Structure of the secondary flow in the bifurcation of a blood vessel: patient-specific modeling and clinical Doppler measurements

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Abstract. The present contribution is aimed at patient-specific and clinical study of the secondary flow in the bifurcation of a blood vessel. Flow visualization is performed both with the ultrasound color Doppler imaging mode and with CFD data postprocessing of the flow in a carotid artery model with narrowing (stenosis). Special attention is paid to obtaining data for the secondary motion in the internal carotid artery. There was a good agreement in the results obtained between the patient-specific modeling and clinical measurements.

1. Introduction

The advancements in numerical simulation in recent years has led to aid the investigation of cardiovascular diseases. These studies shall be helpful to clinicians or physiologists to understand the mechanical environments in normal and diseased arteries. Especially, the flow in regions, such as bifurcation or arterial curvature is quite complex and more prone to development of atherosclerosis. The flow behavior through healthy artery is quite different in contrast to the stenosed artery with elevated stresses and high resistance to flow. The study of such important physiological simulation of flow through stenosis has profound implications for the diagnosis and treatment of vascular disease. The much required observation of flow behavior in critical areas such as bifurcation, carotid bulb, flow separation or turbulence in realistic anatomical models is possible only through patient-specific flow modeling [1,2]. A reliable flow simulation requires the realistic 3D vascular geometric model and unsteady flow boundary conditions. The geometry data is obtained through in-vivo measurements such as MRI slides, ultrasound and angiogram data such as CT, DSA and x-ray [3,4]. There is a good agreement in the results obtained between the numerical simulation and phantom experiments [5]. However, studies aimed at validating patient-specific models by means of comparisons with clinical measurements of the blood flow structure are practically not encountered. Hence, in the present study...
hemodynamics is studied in patient-specific model considering a case study of patient diagnosed with partial narrowing (stenosis) of internal carotid artery. The aim of this work is to compare vortex structures of blood flow, measured clinically by ultrasound Doppler and calculated numerically for the patient-specific model of artery bifurcation with stenosis.

2. Methods

2.1. Image processing
In the present study, patient is taken up whose left carotid system is normal and right common carotid artery is also normal with partial narrowing of internal carotid artery. External carotid artery appears to be normal. The partially stenosed internal carotid artery is shown in Fig. 1.

The geometry of the carotid model was constructed in several stages using the ICEM CFD software that is a part of the ANSYS Workbench platform. First, angio data images of the carotid bifurcation in two mutually perpendicular planes (Fig. 1a) were digitally segmented into several arcs. Equally spaced points were created on the arcs and then two corresponding points from each arc in planes were unified in one point in space. Then smooth space curve was drawn through the points using the 3D-Spline tool. This curve served as an axis for the carotid model. As the final stage, a cylindrical surface simulating the inner wall of the artery was constructed (Fig. 1b).

An angiographic study of the patient showed that the carotid bifurcation under consideration has a spatial curvature, an atherosclerotic plaque is located on the internal carotid in an asymmetric manner, covering the vessel by 64% over the diameter and 90% over the diameter, which corresponds to the case of severe stenosis.

2.2. Ultrasound visualization and quantitative evaluation of swirling blood flow in carotid bifurcation
Swirling blood flow registration and estimation method using the ultrasound Doppler technique was developed and applied by the authors [6]. With the help of this method the authors measured axial and circumferential components of blood velocity for stenosed carotid artery in the instance of maximum flow rate. The axial velocity component was measured in color Duplex imaging mode by registering impulse-wave Doppler spectra in artery longitudinal section using a traditional technique (Fig. 2a).

The circumferential velocity component evaluation was carried out in the color Duplex imaging mode by registering impulse-wave Doppler spectra in an artery cross section, with a sample volume placed in turns into the lateral and medial hemicircles of the artery lumen (Fig. 2b). The sample volume size corresponded to vessel radius, and the angle between the blood flow direction and the ultrasound beam was set to 0°. Circumferential velocity of the blood flow was measured for each position of the sample

Figure 1. Angiograms of stenosed carotid bifurcation in two mutually perpendicular planes (a); 3D stenosed carotid bifurcation (b)
volume with the following averaging of the circumferential velocity values between the lateral and medial hemicircles of the artery.

Figure 2. Doppler ultrasound visualization of swirling blood flow in a human common carotid artery: evaluation of axial velocity component in color Duplex imaging mode (at the top – image of artery longitudinal section, at the bottom – impulse-wave Doppler spectra of blood flow) (a); blood flow circumferential velocity component registration in color Duplex imaging mode (red and blue coloring of blood flow in lateral and medial hemicircles of an artery cross section) (b)

2.3. Mathematical model and computational aspects

Governing Equations

The numerical simulation of the 3D pulsatile flow in the model of the carotid bifurcation with 90% stenosis was carried out. The arterial wall is assumed to be rigid. Actual flow past the stenosis was in transition from laminar to turbulent for Reynolds numbers exceeding a certain critical value. Computations based on the Reynolds-averaged Navier–Stokes equations were made taking into account the results of our clinical measurements showing that there were intense velocity pulsations past the stenosis.

