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HEMODYNAMICS OF A STENOSED CAROTID BIFURCATION

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Abstract. A methodology for computational 3D reconstruction and structured hexahedral meshing for patient-specific hemodynamics analysis of the carotid artery bifurcation with a stenosis is presented. The purpose of this work is the use of anatomically realistic blood flow simulations by the finite element method (FEM) derived from in vivo medical imaging to make patient specific studies of flow phenomena associated with the development of atherosclerosis disease. Blood flow is described by the incompressible Navier-Stokes equations and the simulation is carried out under pulsatile conditions. The study of a diseased carotid bifurcation illustrates the extremely complex hemodynamical behaviour along the cardiac cycle.

1 INTRODUCTION

A long standing hypothesis that correlates fluid dynamic forces and atherosclerotic disease has led to numerous analytical, numerical, and experimental studies over the years. The observation that atherosclerotic disease is focal typically occurring at sites of complex hemodynamics, such as arterial bifurcations, junctions, or regions of high curvature inspired these studies. High wall shear stress (WSS), damage the arterial wall and regions of low or oscillatory shear stress cause monocyte adhesion to the endothelium, an early stage in atherogenesis [1]. The carotid bulb is one of the first sites in the carotid bifurcation to develop late atherosclerotic inflammation consistent with reports that show this to be a region of low WSS. Usually flow separation occurs at the carotid bulb due to the increase in cross-sectional area, being this region of separation greater during deceleration phase of systole, when the fluid undergoes the largest reversal of momentum.

Recent non-invasive medical imaging data acquisition made feasible to construct three dimensional models of blood vessels. B-mode ultrasound is a non-invasive method of examining the intima and walls of peripheral arteries providing measures of the intima-media thickness (IMT) at various sites (common carotid artery, bifurcation, internal carotid artery) and of plaques that may indicate early presymptomatic disease. It also allows measurements of blood flow velocities providing accurate information on flow fields. 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.

Computational modelling of blood flow in realistic arterial geometries has the potential to provide a complete set of hemodynamic data that cannot be acquired by measurement alone. This needs to be performed by combining the latest computational fluid dynamics approaches with the velocity measurements and flow images obtained using ultrasound techniques.

In this work flow characteristics in a patient-specific carotid bifurcation with a stenosis are investigated by using direct numerical simulation. A semi-automatic methodology for patient-specific reconstruction and structured meshing of the right carotid bifurcation is presented. As hexahedral meshes compared to tetrahedral/prismatic meshes converge better, and for the same accuracy of the result less computational time is required [2-4] a tool to generate suitable structured hexahedral meshes for vascular modelling frameworks from Doppler ultrasound images is considered.

Blood flow simulation models [5] using pulsatile inlet conditions based on in vivo colour Doppler ultrasound measurements of blood velocity, allow to compare numerical results with experimental data collected in clinical practice. The three-dimensional, unsteady, incompressible Navier–Stokes equations are solved with the assumptions of rigid vessel walls and constant viscosity (Newtonian fluid).

The ultimate aim of this study is the reconstruction of geometry and flow environment from in-vivo patient data, particularly at the extra-cranial carotid artery, using Doppler ultrasound data.

2 METHODOLOGY

To perform the computational investigation of patient specific arterial morphology and blood flow behaviour using a finite element code four steps are necessary: acquiring the in vivo anatomical data of the arterial segment, image surface reconstruction, 3D finite element mesh definition and blood flow simulation. Data was obtained in Hospital de São João, a university hospital in Oporto, Portugal. Informed consent of each volunteer was obtained using a protocol for the acquisition of a set of longitudinal and sequential transverse Doppler images and velocity measurements at carotid artery bifurcation. Using a standard commercial colour ultrasound scanner (General Electric vivid e) a set of longitudinal and transversal B-mode images of the common carotid artery (CCA), its bifurcation and proximal segments of internal carotid artery (ICA) and external carotid artery (ECA) of a patient was acquired This set of images cover the bifurcation region, about 10 cm long.

