Impedance-based outflow boundary conditions for human carotid haemodynamics

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1. Introduction

Accurate numerical models are important and useful tools to understand atherosclerotic plaque growing-up, progression, vulnerability as rupture in human vessels. In particular, these can be used to evaluate wall shear stress (WSS), to compute wall stresses and to correlate flow phenomena and geometrical features with atherosclerotic pathologies (Ku et al. 1985). The latter is in fact not fully understood. It has been well accepted that low and oscillating flow shear stresses correlate positively with localisation (Perktold et al. 1991; Glagov et al. 1998; Moore et al. 1998, 1999; Ethier et al. 2000; Morbiducci et al. 2010; Gallo et al. 2012), some others analysed mechanical stresses induced by blood pressure using simplified or patient-specific carotid models (Delfino et al. 1997; Kaazempur-Mofrad et al. 2003; Steinman et al. 2003; Hariton et al. 2007). Carotid haemodynamics is very sensitive to geometrical factors as tortuosity, curvature ratio and bifurcation angle (Thomas et al. 2002; Lee et al. 2008). Due to the interindividual variability of geometry and flow dynamics (Thomas et al. 2005; Lee et al. 2008), this sensitivity is crucial and challenging for computational fluid dynamics (CFD) models and fluid-structure interaction (FSI) features. Coupling medical images with computing resources enables the study of blood flow in human arteries. This has been performed through CFD...
simulations and, more recently, through FSI analysis with the aim to evaluate the influence of vessel properties (e.g., geometrical properties and wall compliance) on arterial pathologies. Steinman et al. (2002, 2005) proposed a novel approach for non-invasively reconstructing artery wall thickness and local haemodynamics at the human carotid bifurcation, and reported the first direct comparison of haemodynamic variables and wall thickness. Lee et al. (2008) provided a statistical analysis of a correlation between disturbed flow and bifurcation geometry of several patients. Geometrical factors of several carotid bifurcations were analysed and compared with flow features by Thomas et al. (2002, 2005). On FSI analysis, Perktold and Rappitsch (1995) investigated the effect of a distensible artery wall on the local flow field and determined the mechanical stresses in the artery wall, modelling through the non-linear shell theory, where incrementally linearly elastic behaviour is assumed. Tang et al. (2001, 2003, 2008); Tang, Yang, Kobayashi, Ku 2004 and Tang, Yang, Kobayashi, Zheng, et al. 2004, conducted extensive research on stress analysis in plaque MRI-based models. Gao et al. (2009) compared the differences in stress distribution on plaque location between different diseased carotid bifurcations, and analysed the impacts of the specific combination of fibrous cap thickness and lipid core volume on the stress distribution (Gao and Long 2008). Kock et al. (2008) provided a 2D-FSI study on MRI-based carotid geometry while Yang et al. (2007) focused their work on MRI-based plaque stress analysis. Through FSI analysis, Younis et al. (2003, 2004) tried to validate the hypothesis that low or oscillatory shear and high maximum principle CS correlate with known inflammation sites in the carotid bifurcation, studying also the effects of exercise on haemodynamics in the bifurcation.

CFD and FSI simulations are inherently influenced by the choice of boundary conditions, and imposing either measured or idealised velocity distributions strongly influence the resulting flow field (Moyle et al. 2006; Wake et al. 2009). The boundary condition for a compliant vessel including a carotid bifurcation has been systematically addressed by Formaggia et al. (2001, 2003, 2009), using 1D and 0D approximations of vasculatures at the exit. Milner et al. (1998), Ford et al. (2005) and Wake et al. (2009) among others studied the effect of the inlet velocity condition on carotid flow field. As outflow conditions, CFD studies were used to apply measured flow or velocity rate or normal traction-free conditions (Moyle et al. 2006; Lee et al. 2008; Wake et al. 2009), whereas for FSI computations, pressure (Tang et al. 2008) or combined velocity/pressure conditions (Gao et al. 2009) were used.

The aim of this work was to provide physiological pressure boundary conditions for carotid haemodynamics using the impedance method developed by Olufsen (1999). The impact of these conditions on the overall fluid dynamics was tested by means of a comparison with the standard boundary conditions. Results showed that different outflow conditions applied at the same carotid rigid model led to different computed intravascular pressures, which were physiological only when the impedance method was applied. On the contrary, the WSS remained relatively unaffected. This method can be considered an attractive tool to compute physiological intravascular pressures when these are not available and/or considering the difficulties and the limitations to get these data in vivo. Although blood flow can be measured with non-invasive techniques like ultrasound Laser-Doppler, intravascular pressure requires invasive methods which normally alters the measure. Moreover, as example of application of these pressure conditions, we performed an FSI analysis using the same arterial geometry. For FSI analysis, pressure information is in fact necessary to correctly compute the wall stresses and strains and to evaluate how the wall compliance can affect the blood flow pattern and the WSS. For this reason, a comparison between CFD and FSI is performed. Wall stress in fact, as well as WSS, is an important parameter to take into account when analysing initiation and progression of atherogenesis.

