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Simulations of the blood flow in the arterio-venous fistula for haemodialysis

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The Ciminio-Brescia arterio-venous fistula is a preferred vascular access for haemodialysis, but it is often associated with the development of vascular complications, due to changes in hemodynamic conditions. Computational fluid dynamics methods were involved to carry out seven simulations of the blood flow through the fistula for the patient specific (geometrical) case and various boundary conditions. The geometrical data, obtained from the angio-computed tomography, were used to create a 3-dimensional CAD model of the fistula. The blood flow patterns, blood velocity and the wall shear stress, thought to play a key role in the development of typical complications (stenoses, thromboses, aneurysms, etc.), have been analyzed in this study. The blood flow is reversed locally downstream the anastomosis (where the artery is connected to the vein) and downstream the stenosis in the cannulated vein. Blood velocity reaches abnormal value in the anastomosis during the systolic phase of the cardiac cycle (2.66 m/s). The wall shear stress changes in this place during a single cycle of the heart operation from 27.9 to 71.3 Pa (average 41.5 Pa). The results are compared with data found in the literature.

Key words: *a-v fistula, blood flow, computational fluid dynamics, vascular access*

1. Introduction

Patients who suffer from the end-stage renal disease need kidney transplants or dialysis. In haemodialysis, blood is withdrawn from the patient and circulates through a special filter that removes products of metabolism. The Cimino-Brescia arterio-venous (a-v) fistula is a surgically created connection between the artery and the vein and it is considered the best vascular access for haemodialysis. As the connection is made, high pressure blood from the artery flows directly to the vein and extends its diameter, which ensures an adequate flow rate (5.0–13.3 cm^3/s) [1], [2]. This means five to ten times greater flow rate than the normal one. Moreover, the limp vein wall changes and becomes thicker and more elastic, which enables multiple cannulation of the vein. This process lasts 6–8 weeks and is called maturation (arterialisation) of the fistula.

The fistula is usually created (if possible) between the cephalic vein and the radial artery at the wrist or the forearm and there are generally three types of such connections, namely: side-to-side, end-to-side, end-to-end. A significant number of complications (stenoses, thromboses, aneurysms) of fistulae is related to the geometry of the anastomosis (the place where the artery connects with the vein) and the local hemodynamics [3]. Local flow conditions, in particular the wall shear stress (WSS), are thought to affect sensitive endothelial cells on the inner vessel wall, which leads to intimal hyperplasia and further complications [1], [3]. This paper shows results (blood velocity patterns, WSS) obtained in blood flow simulations performed with the ANSYS CFX code.

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2. Methods and materials

2.1. Geometrical model of the fistula

The aim of this study was to carry out simulations of the blood flow not only near the anastomosis but also in more distal veins receiving blood from the fistula for the patient specific case. Therefore, a set of DICOM images of the matured radiocephalic end-to-end a-v fistula (obtained from the angio-computed tomography – angio-CT) was used as the basis of geometrical data. These images were opened in the RSR2 – medical images viewer featuring a possibility of reading coordinates of the points and special measuring functions. The course of the vessels in space could be found and the main dimensions of vessels could be measured. A full 3D CAD model of the fistula was created in Solid Works 2010 (Fig. 1). In fact, this model represents a volume of the blood flowing through the fistula. Thus, external walls of the model are internal walls of the blood vessels.