For each volunteer velocity measurements were made at different locations in the common, internal and external carotid arteries allowing the definition of the boundary conditions and the validation of the blood flow simulation.

The acquired Doppler ultrasound images made possible the 3D geometry reconstruction and mesh generation. A selected 2D longitudinal image shown in figure 1 was manually segmented by three medical experts and a rough outline of the intima-media region boundaries was defined and imported into the modeling commercial software FEMAP (FEMAP, Siemens PLM, USA & Canada).

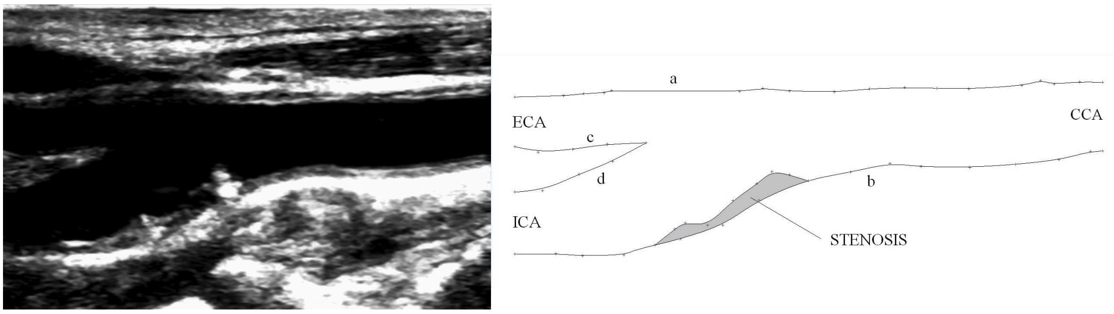

Figure 1: Carotid arterybifurcation of patient P1: (a) input image, (b) estimated boundaries

A computational 3D geometry reconstruction and a structured hexahedral mesh of the lumen were constructed. The centerlines of CCA, ECA and ICA were defined at the adopted mid-plane by creating a curve associated to equidistant points from splines a to b, a to c and d to b. Then, cylindrical geometries are assumed for CCA, ECA and ICA, except at their junctions, and their cross-sections modified according to the drawn lumen boundary and acquired cross-sectional ultrasound images. Artery surfaces are defined as vessels presenting curved axes and cross-sectional shape and diameter variability. As shown in figure 2 cross-sections of internal carotid artery and external carotid artery junctions are the result of overlapped cross-sections, defining non-circular sections.

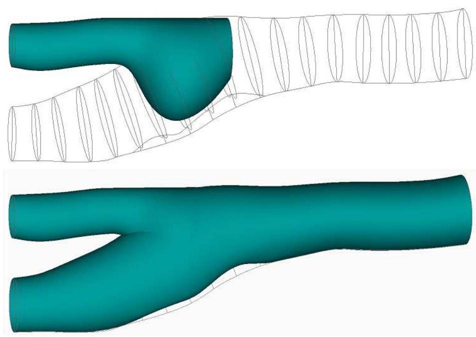

Figure 2: Definition of carotid bifurcation surface

Using software FEMAP, mesh generation of the previously defined surface was performed by dividing the domain in six parts being each part meshed independently and maintaining finite elements continuity at each contact surface as shown in Figure 3. The generation of the volume mesh with hexahedral elements started by building a 2D mesh (quadrilateral) on the confining cross-sections defined at the bifurcation, as artificial separations of CCA, ECA and ICA branches. Then CCA, ECA and ICA branches are treated independently by sweeping or extruding the 2D mesh in order to generate a volume mesh of hexahedrons.