2. Modelling and governing equations

2.1 Vascular reconstruction and numerical grid generation

The patient-specific carotid geometry was reconstructed using computed tomography (CT) scans at one time point of the cardiac cycle. As shown in Figure 1, the arterial tree considered in this study included the common carotid (CCA) and its principle branches: the internal carotid (ICA) and the external carotid (ECA). The actual data, including the intraluminal vessel wall, were imported into the commercial software Ansys Icem CFD (Ansys, Inc., Software, Canonsburg, PA, USA). Then, the initial rough geometry was smoothed and the computational fluid mesh was built. The fluid tetrahedral mesh was generated
starting from the internal shell that represents the numerical fluid–solid interface domain. The final mesh was imported into the software package ADINA R&D, Inc. where the FSI and rigid wall simulations were performed. As usual, in order to establish the adequate element size for the computations, a sensitivity study was conducted. Different grids were evaluated to better compute the WSS. Eight different meshes of about 50,000, 120,000, 200,000, 400,000, 600,000, 10^6, 2 x 10^6 and 4 x 10^6 elements, respectively, were tested. For reason of computational time, the comparison was performed only through rigid wall simulations. The instantaneous maximal WSS at the ECA branch was computed at peak flow during the cardiac cycle \((t = 0.15 \text{s})\) within the eight grids in order to guarantee mesh independence of the provided results. In Figure 2, it can be clearly seen that the first four meshes \((5 \times 10^4, 12^5, 2 \times 10^5 \text{ and } 3 \times 10^5)\) have not converged. Increasing the mesh size, even a fully asymptotic value cannot be reached, the maximal WSS reaches the value of 12.61 Pa and stays around it also for the finer grids. The finest grids, in fact, provide the maximal WSS of 12.61 Pa, which is less than 3% higher. The reason why an asymptotic value is not clearly visible is probably due to the type of the considered grid composing the carotid model that is unstructured. Even by increasing the grid size, in fact, it is not possible, with the used commercial software, to build topologically identical meshes that differ only for the number of elements. This means that, due to different mesh topologies, the maximal WSS value also may change its location within the ECA causing small variations in its values even in the very fine mesh. This problem could be avoided with the use of the hexahedral grid, which is unfortunately not suitable for a patient-specific carotid bifurcation. Taking into account that FSI simulations, especially, require in general very high computational costs, the mesh of 600,000 elements was selected. This grid guarantees in fact, on the one side, adequate convergence of the maximum WSS and, on the other side, reasonable computational costs for both FSI and CFD analyses.

Finally, to assess fully developed fluid flow through the carotid, 5-diameter inlet and 15-diameter outlet extensions were added (Joshi et al. 2004). For the solid grid, since no data were available from real images, a constant thickness of 0.6 mm (Alastrué et al. 2010) was given to the arterial wall. A full hexahedral mesh of about 60,000 elements was created using the meshing tool of the commercial software ABAQUS, Inc.

### 2.2 Intravascular flow modelling

As reported in previous works on haemodynamics (Gao and Long 2008; Lee et al. 2008; Gao et al. 2009), the blood rheology was assumed as Newtonian. In particular, according to literature, the blood density was taken as 1067 kg/m³ and a blood viscosity of 0.0035 N/m² was considered (Lee et al. 2008; Gao et al. 2009). Since the Reynolds number based on the average arterial diameter was about \(Re = 950\), the blood flow was assumed laminar and incompressible under unsteady flow conditions.

### 2.3 Material wall properties

The carotid wall was modelled as a hyperelastic incompressible isotropic and homogeneous material. For this preliminary study, the viscoelasticity, the active behaviour of muscle fibres of the artery and the intrinsic anisotropy, due to the preferential directions of collagen and muscle fibres, were not considered. The carotid bifurcation was then characterised by the following strain energy function \(W\):

\[
W = \frac{a}{b} \left[ \exp \left( \frac{b}{2} (\tilde{I}_1 - 3) \right) - 1 \right] + U(J),
\]

where \(\tilde{I}_1\) is the first invariant of the deviatoric right Cauchy–Green tensor \(\tilde{C} = J^{-2} \tilde{F}^T \tilde{F}\), \(J = \det(\tilde{F})\) is the Jacobian, \(\tilde{F}\) is the standard deformation gradient, \(U\) is the volumetric energy function and \(a\) and \(b\) are material constants. In particular, we considered \(a = 44.2\) kPa and \(b = 16.7\) kPa (Delfino et al. 1997; Younis et al. 2004). This strain energy function was initially proposed by Demiray (1972) and, most recently, used by Delfino et al. (1997) for carotid artery. The CT-reconstructed patient-specific carotid geometry was considered as the stress-free configuration. This is clearly an approximation, since it is well known that the geometry from medical images is under a physiological pressure. Moreover, the load-free configuration is different from the stress-free state, as can be observed by the opening angle experiment. Alastrué et al. (2007, 2010) among others have previously studied these phenomena in patient-specific arteries. Finally, in order to bring the initial carotid configuration to

