Influence of head position on carotid hemodynamics in young adults

F. P. Glor, B. Ariff, A. D. Hughes, P. R. Verdonck, D. C. Barratt, S. A. M. Thom, and X. Yun Xu. Influence of head position on carotid hemodynamics in young adults. Am J Physiol Heart Circ Physiol 287: H1670–H1681, 2004; 10.1152/ajpheart.01186.2003.—Studies in adults have shown marked changes in geometry and relative positions of the carotid arteries when rotating the head. The aim of this study was to quantify the change in geometry and analyze its effect on carotid hemodynamics as a result of head rotation. The right carotid arteries of nine young adult subjects were investigated in supine position with straight and left turned head positions, respectively. The three-dimensional (3D) carotid geometry was reconstructed by using 3D ultrasound (3D US), and the carotid hemodynamics were calculated by combining 3D US with computational fluid dynamics. It was observed that cross-sectional areas and shapes did not change markedly with head rotation, but carotid vessel center lines altered with planarification of the common carotid artery as a main feature (P < 0.05). Measured common carotid flow rates changed significantly at the individual level when the head was turned, but on the average, the change in mean common carotid flow rate was relatively small (0.37 ± 1.11 ml/s). The effect of the altered center lines and flow rates on the atherogenic nature of the carotid bifurcation was evaluated by using calculated hemodynamic wall parameters, such as wall shear stress (WSS) and oscillatory shear index (OSI). It was found that WSS and OSI patterns changed significantly with head rotation, but the variations were very subject dependent and could not have been predicted without assessing the altered geometry and flow of the carotid bifurcation for individual cases. This study suggests that there is a need for standardization of the choice of head position in the 3D US scan protocol, and that carotid stents and emboli diverters should be studied in different head positions.

EFFECTS OF BODY POSTURE and head positioning on respiration and cerebral hemodynamics have been investigated in a number of studies. Effects on respiration seem to suggest that body posture rather than head position has the greatest effects (6). In infants, sleeping in a prone position increases pharyngeal resistance, which may increase inspiratory effort (15). If the head is turned, however, increases in heart and respiratory rates, decreases in oxygen saturation, and marked pallor compared with infants sleeping in a supine position (33). Three changes, however, do not occur with changes in head positions alone (11). In adults, changes in head position also do not seem to change oxygenation or maximum inspiratory pressure. Adopting a supine position, however, increases pharyngeal resistance (38) and has been shown to worsen apneic episodes in those suffering from obstructive sleep apnea (6).

The hemodynamic effects of body posture and head position have also been examined. In infants, whereas cerebral blood flow rate remains constant during head rotations in the supine position, keeping the head straight favors cerebral venous drainage and helps to prevent an elevation of cerebral blood volume (29). Eichler et al. (12) showed that in infants, blood velocity profiles changed in the basilar, vertebral, and internal carotid arteries (ICA) with head rotations. Kinking of the vertebral artery at the base of the skull with head rotation may explain the change in flow in the vertebral and the basilar artery (19). The altered velocity profile in the ICA was not accompanied by a change in flow rate.

Studies (34) in adults have shown marked changes in geometry and the relative positions of the carotid arteries and the internal jugular vein when rotating the head. In addition, rotation of the head is accompanied by significant changes in blood flow patterns and blood velocity profiles in the vertebral and basilar arteries (25, 31) as well as in the ICA (30), although total cerebral blood flow is kept constant (30). The blood flow patterns in the ICA and vertebral arteries also differ from left to right, e.g., the right ICA shows increased blood flow rates with rotation to the contralateral position, but the left ICA did not show the same trend (30).

With the knowledge of potential changes to the carotid geometry and hemodynamics on the one hand, and the potential link between blood flow patterns and atherosclerosis on the other hand (9, 41), the question of how important the effect of the potential hemodynamic changes is on the risk for atherosclerosis is raised. Therefore, the aim of this study was to investigate the effect of the change in geometry on carotid hemodynamics as a result of head rotation. Hemodynamic wall parameters, such as wall shear stress (WSS) and oscillatory shear index (OSI), were evaluated, because these have been linked to different stages of the atherosclerotic process. To our knowledge, this is the first study to examine the three-dimensional (3D) carotid blood flow patterns in different head positions using in vivo, subject-specific data. The hypothesis is that turning the head has an important effect on both the carotid geometry [as suggested by Sulek et al. (34)] and the carotid hemodynamics. This would imply that the risk for atherosclerosis to develop in the carotid arteries, which is associated with low WSS and high OSI, may vary with head positions. The altered distribution of hemodynamic forces could also be the...
cause of so-called “telephone strokes” (28), i.e., strokes occurring at the occasion of prolonged telephone calls during which the patient holds the phone with shoulder and ear.