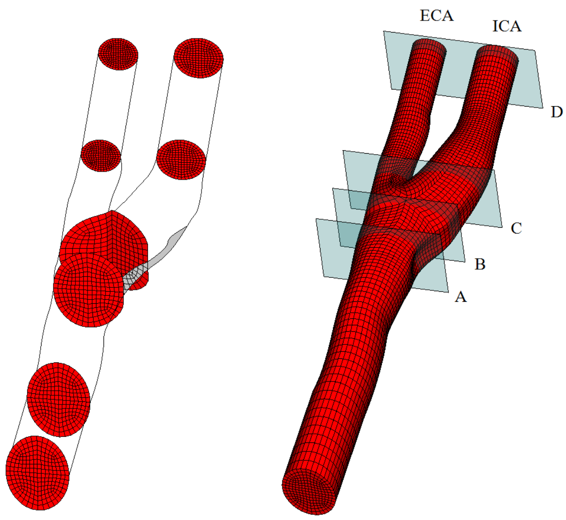

Figure 3: Structured mesh and cross-section slices for spatial velocity variation A–A, B–B, C–C and D–D (figure 4).

The use of computational meshes with well-organized elements along the main flow direction assures faster convergence and more accurate numerical solutions as blood motion in vessels is highly directional [3-5]. More accurate solutions are also obtained with a finer mesh in the bifurcation, near stenosis and near wall regions.

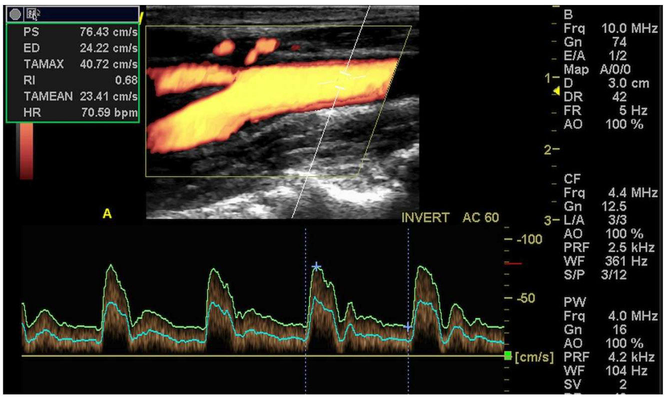

Figure 3: Measured flow wave form in the common carotid distal the flow divider

It is desirable to impose boundary conditions a few diameters upstream and downstream the region of interest therefore the polygonal surface obtained is not directly usable for generating a suitable computational mesh. Cylindrical 4-diameter extensions are added in the direction of the centerlines in order to reflect the vessels geometry as it approaches the domain of interest.

With the inlet flow corresponding to the systolic peak a mesh sensitivity analysis was carried out under steady conditions. Mesh refinement was performed until changes in velocities and maximum nodal WSS became less than 1.5%; a mesh of 55 thousand hexahedrons was chosen and a temporal convergence was performed with a temporal refinement until changes in velocities and maximum nodal WSS became insignificant between the adopted and finer time steps (less than 1.5%). A large amount of computational work is involved and with the chosen mesh the transient study was performed with a constant time step equal to $2.5\text{x}10^{-3}$s.

3 RESULTS AND DISCUSSION

The accuracy and efficiency of the blood simulation is validated comparing velocities given by numerical calculations with experimental data collected in clinical practice. In figure 5 velocities are compared with Doppler ultrasound measurements at different cross-section locations. At all positions in the carotid bifurcation there is a good agreement between the obtained flow velocities and those obtained experimentally in clinical practice as deviations are less than 5% at all positions.