**Figure 2.** Mesh independence study: instantaneous maximum WSS was evaluated for different mesh sizes at peak flow (time \(t = 0.15 \text{s}\)) within the ECA.
physiological pressure values, a ramp pressure was applied to the vessel before the beginning of the cardiac cycle.

### 2.4 Numerical methods

All computational simulations were performed with the finite element-based commercial software ADINA (v8.5, ADINA R&D, Inc.). The details of the numerical approach are given in the literature (Bathe et al. 1999; Bathe and Zhang 2004; ADINA theory and modeling guide 2006). The governing equations are solved with the finite element method (FEM). The FEM equations are obtained by establishing a weak form of the governing equations using the Galerkin procedure (ADINA theory and modeling guide 2006). For the CFD analysis, the continuity equation and the momentum equation are weighted with the virtual quantities of pressure and velocities and then integrated over the computational fluid domain. The fluid domain employs special flow condition-based interpolation (FCBI) available in the commercial software ADINA R&D Inc.

For the FSI model, the fluid domain is deformable so that the numerical approach uses the well-known Arbitrary Lagrange–Euler (ALE) formulation (Donea et al. 1982). For the solid domain, a typical Lagrangian formulation is used (Bathe et al. 1999; Bathe and Zhang 2004; ADINA theory and modeling guide 2006). The fluid and solid domains are coupled through displacement compatibility and traction equilibrium using an equilibrium condition between the stresses acting in the normal direction on both domain boundaries. CFD and FSI simulations were carried out using up to 9000 time steps per cardiac cycle, with a time step size of 0.0001 s, and three cycles were required to damp initial transients. Finally, we used a full Newton method with a maximum of 500 iterations in each time step, and a sparse matrix solver based on Gaussian elimination was used for solving the system.

### 2.5 Physiological boundary conditions

Flow rate entering the CCA was not available for these subjects, so the inlet flow rate waveform was taken from literature. In particular, we used the averaged CCA waveform described by Lee et al. (2008) to define for an average case, followed by the scaling of this average waveform by the ratio of the CCA cross-sectional area. Starting from these flow waveforms, we adopted the impedance recursive method developed by Olufsen (1999) and extended by Steele et al. (2007) for calculating the impedance of the cardiovascular system, we first modelled the circulatory network that was not reconstructed through CT imaging as a structured fractal tree in which we predicted the physiological blood flow and the pressure profiles. This model consisted of two parts: the large arteries (in the case of this study, the carotid artery) and the small arteries, in which a relationship between flow and pressure represents the outflow boundary conditions for the large arteries (Olufsen et al. 2000). In both large and small arteries, the blood flow and pressure are calculated using the incompressible axisymmetric Navier–Stokes equations for a Newtonian fluid. A 1D model is obtained integrating these equations over the cross-sectional area of each vessel (Olufsen 1999; Olufsen et al. 2000). The impedance method is used prior to each FSI computation. Based on the radius of each branch, first the fractal network is created (a radii matrix, which will be the input for the impedance recursive computation). Once the radii are determined for each daughter vessel (two separate trees for the ICA and ECA are created since the radii of each branch are different), the impedance computation allows the computation of the pressure waveforms shown in Figure 3 using the 1D Navier–Stokes equations. The small arteries were modelled as a binary asymmetric-structured tree, in which each vessel was approximated by using a 1D flow dynamics approach. In particular, pressure waveforms on the ICA and ECA of the carotid bifurcation will be computed as outlet conditions. The used inflow and pressure waveforms are shown in Figure 3 (top and bottom right). At the inlet extension, the CCA-adapted velocity was imposed, whereas at the outlet extensions, the computed ICA and ECA pressure waveforms were applied. The resulting boundary conditions were finally applied in the commercial software to the already mentioned branch extensions in order to perform the simulations. Moreover, no-slip wall boundary condition was imposed at the arterial wall, which for the FSI simulation represented also the fluid–structure interface. Finally, at the extension extremities of the solid model, no axial or transaxial motion was permitted. This constraint, though non-physiological, as the inlet and outlets should be allowed to deform radially to correctly simulate tethering of the artery, is commonly assumed in numerical studies of carotid (Gao and Long 2008; Tang et al. 2008; Gao et al. 2009), coronaries (Torii et al. 2006, 2010) and aneurysms (Molony et al. 2009; Torii et al. 2009; Takizawa et al. 2010).

