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STENOSED CAROTID BIFURCATION: HEMODYNAMICS AND WALL SHEAR STRESS

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ABSTRACT

The goal of this work is to present a semi-automatic methodology for patient-specific study of carotid bifurcation hemodynamics using Doppler ultrasound images. In the present study, flow characteristics in two patient carotid bifurcations with stenosis are investigated using direct numerical simulation. A good agreement between ultrasound imaging data and computational simulated results by the finite element method (FEM) was obtained and the study shows that the hemodynamical environment of diseased carotid bifurcations is extremely complex and significantly dependent on the severity of the stenosis

Keywords: carotid artery bifurcation, stenosis, blood flow simulation, Finite element method, Doppler ultrasound images.

INTRODUCTION

Atherothrombotic carotid stenoses, followed by ischemic strokes, are one of the leading causes of mortality and morbidity in western countries. Diseased vessels and regions downstream of severe constrictions, post-stenotic regions, experience a significantly different biomechanical environment with high wall shear stress (WSS) due to area reduction and flow acceleration.

Knowledge of the carotid hemodynamics could clarify the relationship between carotid artery stenosis and symptoms, and, ultimately, the risk of stroke. Blood flow simulations can be used to obtain detailed flow information, including wall shear stress, pressure drops, stagnation and recirculation regions, particle residence times, and turbulence. The local hemodynamics and consequently the wall shear stress are strongly influenced by the morphologic characteristics of the stenosis (Steinman, 2000) and the flow waveforms. The strong dependence of flow patterns on vessel geometry and physiologic conditions limits the use of idealized arterial models. Realistic modeling can yield new insights into the arterial blood flow and due to the large anatomic and physiologic variability among individuals, realistic patient-specific modeling is needed for the purpose of diagnosis and surgical planning (Antiga, 2008; De Santis, 2010).

The authors developed a method to construct realistic patient-specific finite element models of blood flow in carotid arteries from ultrasound images obtained experimentally in clinical practice. A semi-automatic methodology for reconstruction and structured meshing of the right carotid bifurcation is presented. Blood flow simulation models allow to compare numerical results with experimental data collected in clinical practice based on in vivo Doppler ultrasound measurements and images. Finite element simulations of fluid flow

(Sousa, 2012) are used to investigate inter-individual variations in flow dynamics at the carotid artery bifurcation with stenosis, in two volunteers.

METHODOLOGY

Computational investigation of patient specific arterial morphology and blood flow behaviour using a finite element code requires four sequential steps: acquiring the in vivo anatomical data of the arterial segment, image surface reconstruction, 3D finite element mesh definition and blood flow simulation. A set of longitudinal and transversal B-mode images of the common carotid artery, its bifurcation and proximal segments of internal and external carotid arteries of two patients, P1 and P2, was acquired using a standard commercial colour ultrasound scanner (General Electric vivid e). 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.

For each patient a selected good quality 2D longitudinal image 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). Fig. 1 shows the selected image for patient P1 and the estimated boundaries.