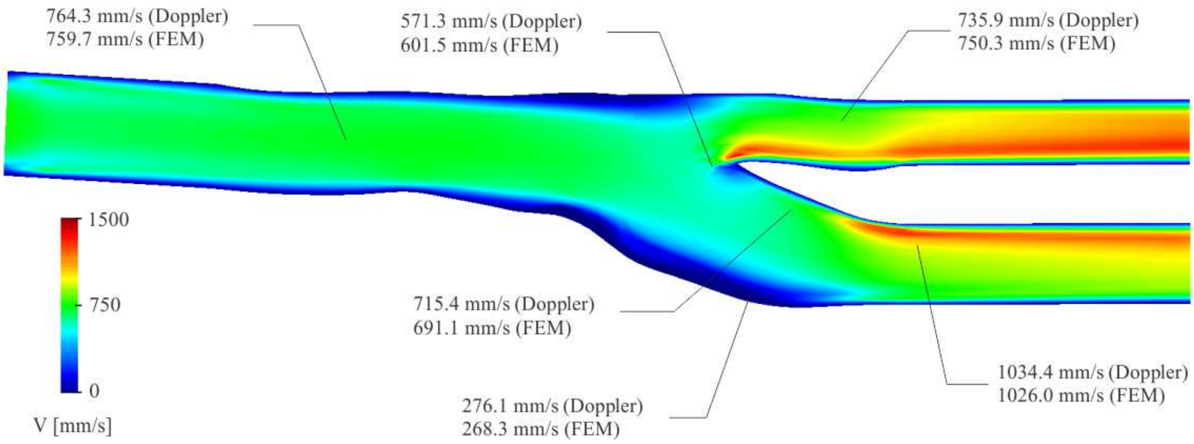

Figure 5: Velocity field in the mid plane of the bifurcation at systolic phase: Power-Doppler measurements and calculated velocities

Numerical velocity field in some sections is presented in Figure 6, for two cardiac phases, near peak systolic and at mid-deceleration phase. Within the stenosis, section A-A, no large velocity gradients can be noticed as there is no variation in lumen section. The increase in lumen section of the carotid bulb just downstream the stenosis contributes to the tendency for flow separation observed in sections B-B and C-C. This recirculation zone, downstream the stenosis is higher during deceleration phase due to the fact that flow parthens change drastically during this phase of the cardiac cycle. In the external carotid artery high velocity gradients occur in section C-C due to the apex-induced separation; high velocity gradients can also be observed in section D-D for both internal and external carotid arteries as the blood vessel diameter decreases significantly.

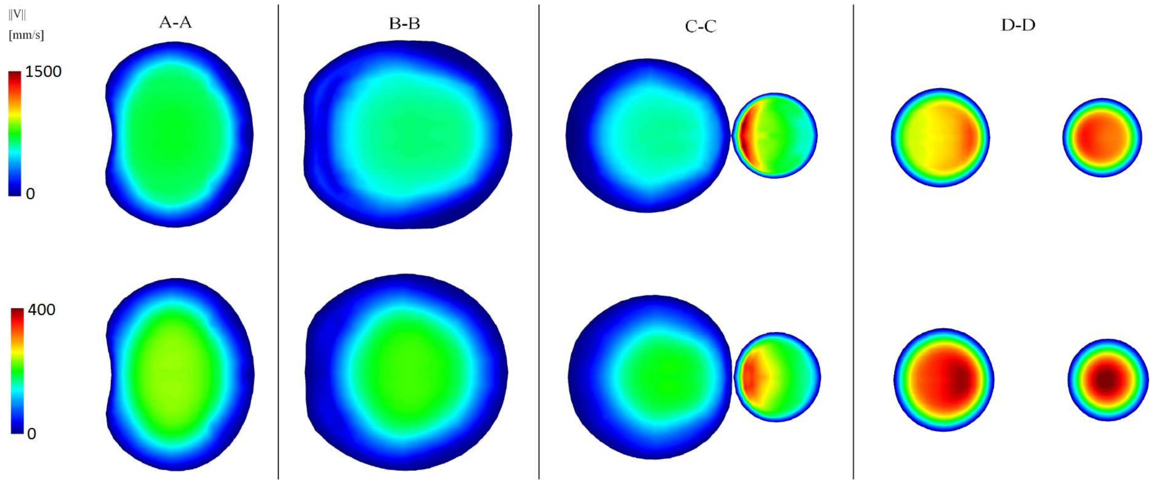

Figure 6: Velocity distribution at four sections near peak systolic and at mid-deceleration phase (section locations in figure 3).