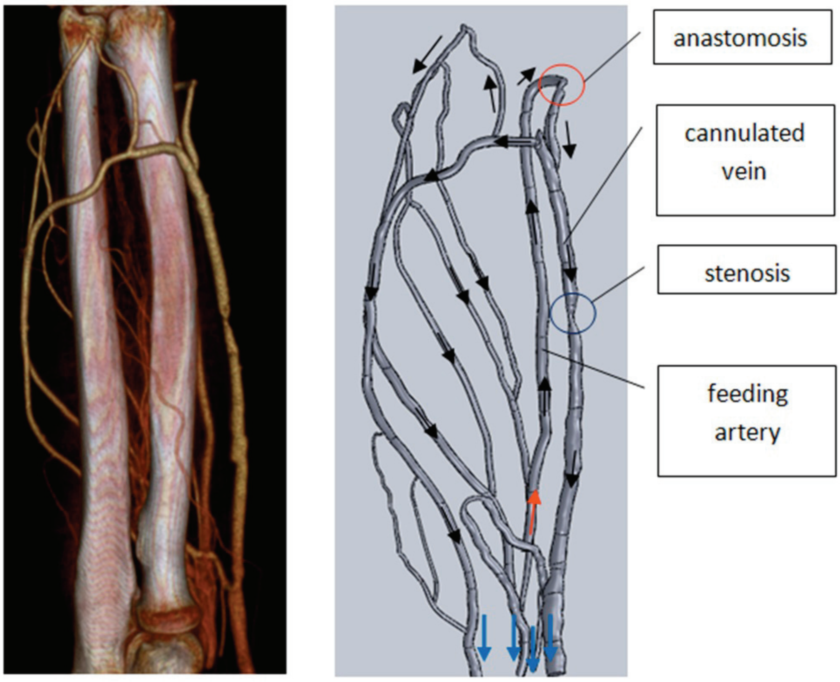

Fig. 1. Model of the fistula observed in the RSR2 (left) and a 3D CAD model created in Solid Works (right); red arrow – inlet, blue arrows – outlets, black arrows – direction of the blood flow through the fistula [5]

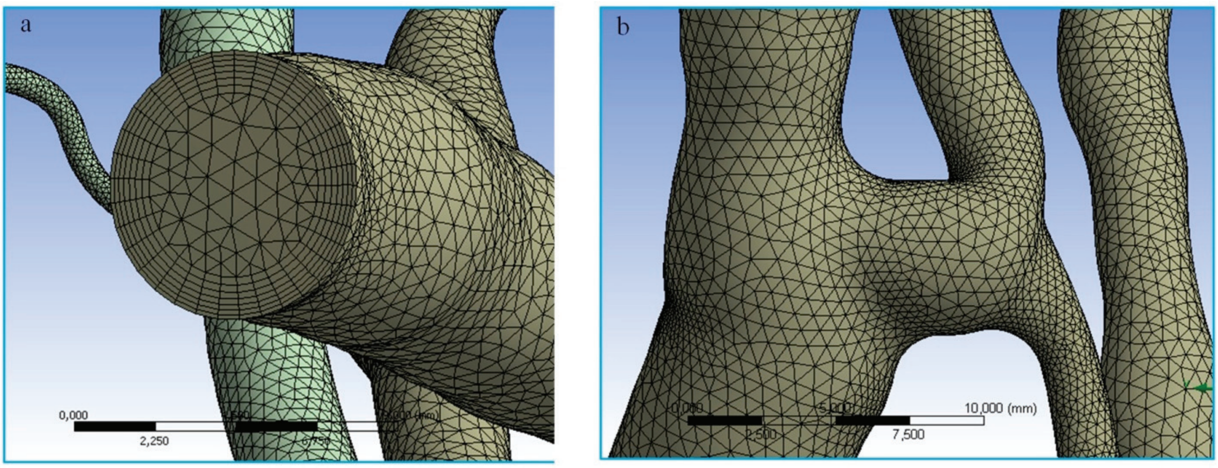

Fig. 2. Mesh of the model: (a) cannulated vein with prismatic elements in the vicinity of the wall, (b) bifurcation of blood vessels [5]

In this study, walls of all vessels were assumed rigid and non-deformable, which is a typical assumption used in other papers concerning simulations of blood flows [3], [4].

A geometrical model of the fistula was meshed and tetrahedral elements were built. Additionally, prismatic elements were introduced near walls of all vessels to increase the accuracy of calculations in the vicinity of the wall. Samples of the mesh are shown in Fig. 2.

2.2. Boundary conditions

As boundary conditions, a typical blood flow velocity waveform [2] at the inlet cross-section (feeding the brachial artery) and a blood static pressure waveform [4] at the outlet cross-sections (receiving veins) were used (see Fig. 3). The pick value of pressure is offset by approximately 0.1 s from the maximum value of velocity. Seven cases for one cardiac cycle (t = 0.84 s) were defined and 7 stationary simulations (A-G) of the blood flow through the fistula were performed.

