N 2020 Numerical and experimental study of the 3D flow in a graft-artery junction model J. Phys.: Conf. Ser. 1675 0120036