METHODS

Subjects and Head Positions

The right carotid arteries of nine healthy subjects, one man and eight women (mean age 26.6 ± 3.0 yr), were investigated with ultrasound (US) by a single operator. All of the subjects were normotensive with no history of vascular diseases. All participants gave written consent and their participation was approved by the institutional review board of their respective participation institutions. They were scanned twice serially in a supine position. In the test. In the second scan, the subject was asked to turn his/her head far to the left and to fix on a point on the wall.

The two scans will be referred to as a straight or a turned scan. In both scans, the subject’s head was tilted forward slightly from the supine position because of the cushion. The head tilt was the same for every subject.

3D US and Computational Fluid Dynamics

The goal of this study was to evaluate the differences in geometry and hemodynamic parameters at the right carotid bifurcation between two head positions. Augst et al. (2) have shown that it is feasible to calculate the flow in carotid arteries on the basis of US images of the arterial geometry. This technique combines carotid US imaging from which the 3D carotid geometry is reconstructed and computational fluid dynamics (CFD) for 3D flow pattern reconstruction. Briefly, the 3D, time-varying, flow profiles were obtained by performing the following steps.

US imaging and slice positioning information. A custom-built, freehand 3D US system was used to acquire 3D images of the carotid bifurcation (3). The system employs a commercial US scanner (model HDI-5000; Philips Medical Systems), equipped with a conventional 5 to 12 MHz broadband linear array transducer, which is tracked in space using an electromagnetic position and orientation measurement (EPOM) device (Ascension Technology) mounted on the US probe. With the use of the system, a series of two-dimensional (2D), ECG-gated transverse images of the carotid bifurcation were acquired for each subject. Images were acquired after a subject-specific delay relative to the ECG R wave (typically ~400 ms and roughly coincident with the end of the T wave). The EPOM device was used to simultaneously record the 3D position and orientation of each slice. The total lengths of artery scanned varied from 2 to 7 cm proximal to the flow divider and 0.5 and 4 cm distal to the bifurcation, depending on the level of the flow divider in relation to the jaw line. After the 3D US scan, velocity waveforms were obtained by using pulsed-wave Doppler from the center stream of common carotid artery (CCA), approximately at the location corresponding to that of the last slice of the 3D US scan.

Image segmentation and carotid reconstruction. The acquired transverse images were segmented by using specially designed software. The software was used to manually select points on the vessel wall to which a smooth cubic spline or ellipse was fitted. The final lumen contours, combined with the positioning information from the EPOM device, allowed reconstruction of a smooth 3D geometry of the carotid bifurcation. This was performed by fitting longitudinal splines through the transverse contours. Reconstruction of the carotid artery bifurcation from 3D US images has been described (14).

Mesh generation and CFD. The mesh was generated by using an enhanced and validated in-house specially built mesh generator (23). This resulted in independent meshes containing between 20,000 and 80,000 hexahedral (structured) cells, depending on the lumen volume.

Figure 1 shows the computational surface mesh for subject 1 in both head positions. Note that the z-axis was the inferior/superior axis, but the x- and y-axis move with the US probe and were not the same in both scans.

The vessel wall was assumed to be rigid and impermeable with zero fluid velocity at the wall (the no-slip condition). At the inlet, Womersley profiles derived from the measured flow waveform in the CCA were imposed. The flow waveform was acquired from the center line velocity, assuming developed Womersley flow in the CCA (39). At the outlet, a constant flow ratio of 55:45 was adopted for all subjects between the internal (ICA) and external carotid artery (ECA) throughout the whole cardiac cycle, on the basis of the average value obtained from healthy young subjects (14). With the vessel geometry and inlet conditions known, the partial differential equations describing the movement of the fluid (mass conservation law and Navier-Stokes equation) can be solved numerically. This was done by using a differencing scheme called Quadratic Upwind Interpolation for Convective Kinematics implemented in CFX4 (1) on a SUN Blade 1000 Workstation (Ultraspire III processor, 512Mb RAM, 750 MHz). To eliminate start-up effects, two cardiac cycles, each consisting of 80 equally spaced time steps, were simulated. Refinement in time step or increase in simulated cycles resulted in instantaneous WSS differences below 0.5 and 0.01%, respectively. For the fluid properties, a blood density of 1,176 kg/m3 was set. The non-Newtonian behavior of blood viscosity was incorporated by using the Quemada model (7).