### 2.6 The impedance-based method

#### 2.6.1 The fractal network

Following the recursive approach developed by Olufsen (1999) and Olufsen et al. (2000) and extended by Steele et al. (2007) for calculating the impedance of the cardiovascular system, we first modelled the circulatory network that was not reconstructed through CT imaging as a structured fractal tree in which we predicted the physiological blood flow and the pressure profiles. This model consisted of two parts: the large arteries (in the case of this study, the carotid artery) and the small arteries, in which a relationship between flow and pressure represents the outflow boundary conditions for the large arteries (Olufsen et al. 2000). In both large and small arteries, the blood flow and pressure are calculated using the incompressible axisymmetric Navier–Stokes equations for a Newtonian fluid. A 1D model is obtained integrating these equations over the cross-sectional area of each vessel (Olufsen 1999; Olufsen et al. 2000). The impedance method is used prior to each FSI computation. Based on the radius of each branch, first the fractal network is created (a radii matrix, which will be the input for the impedance recursive computation). Once the radii are determined for each daughter vessel (two separate trees for the ICA and ECA are created since the radii of each branch are different), the impedance computation allows the computation of the pressure waveforms shown in Figure 3 using the 1D Navier–Stokes equations. The small arteries were modelled as a binary asymmetric-structured tree, in which each vessel was approximated by
A straight compliant segment. Relevant parameters are radii, bifurcation relationships, asymmetry and area ratios, length and compliance. The vascular network resulted in a series of bifurcations composed of a series of parent and daughter vessels, as shown on the left side of Figure 3. The structured tree is constructed such that it is geometrically self-similar (Olufsen 1999). All parameters are specified as function of the vessel radius. A power law determines how the radius changes across each bifurcation:

\[ r_{j+1} = r_j^{\alpha} + r_j^{\beta}, \quad 0 \leq j \leq i, \]  

where \( r_{0i} \) and \( r_{0i2} \) are the radii of the daughter vessels and \( r_{PA} \) is the radius of the parent vessel while \( \xi \) is the power exponent. Equation (2) is well known in the literature (Murray 1926; Zamir 2002) and it is derived by the principle of minimum works in the arterial system. In particular, \( \xi = 3 \) is optimal for laminar flow while \( \xi = 2.33 \) for turbulent flow. More in general, the value of \( \xi \) varies between 2 and 3. In arterial blood flow, a good choice for the power exponent, is \( \xi = 2.76 \) (Olufsen 1999). Each parent vessel bifurcates into two daughter vessels following a scaling guided by the asymmetry factors \( \alpha \) and \( \beta \), according to Equation (3):

\[ r_{i,j} = \alpha^{i} \beta^{j-i} r_{root}, \quad 0 \leq j \leq i, \]  

\[ (r_0)_{d_i} = \alpha (r_0)_{pa}, \quad (r_0)_{d_2} = \beta (r_0)_{pa}. \]

Assuming the radii of the arterial branches ICA and ECA as the roots of the tree, we created two asymmetric vascular networks, as shown on the left side of Figure 3. The structured tree continues branching until the radius of any vessel is less than the given minimum values \( r_{min} \). Normally, a capillar radius, where the pressure was set to 0 mmHg (Olufsen 1999; Olufsen et al. 2000; Steele et al. 2007). Asymmetry and area ratios of the vessels were first defined as

\[ \eta = \frac{(r_0)^2_{d_2} + (r_0)^2_{d_1}}{(r_0)^2_{pa}} \]  

and

\[ \gamma = \left( \frac{r_{0i2}}{r_{0i1}} \right)^2, \]

where \( \eta \), the area ratio, and \( \gamma \), the asymmetry ratio are related to each other as shown in the following equation

\[ \eta = 2 \eta_0 \gamma^n \]

\[ \gamma = \frac{1}{2} \left( 1 + \frac{\eta}{\eta_0} \right)^{1/n} - \frac{1}{2} \]

\[ n = 0.5 \eta_0 \gamma \]

\[ \eta_0 = \left( \frac{r_{0i2}}{r_{0i1}} \right)^{2/n} \]

\[ \gamma = \frac{1}{2} \left( 1 + \frac{\eta}{\eta_0} \right)^{1/n} - \frac{1}{2} \]

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\[ n = 0.5 \eta_0 \gamma \]
Table 1. Parameters used to describe the structured tree.