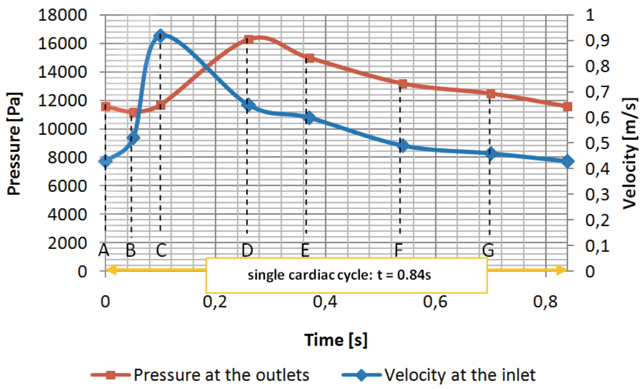

Fig. 3. Changes in time-dependent velocity at the inlet cross-section and time-dependent pressure at the outlet cross-sections during a single cardiac cycle

2.3. Fluid model

From the viewpoint of flow, density, viscosity and heat transfer (neglected in this study) are the fundamental blood parameters. However, blood is very difficult to describe by these parameters because all of them depend on personal features (e.g., hematocrit, age, sex, diet, life-style, etc.).

Blood density values fall within the range 1030–1070 kg/m^3 [6]. In this study, the constant value of blood density equal to 1045 kg/m^3 was assumed. Blood is considered a non-Newtonian fluid because its viscosity depends not only on the kind of liquid but also on the blood velocity gradient (flow parameter). For this reason, numerous rheological models of blood can be found in the literature. For the needs of this simulation, a modification [4] of the Power Law was employed where the dynamic viscosity (η) is defined as

$$\eta = 0.55471 \quad \text{for} \quad \frac{\partial V}{\partial y} < 10^{-9},$$

$$\eta = \eta_0 \left(\frac{\partial V}{\partial y} \right)^{n-1} \quad \text{for} \quad 10^{-9} \le \frac{\partial V}{\partial y} < 327, \tag{1}$$

$$\eta = 0.00345 \quad \text{for} \quad \frac{\partial V}{\partial y} \ge 327,$$

where η_0 = 0.0035 Pa s, n = 0.6 and $\frac{\partial V}{\partial y}$ is the shear stress rate. A representation of different blood viscosity models is shown in Fig. 4.

2.4. Governing equations

To obtain a solution of the problem described, the ANSYS CFX code (version 13.0) was used. According to the ANSYS CFX Theory Guide, the problem is solved by solving equations of mass, momentum, and energy conservation (Navier–Stokes equations). The continuity equation is expressed as

$$\frac{\partial \rho}{\partial t} + \nabla(\rho U) = 0 . \tag{2}$$

The momentum equation takes the following form

$$\frac{\partial(\rho U)}{\partial t} + \nabla(\rho U \times U) = -\nabla p + \nabla \tau + S_m . \tag{3}$$

The total energy equation is noted as

$$\frac{\partial(\rho h_{\text{tot}})}{\partial t} - \frac{\partial p}{\partial t} + \nabla(\rho U h_{\text{tot}})$$

$$= \nabla(\lambda \nabla T) + \nabla(U\tau) + US_m + S_E , \tag{4}$$

where: ρ – density, U – velocity, p – pressure, τ – stress tensor, S_m – sum of body forces, $h_{\text{tot}} = h + 0.5\, U^2$– the total enthalpy, S_E – energy source, λ – thermal conductivity.

In the present investigation, the thermal energy equation is not taken into consideration as the blood

