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Mesh Generation and Blood Flow Simulation in Human Carotid Bifurcation

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Abstract

The ability to associate blood flow features with parameters representative of the vessel geometry is fundamental to perform studies that might help surgeons understand the development of artery diseases. A tool to generate structured computational meshes for vascular modeling frameworks and a three dimensional finite element model for blood flow simulation is presented here. Blood flow is described by the incompressible Navier-Stokes equations and the simulation is carried out under pulsatile conditions. Results of patient-specific reconstruction with structured meshing of the left carotid bifurcation using Doppler ultrasound images together with blood flow simulations demonstrate the performance of the developed methodology.

Keywords: blood flow simulation, finite element method, computational fluid dynamics, Doppler ultrasound imaging.

1 Introduction

Atherosclerosis is a focal chronic inflammatory fibro-proliferative disease of the arterial intima caused by the retention of modified low-density lipoprotein and by hemodynamic stress. The accumulation of plaques on the artery wall is a progressive disease facilitated by local irregular flow field. The diagnosis of atherosclerosis is one of the most important medical steps for the prevention of cardiovascular events, like myocardial infarction or stroke. The assessment of atherosclerotic lesions is based on the measurement of the local arterial lumen diameter from medical images (angiography, MRI, ultrasound).

In recent years non-invasive medical imaging data acquisition made feasible to construct three dimensional models of blood vessels [1]. Validated computational fluid dynamics models using data obtained by these currently available measurement techniques can be very valuable in the early detection of vessels at risk and prediction of future disease progression. Since extra-cranial carotid arteries are superficial, it is quite suited for medical image acquisition using ultrasound devices [2, 3]. Furthermore, ultrasound is inexpensive, widely accessible, fast and safe. Once ultrasound images are captured in real-time the structure and blood flowing through blood vessels can also be analyzed. The definition of high quality surfaces of blood vessels is crucial to guarantee correct numerical results of blood flow simulations.

In order to set up a tool for clinical purposes, it is worthwhile reconstructing the individual vascular morphologies from medical imaging data and then simulating the blood flow in such geometries [4]. In this paper a tool to generate suitable structured hexahedral meshes for vascular modeling frameworks from Doppler ultrasound images is presented. Then, a three dimensional finite element model for blood flow simulation in large arteries is considered [5]. In finite element models, hexahedral meshes are often thought to provide the highest quality solution although generally more difficult to generate presenting an expensive associated operator intervention time. Nevertheless, hexahedral meshes compared to tetrahedral/ prismatic meshes converge better, and for the same accuracy of the result less computational time is required [6].

The finite element model for blood flow simulation in the carotid artery bifurcation [5] considers blood flow described by the incompressible Navier-Stokes equations and the simulation is carried out under pulsatile conditions. The numerical procedure for the transient non-Newtonian equations uses the Galerkin-finite element method and a fractional-step method for the integration in time [6]. At each time step Picard iteration is applied to linearize the non-linear convection and diffusion terms.

The accuracy and efficiency of the blood simulation is validated comparing numerical results with experimental data collected during clinical practice. The experimental data includes left carotid artery bifurcation geometry, blood flow velocities as well as the transient flow waveform defining the cardiac cycle obtained by ultrasound measurements. Simulation results demonstrate a very good spatial and temporal agreement when compared with experimental data at different locations of the carotid bifurcation.

2 Carotid bifurcation model

Report on the development of a semi-automatic methodology for patient-specific reconstruction and structured meshing of the left carotid bifurcation using Doppler ultrasound images is described. For the methodology to be applied, the vessel bifurcation needs to be nearly-planar, meaning that there is at least one direction (usually the bifurcation axis) such that all possible lines oriented in that direction intersect the vessel in none or one single cross-section.

Three main steps involved in developing the carotid bifurcation model with be described in the following sections: 2.1 acquiring the in vivo three dimensional anatomical data of the diseased arterial segment via ultrasound imaging; 2.2 image surface reconstruction of the external wall of the vessels under study; 2.3 structured mesh generation inside the arterial walls.