In order to elucidate the role of carotid hemodynamics on plaque vulnerability WSS distribution at the two same flow phases is studied. In Figure 7 WSS distribution near peak systolic time shows high shear stress at the inner wall of the ECA, corresponding to high velocity-gradients with a skewed velocity profile. However the highest values are found at the inner wall of ICA, downstream the stenosis, due to the reduction of the lumen diameter.

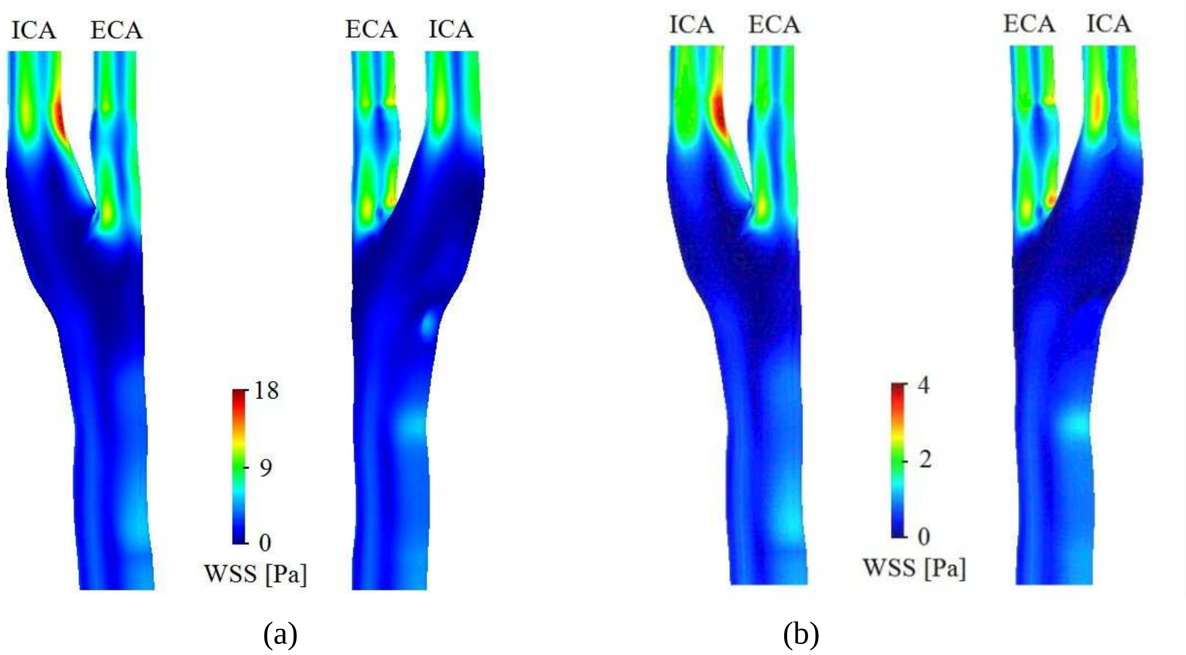

Figure 7: WSS distribution at two flow phases: (a) near peak systolic, and (b) diastole.

Recirculation zones are characterized by patches of low WSS in ICA, where the largest regions are associated with the apex-induced separation near the outer sinus wall and with the stenosis-induced separation along inner and outer walls. At diastolic time high shear stress zones can also be observed, however patterns at the two time instants are different providing further evidence that flow patterns change drastically during the cardiac cycle. One striking similarity prevails between both instants: the largest continuous region of low WSS seems to exist in the carotid bulb, where late atherosclerotic inflammation develops.

This work addresses the hemodynamical environment of a diseased carotid bifurcation concluding to be extremely complex during systolic phase and different from that at diastolic phase.

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