Comparative Study

The effect of change in head position was quantified by three geometric and three hemodynamic parameters.

Geometric parameters. The first geometric parameter is the mean lumen area. The cross-sectional area of the CCA, ICA, and ECA was averaged along its respective center line.

![Figure 1](image-url)
Second, a center line is a space curve that goes through the center of the vessel cross section in the transaxial plane. The nonplanarity parameter, described by King et al. (20), was used to give a measure of the tortuosity of the center line. Briefly, by performing single-value decomposition on the matrix formed by all points on the center line, a quantification measure of bifurcation nonplanarity ($S_{33}$) can be derived. Values of $S_{33}$ range from 0% for planar data to 33% for data distributed uniformly in space. A different measure of center line agreement, $z_{DIST}$, on the basis of a definition by Long et al. (22), is defined as the average distance between the center lines calculated for the two head positions. Because two different sets of center lines can be totally different yet have the same nonplanarity, $z_{DIST}$ is a more robust quantification of center line difference. In addition to $S_{33}$ and $z_{DIST}$, the linearity parameter ($S_{11}$) (20) was also derived for each of the three arteries. The $S_{11}$ describes how well the vessel center line can be approximated by a straight line. Note that the maximum length of each vessel covered in both scans was included in the analysis of nonplanarity and linearity.

Third, shapes of cross-sectional areas were compared between the two scans. To quantify shape difference, a shape factor introduced by Long et al. (22) was adopted. It was defined and evaluated here as follows: 1) the two cross sections to be compared were scaled to obtain a cross-sectional area of 1, 2) the center points were aligned, 3) the area lying inside both cross sections was denoted as $A$, whereas the area outside the overlapping area was $B$, and 4) the ratio $B/(A + B)$ was defined here as the shape factor. It is 0% for a perfect match between two shapes. Shape factors below 10% were within the sensitivity of MRI (22), but with 3D US, the uncertainty of shape factor was ~15% in the ECA (16).

Hemodynamic parameters. First, velocities are obvious parameters for flow comparison. Here, a velocity-dependent parameter was defined as the maximum velocity ($V_{\text{max}}$) at each time step in each cross section, averaged over the cardiac cycle and along the considered artery (CCA, ICA, ECA). The higher this parameter is for a constant flow rate and cross-sectional area, the more peaked the velocity profile is. The theoretical minimum occurs with plug flow and is equal to the mean flow rate divided by the lumen cross-sectional area.

Second, the most common hemodynamic wall parameter is the instantaneous WSS. For laminar flow, the instantaneous WSS can be computed as

$$\tau = \mu \cdot \frac{\partial \bar{u}}{\partial y}$$

here, $\mu$ is the dynamic viscosity, $\bar{u}$ is the velocity parallel to the wall, and $y$ is the normal distance from the vessel wall. The WSS is the tangential force exerted by a moving fluid on a wall. Time-averaged WSS is the WSS averaged over a cardiac cycle.

Third, the OSI is defined as follows (7):

$$\text{OSI} = \frac{1}{2} \left( 1 - \frac{1}{T} \int_{0}^{T} \frac{\tau}{\bar{\tau}} \, dr \right)$$

here $dr$ is the integration over time, $T$ is the cardiac period and ($\bar{\tau}$) is the instantaneous WSS. The OSI can be regarded physically as the fraction of a cardiac cycle during which the angle between the instantaneous WSS and the time-averaged WSS is $>90^\circ$. The theoretical maximum (0.5) is approximated when the instantaneous WSS oscillates around zero.

The average WSS and OSI values alone are not sufficient as markers for atherogeneity. The localization and area of vessel wall experiencing WSS and OSI above certain limits may also be important. For statistical purposes, the average WSS and OSI on the inner and outer walls of the ICA and ECA have been calculated separately. Here, inner wall stands for the half of the daughter vessel on the flow divider side, whereas outer stands for the other half of the wall, i.e., the half that could be viewed as the outer bend of the vessel. Similarly, the CCA has been divided into two portions, one on the ICA side and another on the ECA side.