2.1 Ultrasound scans for anatomical geometry

The arterial geometry was based on in vivo image data obtained with minimum volume at diastolic pressure. The left carotid arteries of an elderly symptomatic patient were scanned using a digital ultrasound acquisition system, consisting of a standard commercial colour ultrasound scanner (General Electric *vivid e*). Both B-and power mode images were obtained to gather artery geometry and blood velocity spectra. Power mode images comprise a grayscale B-mode image with the power Doppler signal displayed as a coloured overlay. Longitudinal ultrasound images were complemented by cross-sectional ultrasound images obtained in sequence along the common carotid artery and the carotid bifurcation.

According to the protocol flow velocity was measured at different locations in the common carotid and also at internal and external carotid arteries. The Doppler signal was obtained from a longitudinal view and with the Doppler sample volume placed in middle of the lumen. Maximizing the peak velocity in the Doppler spatial velocity spectrum by moving the beam across the vessel guaranteed that the volume of interest included the central axis of the vessel lumen. Using this method, we ensured that the quantity measured was the instantaneous spatial peak velocity within the vessel. Ideally instead of centreline velocity, velocity profiles should be acquired and used as boundary conditions for the flow simulations. However, in clinical studies, the prolonged scan time of MR velocity measurement could be rather prohibitive whereas the well subject tolerated and easy access of Doppler ultrasound render it a more practical choice.

2.2 3D surface reconstructions

The external surface of an arterial wall is often invisible in a medical image and needs to be reconstructed based on the local lumen size. Figure 1 presents a 2D image of a quasi-planar carotid bifurcation obtained using B-mode ultrasound from an elderly person in July 2011.



Figure 1: B-mode image of a quasi-planar bifurcation

Areas of blood flow appear darker than surrounding tissue and their contours can be isolated by visual thresholding. At the image the medical doctor was able to make a rough measurement of the intima-media region boundaries and vessels diameters.

The delineation of the boundaries of the intima regions at the carotid walls was achieved by manual segmentation. The output of the algorithm often has irregularities caused by noise or defective detections of the carotid boundaries. After orienting the carotid artery surface along the principal axes of inertia, the user selects a set of points to define smoother lines around the bifurcation and carotid branches. In order to achieve that, on the boundary of the lumen specific points are considered in order to construct splines A, B, C and D defining an estimated lumen boundary for the Internal Carotid Artery (ICA), the External Carotid Artery (ECA) and the Common Carotid Artery (CCA) as shown in Figure 2. In order to guarantee that the surface reconstruction at the intersections of ECA and ICA with CCA are performed continuously, C and D splines are extended using auxiliary lines that are tangent to those splices at their intersection point.

The central axes of the CCA, ECA and ICA branches were defined by creating a curve associated to equidistant points from splines A to B, A to C and D to B and considering their tangent auxiliary lines. All three central axes are due to converge at one single point. Then, assuming CCA, ECA and ICA as circular vessels, except at their junctions, artery cross-sections were considered circular as a function of the length along each central axis with diameters given by the B-mode ultrasound measurements and complemented using cross-sectional ultrasound images. Few extra sections, with the same area of the last slice of the region of interest can be added outside a branch in the direction of the centreline, prolonging the surface with cylindrical extensions. This way, the achieved domain is extended by a few diameters at both arterial inlet and outlets to remove the direct influence of the boundary conditions on the bifurcation blood flow simulations.



Figure 2: Estimated boundaries of the carotid artery bifurcation

For the 3D reconstructing process of the carotid bifurcation geometry, crosssections of ICA and ECA junctions are the result of overlapped cross-sections, defining non-circular sections. Finally the surface of the carotid bifurcation is built with FEMAP function "*Aligned Curves*". Figure 3 exhibits the constructed vessel surface. The next step will be the generation of a hexahedral mesh filling this bifurcation surface. The surface is associated to a quadrilateral mesh with a user defined refinement.

2.3 Structured mesh generation

The surface domain was divided six parts and three confining cross-sections close to the bifurcation were created for each branch that was meshed maintaining finite elements continuity in each contact surface. The structured hexahedral mesh of the lumen of a nearly-planar carotid bifurcation was constructed using software FEMAP and is presented in Figure 4.



Figure 3: Surface definition of the carotid bifurcation



Figure 4: Structured mesh generation

The CCA, ECA and ICA branches are treated independently according to the following procedure: first, a 2D quadrilateral mesh is considered in a defined crosssection; then, by sweeping or extruding a 2D mesh of a section (quadrilateral) along a path, a volume mesh is generated (hexahedrons). At the intersections of ECA and ICA with CCA meshing is performed by superposing common nodes and elements. The lumen section exhibiting the square-based pattern (buttery) used at each quadrant of the inner part of the lumen is shown in Figure 4. By distributing the elements according to the desired accuracy, the local refinement of a hexahedral mesh has the potential to speed up highly demanding computations on large arterial territories.