<table>
<thead>
<tr>
<th>Radius (µm)</th>
<th>α</th>
<th>β</th>
<th>ξ</th>
<th>γ</th>
<th>η</th>
</tr>
</thead>
<tbody>
<tr>
<td>300 &lt; r &lt; 50</td>
<td>0.9</td>
<td>0.6</td>
<td>2.76</td>
<td>0.41</td>
<td>1.16</td>
</tr>
</tbody>
</table>

Note: As in Olufsen et al. (2000), α and β were kept constant, scaling the tree.

(Olufsen 1999):

\[
\eta = \frac{1 + \gamma}{(1 + \gamma^{\xi/2})^{2/\xi}}. \tag{7}
\]

η is also known as asymmetry index and describes the relative relationship between the daughter vessels. Assuming that \( r_{d1} < r_{d2} \), η is between 0 and 1. The value of the asymmetry index widely varies throughout vascular beds and does not appear to be organ specific (Zamir 2002). If the area ratio η and the power exponent ξ are known, γ can be calculated. The value of η in Equation (7) is extensively discussed in the literature. It describes the branching relationship across bifurcations between the radius of the parent vessel and the radii of the daughter vessels at bifurcation points. Using an area ratio of η = 1.16 and the power ξ = 2.76, the asymmetry ratio γ is 0.41. The scaling parameters are finally obtained from the following expressions:

\[
\alpha = (1 + \gamma^{\xi/2})^{-1/\xi}, \quad \beta = \alpha \sqrt[\xi]{\gamma}. \tag{8}
\]

In Table 1, the parameters used to scale the fractal tree are summarised.

The length of each vessel \( L \) is related to the radius \( r_0 \) using a constant length-to-radius ratio \( \ell_r = L/r_0 \). This parameter is well known in the literature (Olufsen et al. 2000; Steele et al. 2007), where different values can be found. Iberall (1967) estimated that the length-to-radius \( \ell_r = L/r_0 = 50 \pm 10 \) for small arteries. Based on these studies, Olufsen et al. (2000) used the value \( \ell_r = 50 \) while Steele et al. (2007) used a multi-level approach. Zamir (2002) suggested that the mean \( \ell_r \) is 20 with a maximum of 70. Other studies showed that this parameter is widely varied in the vascular tree being also organ specific. In this study, we fixed \( \ell_r = 50 \) as suggested in previous works (Iberall 1967; Olufsen 1999; Olufsen et al. 2000), the value that is based on branches from the small arteries such as the renal arteries, the femoral arteries and the cerebral artery (Iberall 1967). For further explanations on the fractal tree, see Olufsen (1999), Olufsen et al. (2000) and Steele et al. (2007).

### 2.6.2 The recursive impedance computation

Outflow conditions representing the impedance due to the peripheral arteries are applied as pressure conditions at the boundaries. The time-dependent pressure, which is applied at the outflow boundary, is obtained as follows:

\[
P(t) = \int_{t-\tau}^{t} Q(t) Z_{tot}(t - \tau) \, dt, \tag{9}
\]

where \( Q(t) \) and \( Z_{tot} \) are the time-dependent outflow and the total impedance arising from downstream of the outlet. Schematics of the combination of 3D and 1D modelling are shown in Figure 3. Here the impedance of the peripheral arterial network is modelled analogous to an electric circuit. The impedance at the root of the vascular tree is calculated in a recursive manner starting from the terminal branch (Olufsen et al. 2000; Steele et al. 2007). As already mentioned, along each vessel in the structured tree, impedance is computed from the linear, axisymmetric, 1D equations of Navier–Stokes. The details of the recursive calculation are given elsewhere (Olufsen et al. 2000). Following the same scheme, we evaluated the input impedance at the beginning of each vessel \( z = 0 \) as a function of the impedance at the end of a vessel \( z = L \):

\[
Z(0, \omega) = \frac{ig^{-1} \sin(\omega L/c) + Z(L, \omega) \cos(\omega L/c)}{\cos(\omega L/c) + igZ(L, \omega) \sin(\omega L/c)} \tag{10}
\]

where \( L \) is the vessel length, \( c = \sqrt{s_0(1 - F_1)/(\rho_F C)} \) is the wave propagation velocity and

\[
F_1 = \frac{2J_1(w_0)}{w_0J_0(w_0)},
\]

being \( J_0(x) \) and \( J_1(x) \) the zero-order and first-order Bessel functions with \( w_0 = i^3 w, w^2 = \rho_F r_0^2 \omega/\mu \), \( i \) the complex unity and \( w \) the Womersley number. The compliance \( C \) is approximately given by

\[
C = \frac{3A_0 r_0}{2 Eh} \quad \frac{Eh}{r_0} = k_1 \exp(k_2 r_0) + k_3 \tag{12}
\]

where \( k_1, k_2 \) and \( k_3 \) are known constants obtained by Olufsen (1999), \( s_0 \) the cross sectional area and \( r_0 \) the root vessel; \( E \) is the Young modulus which is assumed as function of the vessel thickness \( h \) and the root radius \( r_0 \) (Olufsen 1999; Steele et al. 2007), and is given by Equation (12).