Governing equations are the continuity and Reynolds-Averaged Navier-Stokes equations (RANS):

\[
\begin{align*}
\text{div}\mathbf{V} &= 0 \\
\frac{\partial\mathbf{V}}{\partial t} + (\mathbf{V} \cdot \nabla)\mathbf{V} &= \frac{\mu}{\rho} \nabla^2 \mathbf{V} - \frac{1}{\rho} \nabla p + \text{Div}(\Pi) \\
\Pi_y &= -u_i u_j',
\end{align*}
\]

(1)

A widely used k-\omega SST turbulence model was chosen to close the problem formulation.

\[
\begin{align*}
\Pi_y &= \frac{2\mu}{\rho}(S_y - \frac{1}{3} S_{ij} \delta_{ij}), \\
S_y &= \frac{1}{2}(\frac{\partial v_i}{\partial x_j} + \frac{\partial v_j}{\partial x_i}), \\
\frac{\partial k}{\partial t} + (\mathbf{V} \cdot \nabla)k &= \nabla \left[ \frac{\mu}{\rho} + \alpha_s v_i \right] \nabla k + F_k - \beta' \omega k \\
\frac{\partial \omega}{\partial t} + (\mathbf{V} \cdot \nabla)\omega &= \nabla \left[ (\nu + \sigma_s v_i) \nabla \omega \right] + \frac{2}{\nu} F_\omega - \beta \omega^2 + (1 - F_\omega) \frac{2\sigma_s}{\omega} (\nabla k \cdot \nabla \omega)
\end{align*}
\]

(2)
Boundary Conditions and Numerical Procedure
A swirl velocity profile and a variation in the mean flow velocity during the cycle were specified at the inlet boundary (Fig. 3). The mean velocity curve was obtained from the clinical measurements of blood flow in patient by ultrasound Doppler method. The cycle period is 1 s. The velocity increase phase makes up 15% of the total cycle time. The maximum mean flow velocity for the period is 0.7 m/s. The ratio of the maximum circumferential velocity to the maximum axial velocity of the for inlet swirl velocity profile is 0.3. A constant pressure was specified at the outlet of external carotid artery and the mean velocity curve was specified at the outlet of internal carotid artery.

![Figure 3. Variation in the mean flow velocity during the cycle in common carotid artery](image)

Even though blood flow is non-Newtonian physiologically, however in the present study, since the focus in on large arteries, Newtonian assumption is acceptable as relatively high shear rate occurs. The dynamic viscosity coefficient is 0.004 Pa s, the density is 1000 kg/m$^3$. The Reynolds number at the maximum flow rate, based on the inner vessel diameter and the velocity of the mean fluid flow, is 1050.

Geometry and grids were built in ICEM CFD v.16. The finite-volume method was employed to solve RANS equations along with SST turbulence model. Computational domain was discretized into ‘mixed’ mesh with both structured zones with hexahedron cells and unstructured zones with tetrahedron cells. Total amount of cells is approximately 2,600,000. The simulation of carotid artery is carried out for 3 pulse cycle and results in the last cycle is considered for the investigation. All calculations were performed with the ANSYS CFX v.16 software.

3. Results
The secondary flow evolution in the patient-specific model of the stenosed carotid bifurcation is illustrated by Fig. 4, showing tangential velocity field in cross sections along common and inner carotid arteries in the instance of maximum flow rate. Swirling flow is observed in common carotid artery. The computations revealed that Dean vortex pairs in which the fluid rotates in opposite directions form at the ostium of the internal carotid artery. The Dean vortices have the form of two symmetric structures elongated along the outer wall of the inner carotid artery. In front of stenosis flow with Dean vortices is transformed in a converging flow. Immediately downstream the stenosis two asymmetric vortices arise; one of them significantly exceeds the other in size and intensity, and generates a swirling flow.
Clinical (in vitro) data are presented in the form of ultrasound images obtained by color flow mapping mode for measurement of circumferential Doppler velocity (Fig. 5). Numerical results are presented in the form of distributions of the velocity vector projections on the axis of the ultrasonic transducer. Comparison of the fields of circumferential Doppler velocity component, obtained from clinical measurements and through patient-specific modeling, showed good qualitative coincidence of the data.
Fig. 6 shows the numerical and clinical variation of maximum axial and circumferential Doppler velocity components along the common and internal carotid arteries. The axial Doppler velocity variation is characterized by local maximum at the stenosis of the internal carotid artery. The circumferential Doppler velocity variation has maximum on distance nearly 6 mm downstream of stenosis in section where swirl forms; downstream swirl decreases. Discordance of the results does not exceed 20% for axial velocity and 30% for circumferential Doppler velocity.

Figure 6. Variation of maximum axial (a) and circumferential (b) Doppler variation along the common and stenosed internal carotid arteries in the instance of maximum flow rate

4. Conclusions
Using a clinical data with ultrasound measurements allows validating patient-specific model of artery bifurcation with stenosis. The computations revealed that Dean vortex pairs form at the ostium of the internal carotid artery. In front of stenosis flow with Dean vortices is transformed in a converging flow. Immediately downstream the stenosis two asymmetric vortices arise; one of them significantly exceeds the other in size and intensity, and generates a swirling flow. Clinical and numerical results provide a qualitative agreement of secondary flow structures. The measurement error does not exceed 20% for axial velocity component and 30% for circumferential velocity component.

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