RESULTS

Geometry

Figure 2, A–C, summarizes the findings for the arterial cross-sectional areas: both the sample-averaged means (black circles) and the individual changes (gray circles and dotted lines) are represented. Note that sample-averaged mean stands for the value averaged over all subjects in one head position. The $P$ value in Fig. 2 tests the hypotheses that the sample-averaged means in the two head positions are equal (paired Student’s t-test). For clarity, a $P$ value close to zero rejects the hypothesis and suggests that the sample-averaged means are different, whereas a $P$ value close to 1 concludes that the sample-averaged means are equal. Note that the uncertainty of 3D US for cross-sectional area measurement is ~10% for the CCA and ICA and 15% for ECA (P. Glor, unpublished communication). Table 1 gives the shape factor, i.e., the parameter describing the correspondence in cross-sectional shape between the straight and turned scans. Figure 3 overlays the vessel center lines of the straight and turned scans. The quantitative analysis is shown in Fig. 2D using $S_{33}$ (20), and Table 2 shows $z_{DIST}$ (22).

In Table 3, a statistical analysis of the center line agreement is presented. In the first column, the studied center line-related parameter is named. The second column gives the tested hypothesis, and the last column gives the $P$ value for this hypothesis using a Student’s t-test. Whether or not Student’s t-test was one- or two-tailed is reported in the third column. The table compares $z_{DIST}$, $S_{33}$, and $S_{11}$ between the two scans.

Flow

Figure 4A shows the mean flow rate in the CCA for all subjects in all scans, and Fig. 4, B–D contains the velocity-dependent parameter $V_{\text{max}}$. Table 4 shows in the first three rows the $V_{\text{max}}$ averaged over the subjects (means ± SD) in each of the arteries and each of the head positions. Similarly, the last three rows are the results obtained when the same boundary conditions (i.e., the same flow rate) were imposed in the two head positions. Therefore, Table 4 quantifies the change in $V_{\text{max}}$, and, with results given in the last three rows, suggests how much of the change can be attributed to the change in mean flow rate. Table 5 provides the statistical analysis of the velocity-related parameter $V_{\text{max}}$. Similar to Table 3, the studied parameter is in the first column, the hypothesis can be found in the second column, and the last column gives the $P$ value for this hypothesis using a Student’s t-test. Whether or not Student’s t-test was one- or two-tailed, is reported in the third column. A previous study (15) showed that the $V_{\text{max}}$ was reproducible within 10%.

To acquire an overall feel for the effect of head rotation on the distribution of vessel wall parameters, the WSS and OSI distribution in all subjects and all scans are shown in Figs. 5 and 6. Note that low values of WSS (<0.4 N/m²) have been associated with intima-media complex thickening, an early
marker of atherosclerosis (24). Interestingly, high physiological levels of shear stress (>1.5 N/m²) appear to induce endothelial quiescence and an atheroprotective gene expression profile (24). However, WSS values above the threshold of 400 N/m² can cause direct endothelial cell damage (21). It is well known that shear affects many functions of the arterial wall, but the exact thresholds for pathologically low and high WSS are disputable. Similarly, it is known that areas in which the OSI is elevated are prone to intimal thickening. No clear thresholds for “good” or “bad” OSI values have been defined in literature. Figure 7 shows the differences in WSS and OSI in different regions of the arteries.

DISCUSSION

Combination of 3D US and CFD

Atherosclerosis is often a focal disease (10). Plaque typically forms in which complex flow occurs, for example in areas of marked vessel curvature and arterial bifurcations. These observations have led to the implication of local hemodynamic factors, such as shear stress, in the initiation and progression of atherosclerotic disease. The combination of CFD with imaging technology provides a tool to study in vivo hemodynamics. Several imaging techniques have been used to acquire patient-specific 3D geometry and velocity data for CFD simulations. Among these, MRI is often regarded as the gold standard 3D-imaging modality for CFD, because it has been successfully used with CFD in the abdominal aorta (40), the carotid artery (27, 17) and left ventricle (32). MRI is noninvasive and can provide quantitative information about the arterial wall geometry, but it remains expensive and time consuming and is often limited in availability and patient tolerability.

