Influence of the femoral artery-graft junction patient-specific geometry on blood flow structure and wall shear stress

Y F Ivanova\textsuperscript{1,2}, L G Tikhomolova\textsuperscript{2}, A D Yukhnev\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}, G G Khubulava\textsuperscript{2}, V N Vavilov\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 comparative numerical study of pulsatory blood flow in five patient-specific models of femoral-popliteal artery anastomosis. Three-dimensional geometric models of a proximal junction of the common femoral artery/graft were constructed on the bases of CT angiography. The influence of junction geometry on the blood flow and wall shear stress is analyzed. The ratio of the measured CFA and graft diameters and the junction angle are considered as the major geometrical parameters. Numerically calculated velocity fields are analyzed, and stagnant zones in the anastomoses flow are identified. Time-averaged distributions of wall shear stress and oscillatory shear index obtained for five patient-specific model are compared.

1. Introduction

Vascular grafts are used for blood flow recovery in the obstructed vessels. Synthetic grafts are widely used for femoral-popliteal bypass [1]. It is important to monitor blood flow parameters recovery after femoral-popliteal bypass surgery due to hemodynamic disturbances and the high risk on subintimal hyperplasia [2]. Individual femoral arteries geometry and graft-artery junction type substantially determine the blood flow structure in both anastomosis – proximal (upstream) and distal (downstream).

Currently, numerical simulation is used to estimate hemodynamic parameters, which influence upon the neointimal growth rate [3]. These are the time-averaged wall shear stress (TAWSS) and the oscillatory shear index (OSI) [4-6]. Post-surgery observations of neointimal growth in the femoral anastomoses zones will allow us to identify correlations with geometric parameters, flow structure and wall shear stress distribution. It should be noted that all ultrasound measurements for setting the boundary conditions have a higher accuracy for the proximal anastomosis compared to the distal one.

The aim of the present work is numerical study of the patient-specific geometry effects on the blood flow structure and the wall shear stress distribution in proximal femoral-popliteal bypass anastomosis.
2. Methods

2.1. Geometric model

Five individual geometries of real patients were considered. 3D geometric models, constructed on the base of angiography data, are shown in figure 1. Each of the models represents a patient-specific junction of the common femoral artery (CFA) with a synthetic vascular graft in the end-to-side type. There is one inlet, CFA, and two outlets, DFA and graft, in each of the models.

The major geometry characteristics are the ratio of the CFA and graft diameters and the junction angle. For patients K., E., V., I. and R. the CFA diameters is equal to 6.9, 4.8, 6.4, 4.5, 7.2 mm, correspondingly, the CFA-graft diameters ratios are 1.4, 1.8, 1.9, 1.2, 1.2 and the junction angles - 37°, 42°, 44°, 56°, 85°. In Figure 1, the models are positioned in order of increasing the artery-graft junction angle.

![Figure 1. Patient-specific models of the femoral-popliteal proximal anastomosis region](image)

2.2. Mathematical model and computational aspects

Numerical simulation of blood flow was carried out using the ANSYS CFX fluid dynamics finite-volume package. Numerical integration of non-stationary three-dimensional Navier-Stokes equations was carried out. Typical for large vessels The newtonian fluid model with density of 1000 kg/m³ and viscosity of 0.004 Pa·s was adopted that is typical for modelling of hemodynamics in case of large vessels. The walls elasticity was not taken into account. For each of the models, an unstructured mesh consisting of about 3 million elements, including a prismatic layer near the wall, was constructed. The time step was chosen as 0.01 s.

Despite the temporal variation of the inlet bulk velocity (the so called “velocity curve”) was measured for each patient, in the present numerical study an identical, typical, velocity curve shape was set for all models. This approach allows us to extract the influence of geometry on blood flow structure without a shadowing effect of the velocity curve shape. In fact, the inlet velocity curve shape got for the patient K. was taken (shown in figure 2). As well, for all models, the same time-average flow rate was set: at the inlet - 300 ml/min, at the graft outlet - 150 ml/min (50% of the inlet flow). The indicated values were
obtained by averaging our blood flow ultrasound measurements for 5 operated patients. Inlet velocity profiles were assumed flat. At the outlet of the DFA, a constant (zero) value of the reduced pressure was prescribed. The no-slip condition was set on the walls. Maximum values of the Reynolds number were evaluated based on the CFA inner diameter at the inlet and the bulk velocity at the instance of maximum flow. The patient-averaged Reynolds number was about 1500.