The computation of the pressure waveforms is performed through the software Matlab (MathWorks®) prior to the numerical analyses. For each time step, a spatial uniform pressure value is given to each outlet, at the bottom of the model extensions, in order to guarantee parabolic and streamwise-independent profiles. This value varies during the cardiac cycle following the curves.
represented in Figure 3(b) for the ICA and ECA, respectively.

3. Results and discussion

3.1 Role of the boundary conditions: intravascular pressure and flow comparison

The importance of using impedance-based pressure conditions is assessed first comparing different outflow conditions using the same geometry and the same approach (CFD). In particular, standard traction-free and flow splitting through the bifurcation (40% to the ECA and 60% to the ICA; Thomas et al. 2002, 2005; Lee et al. 2008) are compared with the results obtained using impedance-based conditions. The most notable change between standard traction-free, flow splitting and impedance outflow conditions is the pressure distribution, both spatially and temporally. In Figure 4, the temporal pressure variation is compared for these three different boundary conditions. Evidently, there are marked differences. In particular, even similar spatial distribution was found, only the impedance-based conditions provide physiological intravascular pressure. Moreover, with the impedance conditions, the maximum pressure is approximately 40% higher than that obtained using stress-free conditions at the outlets of the model. Moreover, the pressure history revealed by imposing flow conditions is approximately constant along the cardiac cycle with a maximal difference of 40% with respect to that of the impedance conditions. The intravascular pressures of Figure 4 are compared at an intermediate point on a bifurcation cutting plane. In addition, the registered pressure variation between systole and diastole is about 30% higher when the impedance-based conditions are applied. Imposing the flow at the model inlet and outlets causes an almost constant pressure along the cardiac cycle. Even the average value along the cycle is physiologic (≈100mmHg), and the pressure gradient is very small in comparison with that found in other works (Perktold and Rappitsch 1995; Younis et al. 2003, 2004) and not in physiological range (≈40–60mmHg). This gradient, on the contrary, is well caught using impedance-based pressure conditions.

The time history of the pressure is completely different for the three scenarios. From a physiological perspective, this finding has a high relevance because the correct evaluation of the intravascular pressure plays an important role in the cardiovascular mechanics, especially in diseased patients. Moreover, intravascular pressure is crucial for evaluating arterial stress distributions which, as reported by Delfino et al. (1997), Younis et al. (2004) and Thubrikar (2007) among others, is recognised to play an important role in combination with the WSS distributions for initiation, development and progression of atherosclerosis. On the contrary, the intravascular blood flow looks relatively unaffected by the different outflow conditions. The simulation results show in fact the well-known characteristics for flow in an arterial bifurcation; in particular, due to curving vessels flow phenomena like flow recirculations appear. The latter arises from the need to satisfy the continuity condition. Furthermore, the tapered vessel geometry causes acceleration of the fluid after the bifurcation, and the flow is relatively similar when comparing the streamlines of results obtained through impedance conditions with the aforementioned standard outflow conditions, which means that the application of the impedance boundary conditions in these locations is acceptable.

3.2 Vessel haemodynamics

As an example for the application of the provided boundary condition, we performed an FSI analysis using the same intraluminal arterial geometry. FSI computations in haemodynamics in fact need pressure information in

Figure 4. Comparison of the temporal pressure history (in mmHg) at a specified location for rigid wall by applying impedance-based pressure conditions, simply outflow stress-free conditions and imposing the flow at the outlets.
order to correctly compute stresses, strains and lumen compliance. These variables are very important in atherogenesis as reported in the literature (Ku et al. 1985; Milner et al. 1998; Thubrikar 2007). The obtained results were compared with the CFD in order to examine the potential effect of the wall compliance on the haemodynamic variables. Although not necessary for CFD analysis, intravascular pressure is a crucial information for the FSI technique. The following comparison shows the capability of the impedance method to generate adequate and useful conditions for fluid–structure coupled approach. The latter, in the present work, has to be considered as a generic application and anyway as a first step. Further investigations and a more adequate and detailed model will be provided in future works.