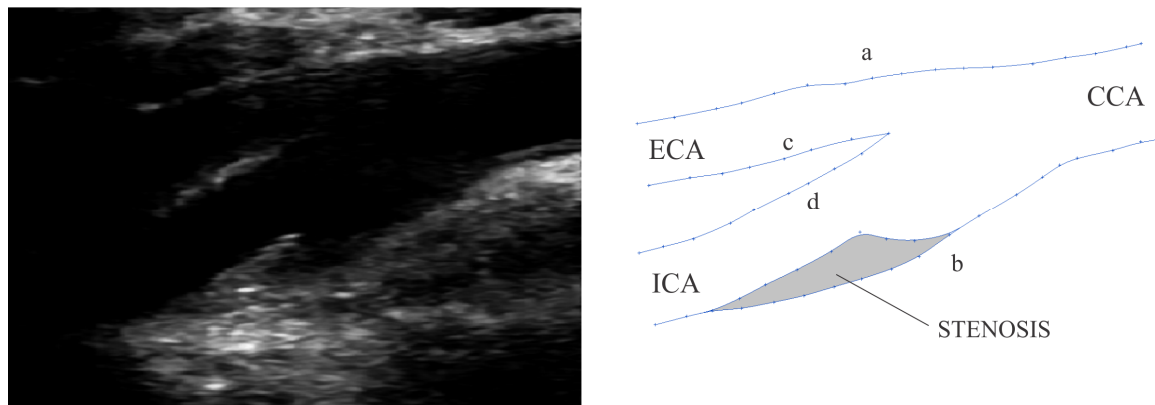


Fig. 2 Carotid artery bifurcation of patient P1: input image and estimated boundaries

Using the estimated bifurcation boundaries and frozen end-diastole transverse images a 3D geometry reconstruction and mesh generation was made in order to study carotid bifurcation artery hemodynamics. Carotid bifurcation surface definition and a structured hexahedral mesh of the lumen of the nearly-planar stenosed carotid bifurcation of volunteer P1 are presented in Fig. 2. As shown in this figure cross-sections of internal carotid artery (ICA) and external carotid artery (ECA) junctions are the result of overlapped cross-sections, defining non-circular sections. Mesh generation was performed by dividing the previously defined surface in six parts and being each part meshed independently using software FEMAP. First a 2D mesh (quadrilateral) was built on the confining cross-sections defined at the bifurcation, as artificial separations of the CCA, ECA and ICA branches. Then the generation of the volume mesh with hexahedral elements was made, by sweeping or extruding the 2D mesh defined in each contact surface maintaining finite elements continuity, as shown in Fig. 2. 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 (Antiga, 2008; De Santis, 2010; Verma, 2005).

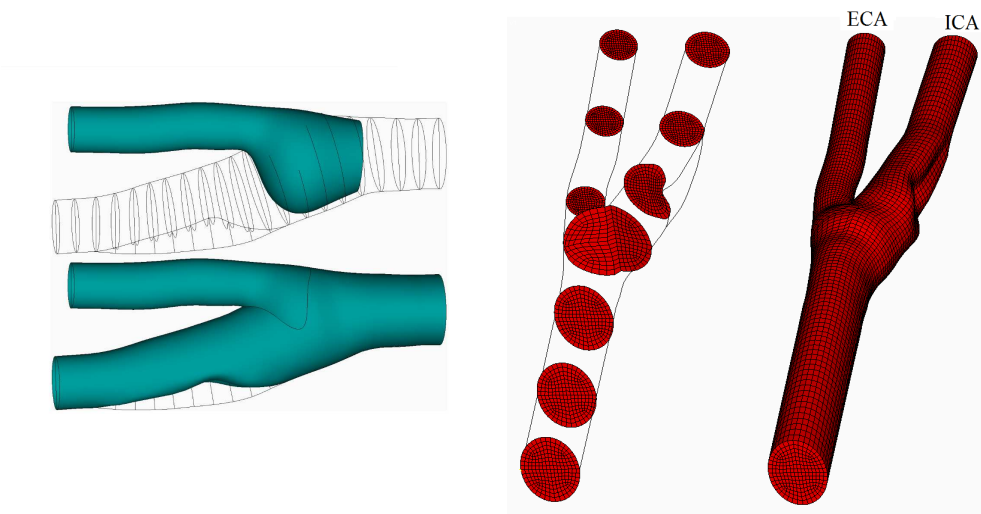


Fig. 2 Building the surface of carotid bifurcation and structured mesh of volunteer P1

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 flow extensions were created at both inlet and outlet locations in order to ensure fully developed velocity profiles at the inlet and to minimize the influence of outlet boundary conditions.