US provides an alternative noninvasive method for visualizing the arterial tree and its hemodynamics. Widespread availability, relatively low cost and ability to provide simultaneous, real-time acquisition of both flow and vessel wall data have allowed US to become a first-line vascular imaging technique. With the development of 3D US techniques, US provides an alternative to MRI for acquiring the 3D geometry of large superficial arteries, such as the carotid arteries. The combination of 3D US with CFD has been validated in a number of technical studies. Barrat et al. (5) proved that the geometrical accuracy of a reconstructed carotid phantom was within −0.53 ± 3.39%. Augst et al. (2) found an accuracy of 0.5 N/m² for WSS values in the most sensitive regions of a carotid phantom. The combination of CFD and 3D US was also validated in vivo. In a reproducibility study, 3D US was found to be reproducible in the absence of area and center line misregistrations (15). Finally, 3D US was compared with black blood MRI as an imaging tool for carotid geometry reconstruction (16), and was proved to be an inexpensive substitute for MRI.

In this study, 3D US was preferred to MRI because of the relative small nuisance to the subject, the speed of imaging, and lower cost. Clearly, the applicability of 3D US to the carotid is affected by the location of the bifurcation point in the

Table 1. Shape factor for all subjects

<table>
<thead>
<tr>
<th>Subjects</th>
<th>CCA</th>
<th>ICA</th>
<th>ECA</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2.53</td>
<td>6.09</td>
<td>5.95</td>
</tr>
<tr>
<td>2</td>
<td>4.08</td>
<td>2.02</td>
<td>11.02</td>
</tr>
<tr>
<td>3</td>
<td>6.05</td>
<td>6.78</td>
<td>5.66</td>
</tr>
<tr>
<td>4</td>
<td>1.54</td>
<td>2.90</td>
<td>5.80</td>
</tr>
<tr>
<td>5</td>
<td>1.54</td>
<td>6.53</td>
<td>10.45</td>
</tr>
<tr>
<td>6</td>
<td>3.25</td>
<td>6.88</td>
<td>4.93</td>
</tr>
<tr>
<td>7</td>
<td>2.51</td>
<td>4.57</td>
<td>3.52</td>
</tr>
<tr>
<td>8</td>
<td>1.39</td>
<td>8.04</td>
<td>4.80</td>
</tr>
<tr>
<td>9</td>
<td>3.36</td>
<td>6.51</td>
<td>5.66</td>
</tr>
<tr>
<td>Mean</td>
<td>2.92±1.50</td>
<td>5.59±2.00</td>
<td>6.42±2.56</td>
</tr>
</tbody>
</table>

Values are percent in means ± SD. CAA, common carotid artery; ICA, internal carotid artery; ECA, external carotid artery.
neck. In those subjects with relatively high bifurcation points, the angle of the jaw often physically limits the accessibility of the US transducer to the carotid bulb, and/or the internal and external carotid arteries. This can lead to an inadequate geometric coverage for subject-specific flow simulations. In this study, all have relatively low bifurcation points, allowing sufficient geometrical information to be acquired.

**Geometrical and Hemodynamic Differences**

**Geometry.** In Fig. 2, A–C, it can be seen that the sample-averaged mean of the arterial cross-sectional area did not change significantly \( P < 0.05 \). Individual area changes were considered insignificant given that the 3D US uncertainty in measuring cross-sectional area was around 10% for the CCA and ICA, and 15% for the ECA. In a separate validation study (16), it was similarly shown that the shape factor was reproducible to within 10% for the CCA and ICA and 15% for the ECA. Table 1 shows that all values for the shape factor are lower than these thresholds, and therefore changes in cross-sectional shapes could be considered as insignificant.

Although cross-sectional areas and shapes were little affected by head rotation, the vessel center lines showed marked differences. Figure 3 gives a general impression of how much the center lines changed with head rotation, noting the large changes for subjects 1, 2, 6, and 7. Interestingly, this observation was consistent with the values of \( z_{DIST} \) in Table 2.