The WSS field in the carotid bifurcation is determined by the flow field. The blood flow near the bifurcation separates from the arterial wall generating a recirculation area that causes low WSS regions at the proximal section of the bifurcation, inside both the ICA and ECA branches and at the carotid bulb. The flow separation at the carotid bulb and inside the branches is expected due to the abrupt change in curvature and the associated adverse pressure gradient along the outer wall of the artery as found also in other works (Younis et al. 2003, 2004). High momentum fluid is incapable of instantaneously changing the direction to accommodate the curvature. The found WSS distributions are similar for both CFD and FSI simulations. This distribution which is due to the velocity field inside the carotid matches well with the other results found in the previous works by Motomiya and Karino (1984), Moyle et al. (2006), Lee et al. (2008) and Wake et al. (2009), among others. Moreover, the obtained range of WSS values are confirmed by other works (Younis et al. 2003, 2004). Reverse flow, and low or oscillating WSS correlate with atherosclerosis development in the carotid artery bifurcation as documented in Bharadvaj et al. (1982) and Ku et al. (1985). In Figure 5, a comparison between the temporal minimal shear stress (in each arterial branch corresponding to the regions 1, 2 and 3 in this figure, bottom-left), respectively, for rigid wall and FSI simulations is shown. As expected, the computed curves show the same trend as the inlet velocity condition (see Figure 3), but no negligible differences are found in the WSS values. The post-processed WSS temporal variation shows higher values if computed with CFD simulation rather than the FSI analysis. In the ECA (see Figure 5-3), we can observe a constant gap between the minimal WSS values computed with FSI and with a rigid wall. The maximal gap between both curves is reached at time \( t = 0.38 \) s (see Figure 5-3), where the value computed through rigid wall analysis is approximately twice those calculated with FSI. For the ICA, the maximal gap between the two curves of Figure 5-2 is registered at time \( t = 0.25 \) s, where the value computed with a rigid wall simulation is almost 10 times those computed with FSI.

Figure 5. Comparison of the minimum WSS value of each arterial branch as a function of the time between rigid wall and FSI computations.
Smaller gaps are registered at the beginning of the cardiac cycle, for time $0 \, t \, 0.1 \, s$. Finally, in the CCA (see Figure 5-1), the gap between the aforementioned curves is reduced. While a maximal gap is registered at time $t = 0.1 \, s$, where the minimal WSS computed with a rigid wall is approximately twice of those computed with FSI, both curves are quite near at the beginning and at the end of the cardiac cycle. From these results, we can conclude that the arterial compliance strongly affects the WSS evaluation, especially in the areas characterised by high displacement. The compliance of the arterial walls dilates the vessel, and the WSS is consequently altered as observed in other works (Perktold and Rappitsch 1995; Younis et al. 2004).

Since low and oscillatory WSS plays an important role in atherogenesis, as in other studies (Ku et al. 1985; Younis et al. 2004), the time-averaged WSS (TAWSS) was considered. This parameter was compared between the FSI and rigid wall models to assess the importance of FSI in determining carotid haemodynamics. The TAWSS is defined as

$$\text{TAWSS} = \frac{1}{T} \int_{0}^{\infty} |\tau_{W}| \, dt,$$  

where $\tau_{W}$ is the WSS vector and $T$ is the period of the cardiac cycle. As shown in Figure 6, the TAWSS, displayed in logarithmic scale (Younis et al. 2004), is low at the aforementioned locations 1, 2 and 3 of Figure 5 due to the flow divider which causes recirculation flow and consequently low velocity gradient near the wall. In these three regions, the values computed through FSI are lower than those found with CFD analysis as shown in Figure 6.

In particular, the impact of the distensible walls in the TAWSS evaluation is visible in the region 2 that registers the lowest value ($\leq 1 \, \text{Pa}$). Moreover, the dilatation of the vessel due to the pulsating flow induced a strong reduction of the blood velocity causing consequently accentuated lower WSS values along the entire ICA (see Figure 6) for the FSI model with respect to the CFD.

3.3 Arterial stress and wall compliance

In Figure 7, the temporal variation of three sections before and after the bifurcation (a median section on CCA (section 1), ICA (section 2) and ECA (section 3) is shown). Maximal diameter variations during a cardiac cycle are 0.09 mm, 0.29 mm and 0.79 mm on CCA, ICA and ECA, respectively. By normalising the changes with their corresponding diameter, we obtained 2.1%, 1.1% and 2.8%, respectively. It has to be noted that these values refer to the diameter variation during the cardiac cycle. The maximal diameter variation of each branch from the initial configuration to the beginning of the cycle was of about 18.5%, 20.4% and 16.3%, respectively. Our results agree with those obtained in other works (Perktold and Rappitsch 1995), which shows that the arterial wall compliance for carotid remains small during the cardiac cycle. Primarily the WSS magnitude is reduced in the distensible model. The results of the flow study have been compared as far as possible with other numerical results (Perktold et al. 1991; Perktold and Rappitsch 1995). The comparison can be performed only in a qualitative manner because the conditions (pressure pulse wave, wall