The time-varying 3D velocity fields were computed using an in-house developed code (Sousa, 2012). As commonly considered blood was modeled as an isotropic, incompressible, Newtonian viscous fluid, with a specific mass value equal to 1060 kg/m^3 and a constant dynamic viscosity value equal to 0.0035 kg/(m.s) . The backward Euler implicit time integration scheme was implemented to obtain the solution at each time step of the time-dependent problem and the upwinding method was applied to solve the nonlinear system of matrix equations derived from the discretization of the flow equations on the computational grid.

To assure grid independence a mesh sensitivity analysis was carried out under steady conditions with the inlet flow corresponding to the systolic peak. For each patient-specific carotid bifurcation mesh refinement was performed until changes in velocities and maximum nodal WSS became less than 1.5%; then, with the chosen mesh, a temporal convergence was performed with a time step refinement until changes in velocities and maximum nodal WSS became insignificant between the adopted and finer time steps (less than 1.5%). Due to the large amount of involved computational work we limited the transient study to the chosen mesh, adopting a constant time step equal to $2.5 \times 10^{-3} \text{ s}$.

RESULTS

The developed blood simulation code is validated comparing velocities given by numerical calculations with experimental data collected in clinical practice. For the patient P1 numerical velocity field in some sections is presented in fig. 4. High velocity gradients are observed at the inner wall of the ECA near the apex-induced separation.

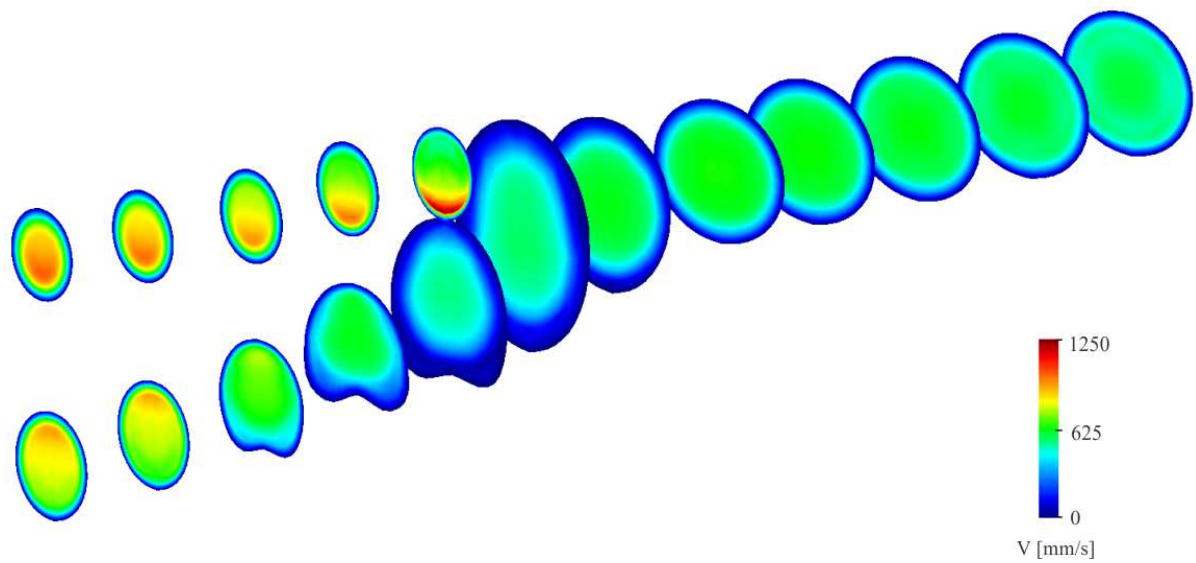


Fig. 3 Velocity distribution at systolic phase for patient P1

In fig. 4 and Fig. 5 calculated velocities for patients P1 and P2 are compared with Doppler ultrasound measurements at different cross-section locations. For both individuals, at all positions in the carotid bifurcation there is a good agreement between the obtained flow velocities and those measured experimentally in clinical practice. For patient P2 high velocity gradients can be noticed in ECA within and downstream the apex-induced separation; in ICA high velocity gradients are also found within and downstream the stenosis, being the hemodynamical environment quite different from that of patient P1 due to the larger lumen reduction.