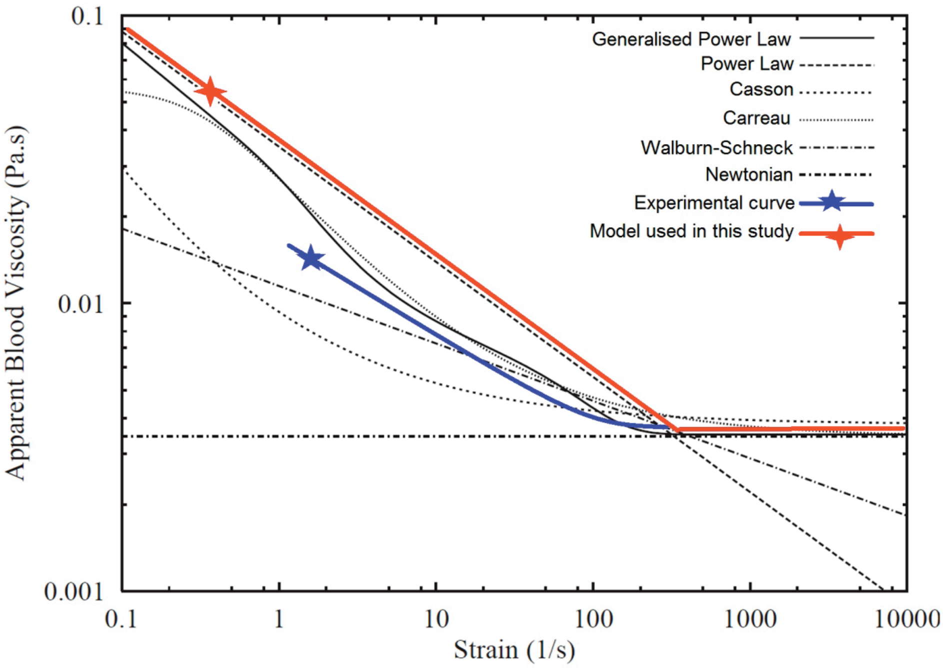

Fig. 4. Apparent blood viscosity vs. strain for different blood models [7], [8]

flow was assumed to be isothermal. Thus, energy dissipation and heat conductivity are neglected.

Blood flows observed in real anastomoses are usually turbulent, therefore the Shear Stress Transport (SST) model of turbulence was used in this study.

3. Results

For the seven cases investigated in this study, the results that allow one to analyse velocity, flow patterns and the WSS were obtained.

The results from the blood flow patterns in the fistula (anastomosis and veins) are shown in Fig. 5a. The maximum value of velocity of blood in the whole domain was observed in the anastomosis and it varied from 1.28 to 2.66 m/s (average 1.66 m/s) during one cardiac cycle (Fig. 5b). The blood velocity reaches high values also in the cannulated vein stenosis but this is a narrowing of the vessel, thus an increase in the blood velocity value was expected (Fig. 7a).

A turbulent flow with large vortices and back flows are observed in the aneurysm just downstream of the anastomosis (Fig. 6a). Similar phenomena, depicted in Fig. 7a, occur downstream of the cannulated

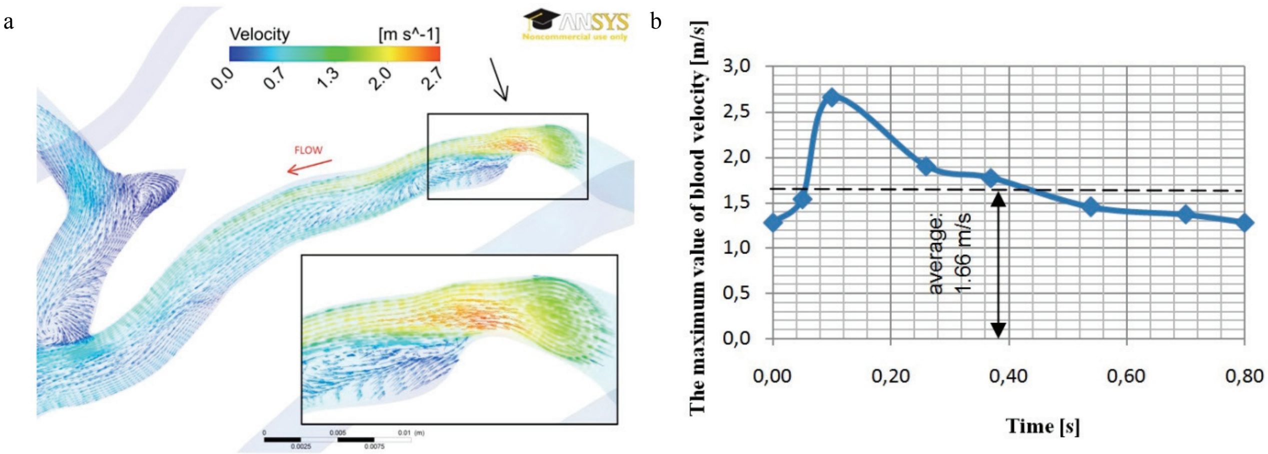

Fig. 5. (a) – visualization of blood flow patterns near the anastomosis (black arrow) extracted from simulations using the systolic velocity of blood at the inlet (case C); (b) – changes in the maximum value of blood velocity in the anastomosis versus time during a single cardiac cycle

vein stenosis where blood velocity rapidly decreases and the blood flow is locally reversed.

