SENSITIVITY OF HEMODYNAMIC PARAMETERS TO WAVEFORM, FLOW DIVISION, AND HEAD ROTATION IN THE HUMAN CAROTID BIFURCATION

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BACKGROUND
Hemodynamic parameters such as time averaged wall shear stress (TAWSS), wall shear stress temporal gradient (WSSTG) and Oscillatory Shear Index (OSI) have previously been cited as parameters associated with the development of atherosclerotic disease at the human carotid bifurcation [1,2]. The sensitivity of these important parameters however, with variations of driving flow waveform, flow division and posture changes are not well known. To investigate these changes, we have used image based CFD, to analyze the flowfield of the carotid bifurcation of a healthy volunteer for five different input waveforms, three flow division ratios and two head postures.

MATERIALS AND METHODS
The right carotid artery of a healthy volunteer was scanned using a 3.0 T Philips Achieva MRI instrument at the neutral head position and at a rightward head rotation (45 degrees). The employed 3D gradient-echo pulse sequence (TE=3.5ms, TR=23ms, φ=20°, acquisition voxel of 0.36x0.36x1.2mm³) provided 100 overlapping (by 50%) slices with a reconstruction voxel of 0.2x0.2x0.6 mm³. The inflow and outflow boundary conditions were obtained by MR phase contrast velocity mapping at the limits of the TOF covered anatomic region. The sequence used was a 2D, gradient-echo sequence with a TE of 12 ms, a TR of 19 ms, a flip angle of 10°, an acquisition voxel of 0.69x0.69x5.0 mm³ (reconstruction voxel of 0.35x0.35x5.0 mm³) and a velocity window of ± 65 cm/sec. Peripheral pulse triggering was implemented and 30 temporal phases per RR cycle were obtained. Magnitude and phase-difference velocity-encoded images were derived using a view sharing reconstruction.

Segmentation and 3D surface reconstruction of the MR images were implemented using purpose-developed software (3). From the segmented MR images the 3D true vessel lumen surface was reconstructed. Abnormal, small scale surface irregularities introduced during the imaging and reconstruction processes applied were excluded from the computational model by applying pixel width constrained smoothing of the reconstructed surfaces prior to mesh generation. Finally, smoothly matched cylindrical extensions of both inflow and outflow segments were added to facilitate the application of fully developed boundary conditions for the numerical simulation of the flow field. To investigate the complex flow dynamics of the carotid bifurcation a numerical approximation of the Navier-Stokes equations solution was computed using Fluent v6.2 (Ansys Inc.). The computational grid was generated with ANSA (Beta CAE Systems, Greece). It was constructed using 4.5 10⁵ tetrahedral / pentahedral elements and non-uniform grid node spacing with higher grid density in the vicinity of the bifurcation. The boundary conditions were varied in terms of input waveform, flow division and geometry due to posture change.

Inlet Waveforms: Five inlet flow waveforms were used: a) The CCA waveform (13 harmonics) at resting conditions (T=1.032s), b) The CCA waveform at exercise conditions (T=0.516s) and c) the first harmonic d) the first 2 harmonics and e) the first 3 harmonics of the physiological waveform as shown in Fig.1. The CCA inflow waveform was obtained by in vivo MR velocimetry. The waveforms were chosen based on their frequency content to investigate how it influences the flow field.

Flow Division: Three different flow divisions through the internal carotid artery (ICA) and external carotid artery (ECA) were used: a) 60/40 b) 65/35 and c) 70/30 (ICA/ECA).

Head Posture: Two different head postures were used: a) the neutral position and b) rotation of the head 45 degrees clockwise. This latter
posture was previously shown to cause geometric morphology changes to the artery.

Based on the discrete Fourier series of the measured waveform, the fully developed Womersley solution was prescribed at the model inlet (Re=340, α=4.1) for all studied flow waveforms except for the exercise waveform where α=5.79. From the simulation results, the OSI was computed as:

\[ OSI = \frac{\int_{0}^{T} \left( \mathbf{\tau} \cdot \mathbf{n}_m \right) dt}{\int_{0}^{T} \left| \mathbf{\tau} \cdot \mathbf{n}_m \right| dt} \]

\[ \mathbf{n}_m = \frac{1}{T} \int_{0}^{T} \left( \mathbf{\tau} \right) dt \]

where \( w=0.5(1-\cos a) \) and \( a \) the angle between the shear vector \( \mathbf{\tau} \) and the mean shear direction \( \mathbf{n}_m \). OSI varies between 0 for unidirectional shear flow and 0.5 for the purely oscillatory shear case. The WSSTG was also computed at each time step using a second order central difference scheme.

RESULTS
Fig. 2abcde shows the TAWSS magnitude distribution for the neutral posture 65/35 ICA/ECA flow division with sinusoidal waveform (a), 2-harmonics (b) 3-harmonics (c) and physiological rest waveform (d) and physiological exercise waveform (e). Fig. 2fg shows the same characteristics for the 70/30 flow division (f), and 60/40 flow division (g). Fig.2h shows the TAWSS distribution for the right flexion head posture for a 65/35 flow division and a physiologic rest waveform. To identify regions of low and oscillatory shear we consider a non-dimensional parameter (nOSI) which is calculated by dividing the OSI by the TAWSS magnitude normalized by the mean Poiseuille flow WSS (Re=340). The OSI and nOSI distribution over the surface is shown in Fig. 3 for the same conditions as in Fig. 2abcde and fgh. The corresponding changes in time averaged WSSTG are shown in Fig. 4abcde and Fig. 4fgh. Table 1 shows the total wall surface area divided by inlet cross-sectional area exposed to low or highly oscillatory shear stress (TAWSS values below 0.4 Pa and OSI values above 0.3).

| Table 1. Total wall area divided by inlet cross-sectional area exposed to low or highly oscillatory shear stress. |
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| | a | b | c | d | e | f | g |
| TAWSS<0.4 Pa | 2.7 | 2.5 | 2.3 | 1.9 | 1.0 | 1.1 | 2.5 | 1.2 |
| OSI>0.3 | 0.1 | 0.1 | 0.2 | 0.4 | 0.6 | 0.1 | 0.4 | 0.1 |

DISCUSSION
The results show that the sinusoidal waveform had augmented the regions of low TAWSS, while the “exercise waveform” augmented the regions of high OSI and time averaged WSSTG in comparison to the sinusoidal and resting conditions waveform. The addition of higher harmonics to the waveform reduces the extent of low TAWSS but augments regions of high OSI. The effect of flow division showed that as more flow goes through the ECA, the regions of low TAWSS, high OSI and high WSSTG also increase. When the morphology changes due to head rotation, it seems that all the regions of low TAWSS, high OSI and high WSSTG are reduced in comparison to the neutral position. The results presented in Table 1 provide a measure of the influence of the various parameters investigated on the extent of wall exposure to hemodynamic conditions that have been found to promote vascular disease.

REFERENCES

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