![Figure 6. Comparison of TAWSS (with the logarithmic scale in Pa) between the FSI (top) and the rigid-wall models (bottom).](image-url)
parameter, Reynolds number and rheological blood property) are different. The qualitative comparison shows agreement in the essential features. During diastole, a reduction of the internal axial flow velocity is registered so that the pressure and, thus, the vessel lumen increase as reported also by Perktold and Rappitsch (1995). During systolic acceleration, the pressure decreases, the vessel lumen contracts and the internal axial flow velocity increases. In Figure 8, the maximum principal stress at peak maximal and minimal flow is represented in a median section of the carotid bifurcation. The maximal stress (around 320 kPa) is located at the side-wall region of the intersection of the two branches, as reported in other works (Perktold et al. 1991; Perktold and Rappitsch 1995). The computed values are higher than those found by Perktold and Rappitsch (1995) probably due to the different used boundary conditions (see Figure 8 top) and a cause of the different used material properties. Moreover, these values are lower than those found by Delfino et al. (1997), even they used a pure solid mechanics approach. The carotid bulb is one of the first sites in the carotid bifurcation to develop late stage atherosclerotic inflammation as reported by Ku et al. (1985), Delfino et al. (1997) and Younis et al. (2004). The proposed correlation between the low WSS and the maximal arterial wall stress (approximately 180 kPa) exhibits in fact a low WSS distribution (see Figures 6 and 8). This coupled effect is recognised to be subjected to high risk of atherosclerosis initiation and development as reported by Thubrikar (2007). Smooth muscle proliferation is a key element in the development of the lesion. Injured cells induce a chronic inflammatory response followed by a healing or fibroproliferative response (Thubrikar 2007). This injury can be caused by CS, especially in the carotid artery joined to low WSS values which play nevertheless an important role in this process.

3.4 Study limitations

This work, even showing improvements with respect to other previous studies, is affected by some assumptions and simplifications. First of all, the impedance computation was applied following the method developed by Olufsen (1999) for a tapering generic big artery model. Specific parameters used for the carotid tree scaling and for the impedance evaluation were not available in literature, so some additional assumptions on the arterial geometry were made. The impedance-based pressure waveforms were computed starting from experimental velocity waveforms taken from the literature (Lee et al. 2008). In future works, inflow waveforms belonging to the same patient of the carotid model should be provided for the impedance computation. Moreover, we applied all velocity waveforms with a time variation of a flat profile.
applied at the end of the aforementioned branch extensions. The blood flow velocity profile entering in the carotid artery is probably more complicated. Finally, a Newtonian blood rheological model was assumed. Even the non-Newtonian nature of the flow may not have significant impact on the considered model, and real blood rheology will be included in future works. In absence of real data, a constant arterial wall thickness was assumed. Even in this way, the solid model was strongly approximated, in future work, in order to evaluate wall compliance; the carotid wall should be modelled starting from patient-specific data. This work has to be considered the first step to study the carotid hemodynamics with useful boundary conditions for including in the future the arterial compliance.

4. Conclusions

The proposed impedance method can be considered a very attractive tool to compute physiological pressures starting from blood flow measures. In fact, while intravascular flow can be obtained with standard techniques through non-invasive measurements (as ultrasound laser Doppler for example), intravascular pressures required the use of invasive probes which can affect the overall blood flow during the measure. In addition, laser Doppler is considered a standard technique in the daily medicine practice while pressure measurements are performed only in special cases due to its difficulty.

Impedance-based pressure waveforms were computed and applied to a patient-specific human carotid bifurcation. Comparison with standard outflow conditions demonstrated clear differences between the numerical results of the computed pressures. On the contrary, the intravascular blood flow is relatively unaffected by these conditions. Moreover, while standard boundary conditions provide almost the same WSS distributions in comparison with those obtained with impedance-based conditions, these cannot be used to compute the wall stresses and strains which are important factors to take into account in atherogenesis. To correctly evaluate the arterial wall behaviour, time-dependging pressure conditions are in fact necessary. Although not necessary for CFD simulation for evaluating the WSS, intravascular pressure must be correctly computed for compliant vessels. For this reason, as application example, an FSI analysis of the carotid artery was carried out applying the impedance method as boundary conditions, in order to investigate the effect of the arterial wall compliance on the overall carotid haemodynamics. WSS distributions were then compared between FSI and CFD simulations. The impedance method, through physiological boundary conditions, led to physiological intravascular velocities and pressures that provided wall stresses and WSS spatial distributions comparable with previous studies. The FSI numerical computation showed lumen compliance values in agreement with previous studies. Comparison of computational results between rigid and compliant arterial walls showed

Figure 8. Maximum principal stress (in kPa) of the carotid artery bifurcation at time \( t = 0.15 \) s (a) and \( t = 0.8 \) s (b).
qualitative discrepancies in the WSS distribution, but also relevant differences in the WSS temporal profiles confirming the feasibility of these types of outflow conditions and their utility especially where, as in the case of FSI analysis, pressure conditions are necessary.

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