The WSS was investigated on the walls of the radial artery, the anastomosis and veins. The maximum value of the WSS, depicted in Fig. 6a, was observed in the anastomosis and it changes within a single cardiac cycle from 27.9 to 71.3 Pa (average 41.5 Pa). The second location of the area of high values of the WSS was found near the vein stenosis. Figure 7b shows a distribution of the WSS near the stenosis obtained for the case C for which the blood systolic velocity was used as the boundary condition at the inlet. The WSS distribution is irregular and asymmetric around this place and its maximum values change during a single cycle of the heart operation from 14.5 Pa to 36.4 Pa (average 21.5 Pa).

The flow rate at the inlet of the fistula investigated was 18.3 cm^3/s and this value should provide adequate haemodialysis [2]. From the viewpoint of haemodialysis efficiency, the most important parameter describing the blood flow through the fistula is a flow rate measured in the cannulated vein. Thus, on the basis of the blood velocity values obtained for the outlet of the punctured vein in the simulations performed and knowing the diameter (7.5 mm) of the outlet, an average value of the blood flow rate was determined. It equals 4.3 cm^3/s.

4. Discussion

The objective of this study was to evaluate and quantify the blood flow through the 3-dimensional model of end-to-end fistula used in haemodialysis on the basis of the patient's specific vessel spatial shape. A possibility of modelling the complicated geometry of the blood vessels involved, taking into account the

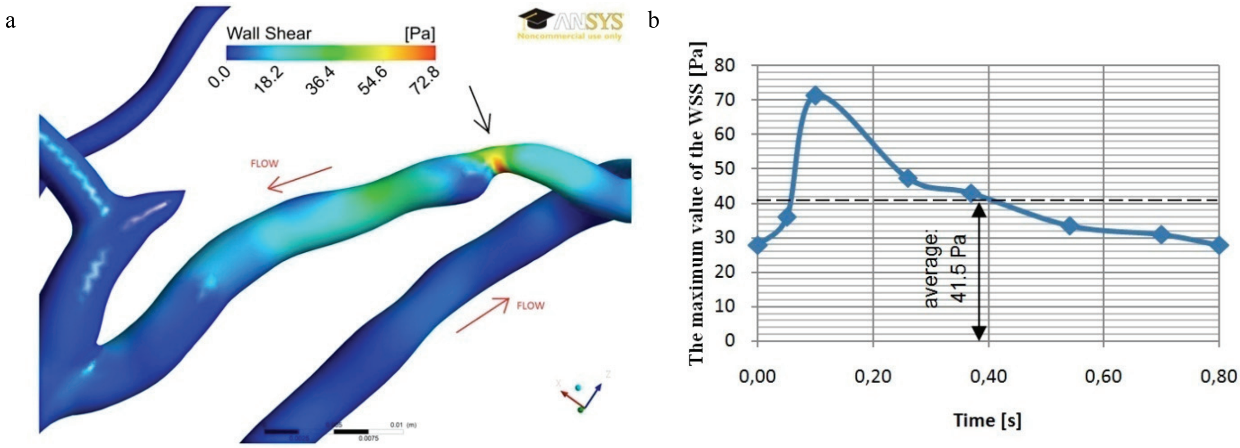

Fig. 6. (a) – visualization of the WSS near the anastomosis (black arrow) extracted from simulations using the systolic velocity of blood at the inlet (case C); (b) – changes in the maximum value of the WSS in the anastomosis versus time during a single cardiac cycle