3. Results

Figure 3 shows the streamline patterns plotted for five models at the instance of maximum flow. These patterns show the presence of stagnant zones downstream of the artery-graft junction region. The stagnant zones is close in size for models K., E., V., I., whereas model R. is characterize by a notably smaller stagnant zone. It can be attributed to the most unimpeded flow direction in the artery-graft junction for this case. As expected, higher velocity in CFA are observed in models with smaller inlet diameters.
It is known that the presence of stagnant zones is closely related to occurring areas of low values of time-average wall shear stress (TAWSS) and high values of the oscillatory shear index (OSI). These parameters are define as:

\[ TAWSS = \frac{1}{T} \int_{0}^{T} |\vec{\tau}_w| dt \]  

\[ OSI = \frac{1}{2} \left( 1 - \frac{\int_{0}^{T} \vec{\tau}_w dt \sqrt{\int_{0}^{T} |\vec{\tau}_w| dt} }{\int_{0}^{T} |\vec{\tau}_w| dt} \right) \]  

where \( \vec{\tau}_w \) is the wall shear stress vector, \( T \) – cycle time, \( t \) – time. Note also that low TAWSS (less than 1 Pa) and high OSI (about 0.5) are associated with higher risks of subintimal hyperplasia after surgery.

Figure 4 shows time-averaged wall shear stress distributions computed for all the geometrical models. Only values below 5 Pa are shown to illustrate occurrence of “dangerous” areas, where TAWSS is less than 1 Pa. One can see that in all the cases the areas with low TAWSS-values are positioned downstream of the artery-graft junction area. The same can be concluded for the oscillatory shear index fields. As illustrated in figure 5, higher values of OSI are observed downstream of the graft-artery junction.

Together with the geometric characteristics of the models, table 1 contains data on the TAWSS and OSI values averaged over a walls-surface section covering a length of 20 mm in the area of the artery/graft junction (figure 6). It can be seen that the highest OSI value is observed in case of the patient K. model (0.21). The Patient V. model is characterized by the lowest TAWSS value (1.1 Pa). Notably, that both the models have the smallest artery-graft diameters ratio.
Figure 5. Oscillatory shear index distribution

Figure 6. Surface section for integration of TAWSS and OSI (the patient R. case)

Table 1. Geometrical parameters of five models and calculated wall shear stress parameters

| Patient | K. | E. | V. | I. | R. |
|---------|----|----|----|----|----|
| $D_{CFA}$, mm | 6.9 | 4.8 | 6.4 | 4.5 | 7.2 |
| $D_{graft}/D_{CFA}$ | 1.2 | 1.9 | 1.2 | 1.8 | 1.4 |
| $\alpha$, $^\circ$ | 37 | 42 | 44 | 56 | 85 |
| TAWSS, Pa | 1.6 | 3.5 | 1.1 | 4.7 | 1.4 |
| OSI | 0.21 | 0.15 | 0.07 | 0.16 | 0.17 |
4. Conclusions
Numerical study of the patient-specific geometry effects on blood flow structure and wall shear stress distribution in proximal femoral anastomosis has been performed. Five three-dimensional models were considered, which had different values of the ratio of the artery-graft diameters and the artery-graft junction angle.

Stagnant zones were found downstream of the anastomosis area, which occurrence is associated with the curvature of the artery-graft junction. Typical characteristics of flows predicted for all models are lower values of the time-averaged wall shear stress and higher values of the oscillatory shear index in the graft downstream of the anastomosis area. A further analysis is needed to establish definite correlations between geometric and hemodynamic parameters.

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

References
[1] Donadoni F, Pichardo-Almarza C, Bartlett M, Dardik A, Homer-Vanniasinkam S and Díaz-Zuccarini V 2017 Patient-Specific, Multi-Scale Modeling of Neointimal Hyperplasia in Vein Grafts Front. Physiol. 8 20
[2] Kuryanov P C, Razuvaev A S, Vavilov V N 2008 Intimal hyperplasia within a vascular anastomosis Angiology and vascular surgery 14 146–51
[3] Díaz-Zuccarini 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
[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] Suo J, McDaniel M C, Eshtehardi P, Dhawan S S, Taylor R W, Samady H and Giddens D P 2011 Computing in Cardiology 38 217-19
[6] McGah P M, Leotta D F, Beach K W, Riley J J 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