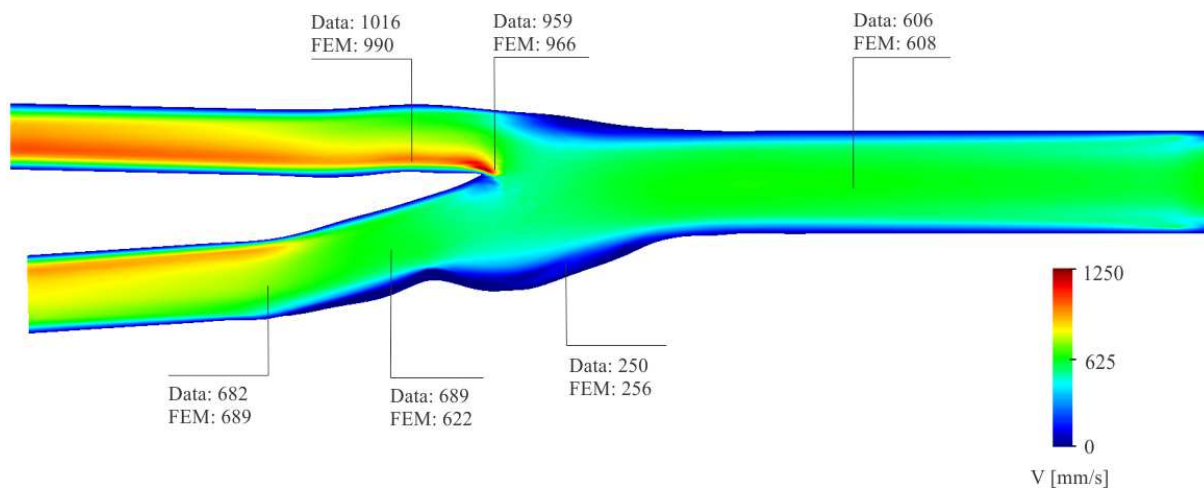


Fig. 4 Velocity field at the mid plane of the bifurcation at systolic phase for patient P1:
Power-Doppler measurements and calculated velocities (FEM)