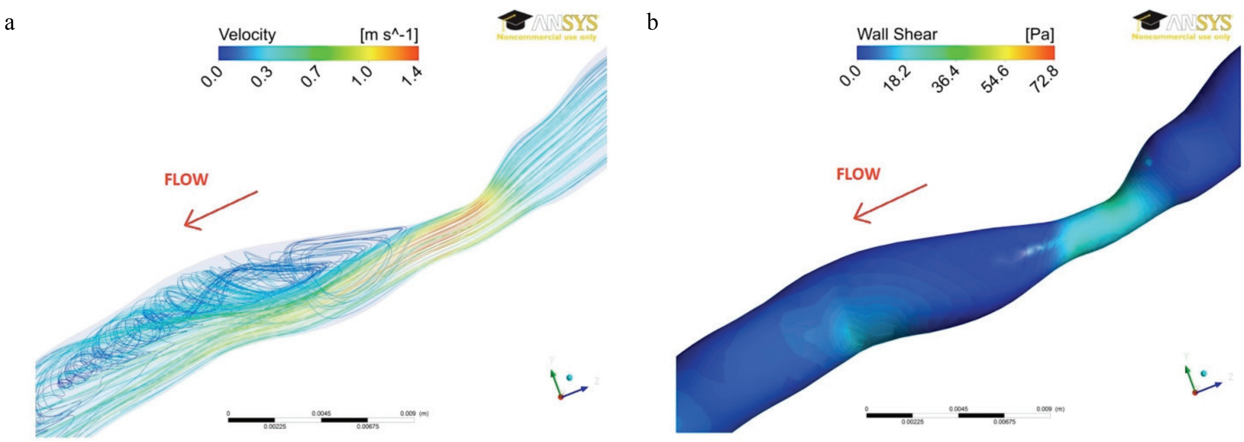

Fig. 7. Visualization of the blood flow – streamlines – (a) and the WSS (b) near the cannulated vein stenosis extracted from simulations using the systolic velocity of blood at the inlet (case C)

feeding artery (radial artery), the anastomosis of veins receiving blood was confirmed.

The flow in the anastomosis and the stenosis of the cannulated cephalic vein was turbulent and flow patterns that occur in those places were strictly connected with their geometrical shape. The anastomoses and stenoses are considered to be places of occurrence of clot risk because the blood velocity rapidly decreases and the blood flow can be even reversed locally around them. The clot formed, for instance, in the vortex centre, can be transported by the blood flow and may clog smaller blood vessels in the brain or the lung and cause instantaneous death of the patient. Narrowings cause also a pressure drop that usually leads to further thrombosis and, consequently, to a reduction of the diameter of the vessel. This may be the reason of decrease in the flow rate and can make the fistula completely useless for haemodialysis.

The velocity fields investigated in this study provide one with some knowledge about the possible values of blood velocity in the fistula with an emphasis on the anastomosis. The average maximum value of blood velocity during a single cardiac cycle obtained in this investigation is 2.66 m/s. Thus, it confirms the information found in the literature about extremely high values of blood velocity that can occur in anastomoses [1], [3]. However, one should remember that rigidity of the blood vessel walls was assumed in this study. This assumption means that the energy is not accumulated during the contraction phase of the heart. If the deformability of blood vessel walls had been taken into account, we could have obtained more accurate results. Nevertheless, wall deformation would play a more important role in the non-stationary analysis and, for the steady state, this assumption of wall rigidity is adequate.

The oscillating wall shear stress and its abnormal values are thought to be a fundamental factor influencing the endothelium and to play a key role in the development of intimal hyperplasia. This means a biological effect of a dramatic change in hemodynamic conditions after creation of the fistula and always leads to its dysfunction. The average maximum value of the WSS obtained for the anastomosis is 41.5 Pa. This order of magnitude is consistent with other results obtained from the in vitro tests or the blood flow simulations through the fistula models that have been found in the literature [1], [3]. Thus, the steady state solution may be used whenever limited computational resources are achievable or whenever short response time is a key factor. This may be important when clinicians want to consider numerous variants of the fistulas that may be developed in the particular patient.

CFD methods may be helpful in choosing the best geometry and location of the fistula and may support its maintenance. However, further investigations should be devoted to various fistula models with an assumption of time-dependent pressure and blood velocity and deformability of vessel walls. Additionally, a blood velocity waveform and time-dependent pressure should be measured for the selected cross-sections of the real fistula and be employed as boundary conditions. The results should be compared with the experimental data obtained from ultrasonic imaging (colour Doppler sonography). Finally, the fistula examined should be investigated after one month or longer time to check if any complications are formed. Thus, a method should be developed to provide a deeper insight into the correlation between the results obtained in simulations and measurements but also to provide one with some new knowledge on the formation of common vascular complications.

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