3 Numerical simulation model

In order to set up a tool for clinical purposes, it is worthwhile reconstructing the individual vascular morphologies from medical imaging data and then simulating the blood flow in such geometries. In large arteries, the blood flow can be assumed to behave as a continuum, as well as incompressible, apart from severe pathological situations. The Navier-Stokes equations describe the mechanics of fluid flow. They state the dynamical balance between the internal forces due to pressure and viscosity of the fluid and the externally applied forces. Considering isothermal conditions the time dependent incompressible blood flow is governed by the momentum and mass conservation equations, the Navier-Stokes equations given as:

$$\rho \left(\frac{\partial \mathbf{u}}{\partial t} + \mathbf{u} \cdot \nabla \mathbf{u} \right) = \nabla \boldsymbol{\sigma} + \mathbf{f}$$
(1)
$$\nabla \cdot \mathbf{u} = 0$$

where **u** and σ are the velocity and the stress fields, ρ the blood density and **f** the volume force per unit mass of fluid.

For solving the fluid equations, the following boundary conditions were imposed. In an elderly person the walls of the arteries thicken, lose their elasticity, become stiffer and the effects of wall compliance can be neglected; then it appears reasonable to assume no slip at the interface with the rigid vessel wall. At the flow entrance (host artery) Dirichelet boundary conditions are considered assuming a parabolic distribution for the time dependent value \mathbf{u}_D of the velocity on the portion Γ_D of the boundary $\mathbf{u}(\mathbf{x},t) = \mathbf{u}_D(\mathbf{x},t)$, $x \in \Gamma_D$ at an outflow boundary Γ_N the condition describing surface traction force \mathbf{h} is assumed. This can be described mathematically by the condition:

$$\left(-p\delta_{ij} + \mu\left(\frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_j}\right)\right)n_j = h_i \quad i, j = 1, 2, 3 \text{ on } \Gamma_N$$
(2)

where n_j are the components of the outward pointing unit vector at the outflow boundary.

3.1 Finite element method (FEM)

A finite element approach for the solution of the time-dependent incompressible Navier-stokes equations is presented. In this work a spatial discretization with isoparametric brick elements of low order with trilinear approximation for the velocity components and element constant pressure is adopted. The numerical procedure for the discretization in space uses the Galerkin-finite element interpolation. The discretization in time is obtained by an implicit fractional stepmethod in which the time advancement is decomposed into a sequence of two steps [4, 5]:

- Calculation of an auxiliary velocity field $\mathbf{u}^{n+1/2}$ from the equations of motion, which include the viscous and convective effects; known pressure values from the previous time *n* step or previous iteration step *m* are used;
- Calculation of the end-of-step velocity field **u**^{*n*+1} and pressure **p**^{*n*+1} by the solution of the Stokes problem using lumped mass matrix. The method includes a diffusion term in the incompressibility step, which allows the imposition of the full boundary conditions for the velocity while needing no pressure boundary conditions [5].

Due to high Reynolds numbers, the treatment of the convective term needs a numerical scheme with a stabilization technique like the Streamline upwind/Petrov Galerkin method (SUPG-method) in order to avoid oscillations in the numerical solution [8, 9]. The SUPG-method produces a substantial increase in accuracy as stabilizing artificial diffusivity is added only in the direction of the streamlines and crosswind diffusion effects are avoided.

4 **Results and conclusions**

To show the potential of the numerical modelling to reproduce realistic flow fields relevant for medical investigations, simulated results obtained after application of the techniques discussed in the previous sections are presented here. One of the most evident features of blood flow in arteries is the periodic unsteadiness, or, more precisely, the pulsatility. This term refers properly to the feature of a first rapid increase and decrease of the flow rate, followed by a longer phase, when the flow rate becomes small and almost constant. So, the carotid flow pattern exhibits three different steps of the cardiac cycle: systole, deceleration phase and diastole. At CCA it is commonly accepted that the peak velocity lies along the vessel lumen central axis throughout the cardiac cycle.