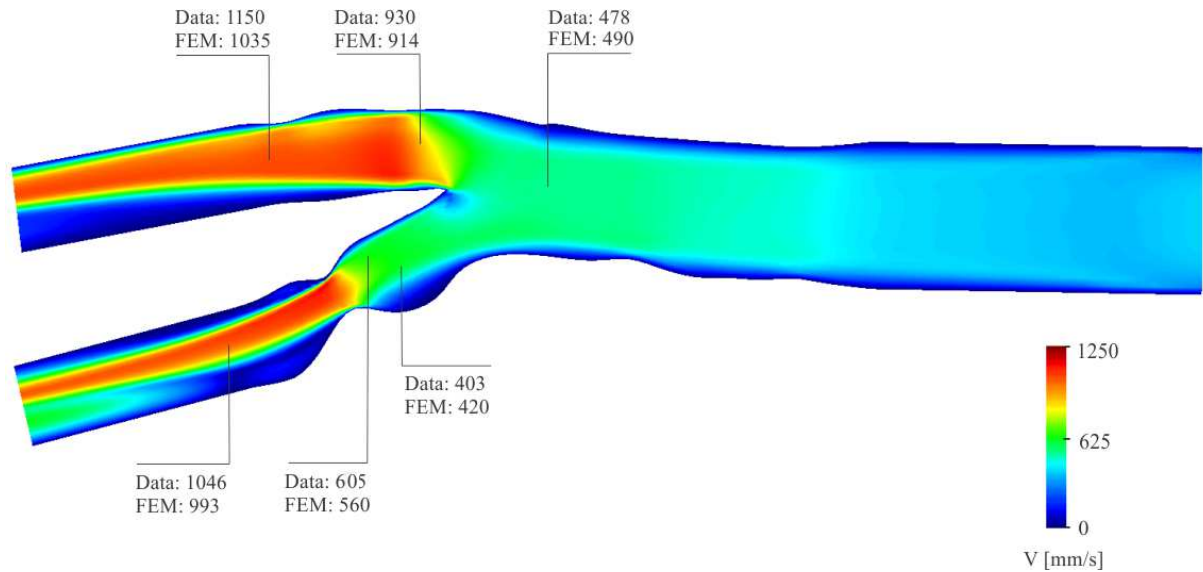


Fig. 5 Velocity field at the mid plane of the bifurcation at systolic phase for patient P2: Power-Doppler measurements and calculated velocities

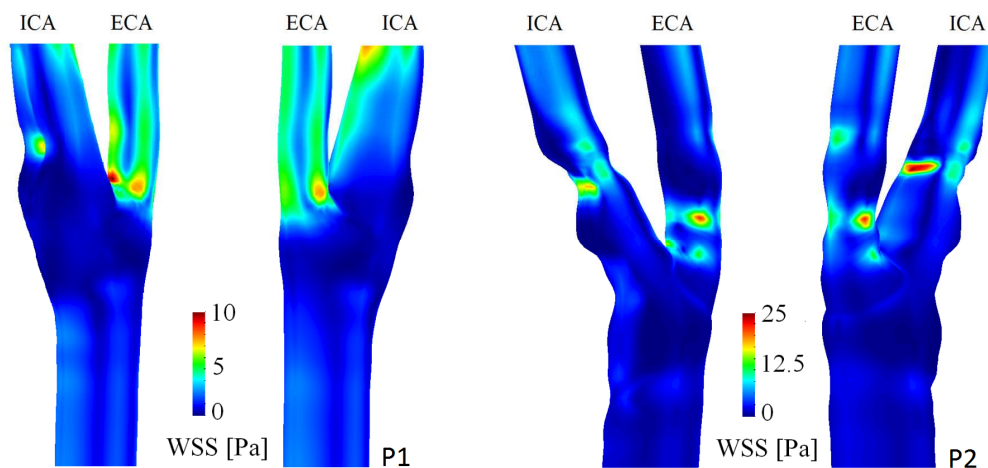


Fig.6 WSS distribution near peak systolic for two patients, P1 and P2 mesh

In order to elucidate the role of carotid hemodynamics on plaque vulnerability WSS distribution near peak systolic phase is presented at Fig. 6 for both patients. The WSS distribution shows high shear stress at the inner wall of the ECA, corresponding to high velocity-gradients with a skewed velocity profile. However for patient P2 the highest values are found at the inner wall of ICA, within the stenosis, also associated with high velocity gradients. For this patient P2 high WSS are also detected at the outer wall of ICA due to the prominent narrowing of the internal carotid artery. 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. These recirculation zones downstream the stenosis can be noticed in Fig. 6.

CONCLUSION

In this work flow characteristics in two patient-specific carotid bifurcations with ICA stenosis of different degree are investigated using an in house developed finite element code. Preliminary validation studies were made comparing numerical velocities with experimental

data based on Doppler ultrasound measurements.

The main features expected from fluid dynamics, such as low WSS values in the bulb region of the ICA with a return to ordered flow further downstream, were successfully captured. As expected, high WSS values are also observed at the inner wall of ECA. Furthermore for the stenosed ICA with greater degree of stenosis, large WSS values are observed within the stenosis, as it presents a greater narrowing of the internal carotid artery.

This work addresses the hemodynamical environment of two diseased carotid bifurcations concluding to be extremely complex during systolic phase and significantly different from each other mainly due to different degrees of stenosis.

The method needs to be tested further in cases of more severe degrees of stenosis, which involve more complex flow patterns. Although the method would still be feasible, an increase in the required computing time is expected, because the temporal and spatial resolutions must be increased to resolve flow features due to turbulence.

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