Figure 5 illustrates the peak velocity waveform for one cardiac cycle given by ultrasound measurements at the distal common carotid artery section (DCCA), 1.5 cm before bifurcation as extracted from the Doppler spectra. The analysis presented here corresponds to Doppler data obtained in July 2011. As it can be seen, the waveform does not correspond to a healthy person. On the contrary, the observable decreased waveform flow indicates a carotid artery occlusive disease. In vivo data may contain artifacts due to both prescription errors and signal from other tissues near the vessel of interest. These artifacts cause errors in estimates of wave velocity. For a healthy person this problem can be solved at the expenses of data acquisition time and then taking an averaged waveform. With an unhealthy person this subject is not so easy since temporal variations might be due not only to involuntary movements but also to real flow obstructive problems.

Considering pulsatile blood flow with time dependent parabolic inlet velocity as given by this waveform simulation of blood flow in the reconstructed bifurcation artery geometry is performed using the developed software [5]. Convergence has been carefully tested. Refinement of both temporal and spatial resolutions was performed until changes in predicted velocities and wall shear stress became negligible. A mesh of 40 thousand hexahedrons (Figure 4) with trilinear approximation for the velocity components and element constant pressure was adopted.



Figure 5: Doppler velocity waveform at the distal common carotid artery

Calculated velocity values at four locations were extracted from the simulated velocity fields and compared with the Doppler experimental data. Peak velocity values are presented in Table 1 with the three first columns giving numerical simulation values and the last columns presenting the experimental measurements. Each line of the table represents a different location: APEX represents the bifurcation entrance, the proximal external carotid artery (PECA) corresponds to the most proximal point at the external carotid artery without flow disturbance from bifurcation (1.68 cm from DCCA) and PICA, the proximal internal carotid artery is situated 1.65 cm after DCCA, at the internal carotid artery.

Location	Max [mm s ⁻¹]	Min [mm s ⁻¹]	Min/Max	Data Max [mm s ⁻¹]	Data Min [mm s ⁻¹]	Data Min/Data Max
APEX	391.0	117.0	30%	384.1	107.1	28%
PECA	488.4	136.0	28%	596.6	106.4	18%
PICA	317.5	87.0	27%	347.8	122.7	35%

Table 1: Simulated and Doppler measured flow velocities

Comparing end-diastolic velocities with peak systolic velocities at each location, the computed ratios using the simulated values vary from 30% at APEX to 27% at PICA. On the contrary the ratios calculated using the Doppler data vary from 18% at PECA, 28% at APEX and 35% at PICA. Calculating the same ratio at DCCA the

obtained value is around 35%, as deduced from Figure 5. The Doppler waveform observed at DCCA is used as boundary entrance values for the simulations. Then, along the bifurcation the FEM simulations smooth that ratio to values just under to 30%. Velocity differences between observed and simulated values can be explained using different arguments. The first one is that the presented simulation results are taken exactly at the central axis of the simulated vessel where the systolic and end-diastolic velocities are maxima and that cannot be guaranteed during Doppler acquisition where axial location might be uncertain or averaged. Another argument comes from the uncertainty of defining the estimated boundaries of the carotid artery bifurcation. With a modified geometry the simulated results would be different.

Figure 6 presents the measured and simulated velocity waveforms, during one cardiac cycle at PICA. Again, the observable decreased waveform indicates a possible carotid artery occlusive disease but this subject is outside the scope of this article. The resemblances of both observed and simulated velocities and shapes are evident. At PICA the direct influence of the inflow pulse as given by the DCCA inflow pulse (Figure 5) is largely preserved. So, besides all possible complications along the carotid bifurcation, the shapes of the velocity waveforms at PICA and DCCA are similar.

As noticed in Table 1, both systolic and end-diastolic velocity values are higher in the collected data as compared to the simulated results. Most likely the uncertainty associated to the geometry definition is one of the reasons. In the near future, research will be devoted to analyze the uncertainties associated to local boundary variations in the computational domain.



Figure 6: Doppler (left) and simulated (right) velocity waveform [mm s⁻¹] at PICA

This study shows that blood flow simulation of the carotid bifurcation is in good agreement with the one obtained experimentally using Doppler ultrasound measurements in clinical practice. Further work is necessary to determine whether this technique can maintain the demonstrated performance and whether that performance and the achievable spatial and temporal resolution are sufficient to make clinically meaningful considerations.

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