**Table 2. Center line distance for all subjects**

<table>
<thead>
<tr>
<th>Subjects</th>
<th>( z_{DIST} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>106.9</td>
</tr>
<tr>
<td>2</td>
<td>91.4</td>
</tr>
<tr>
<td>3</td>
<td>37.2</td>
</tr>
<tr>
<td>4</td>
<td>63.4</td>
</tr>
<tr>
<td>5</td>
<td>35.8</td>
</tr>
<tr>
<td>6</td>
<td>111.8</td>
</tr>
<tr>
<td>7</td>
<td>111.3</td>
</tr>
<tr>
<td>8</td>
<td>60.1</td>
</tr>
<tr>
<td>9</td>
<td>67.7</td>
</tr>
<tr>
<td>Mean</td>
<td>76.2 ± 30.3</td>
</tr>
</tbody>
</table>

Values are in micrometers and given as means ± SD. \( z_{DIST} \), measure of center line agreement in distance.

**Table 3. P values from Student’s t-test for center line agreement parameters**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Hypothesis</th>
<th>Tailed</th>
<th>( P ) Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>( z_{DIST} )</td>
<td>&lt;0 µm</td>
<td>one</td>
<td>3.3 ( 10^{-3} )</td>
</tr>
<tr>
<td>( S_{33} ) difference CCA</td>
<td>0%</td>
<td>two</td>
<td>0.57</td>
</tr>
<tr>
<td>( S_{33} ) difference ICA</td>
<td>0%</td>
<td>one</td>
<td>5.3 ( 10^{-3} )</td>
</tr>
<tr>
<td>( S_{33} ) difference ECA</td>
<td>0%</td>
<td>two</td>
<td>0.99</td>
</tr>
<tr>
<td>( S_{11} ) difference CCA</td>
<td>0%</td>
<td>two</td>
<td>0.33</td>
</tr>
<tr>
<td>( S_{11} ) difference ICA</td>
<td>0%</td>
<td>two</td>
<td>0.42</td>
</tr>
<tr>
<td>( S_{11} ) difference ECA</td>
<td>0%</td>
<td>two</td>
<td>0.12</td>
</tr>
</tbody>
</table>

\( n, 9 \) subjects. Difference, signed value of the difference between the turned and the straight position. \( S_{33} \), quantification measure of bifurcation nonplanarity; \( S_{11} \), linearity.
Conversely, the \( S_{33} \) seemed to be less of an indicative parameter, supporting the statement that \( z_{DIST} \) is a better measure of center line agreement than a nonplanarity parameter. Figure 2D shows that individual changes in nonplanarity can be significant (i.e., >1%, Ref. 20), but the sample-averaged value is not affected (\( P = 0.565 \); see both Fig. 2D and Table 3). Table 3 also provides the nonplanarity and linearity in each of the three branches (CCA, ICA, and ECA) separately with the paired Student’s \( t \)-test. The only general observation (at \( P < 0.05 \)) was that the CCA became more planar after head rotation (overall straight CCA nonplanarity greater than overall turned CCA nonplanarity; \( P = 0.0053 \), one-tailed paired Student’s \( t \)-test).

Hemodynamics. From Fig. 4A it can be seen that the mean flow rate in the straight CCA changed considerably when the head was turned to the left. Examination of individual data showed that subjects 3, 4, and 5 had a sharp increase in mean flow rate from the straight to turned position (+26.7 ± 4.5%), whereas subjects 1, 2, 6, 7, and 9 had a considerable reduction in flow rate (−35.1 ± 13.5%). However, the sample-averaged common carotid flow rate showed a relatively small change (−10.6 ± 31.8%).

In Table 4, it can be seen that change in \( V_{\text{max}} \) with turning of the head should not be ignored; the standard deviation of the error exceeded the 10% uncertainty found in a previous study (15). Figure 4, B–D and the first three rows of Table 5 confirmed this statement: the average change in \( V_{\text{max}} \) was found to be >10% with a significance of \( P < 0.05 \) (one-tailed paired Student’s \( t \)-test). Again, there was certainty of change in the velocity profile, but it was impossible to predict whether the velocity parameter would increase or decrease. In Table 5 (rows 4–6), the sample-averaged values proved to be equal (mean difference in \( V_{\text{max}} = 0\% \), \( P > 0.05 \)). The question of what causes the \( V_{\text{max}} \) to change is raised. With little differences in areas and shapes of carotid artery cross sections, the change has to be caused by the altered center line or flow rate. The simulations were repeated with identical flow rates as boundary conditions (i.e., the same boundary conditions (i.e., the same flow rate) were imposed in the 2 head positions).

Table 4. Change in maximum velocity

<table>
<thead>
<tr>
<th>Artery</th>
<th>Straight, cm/s</th>
<th>Turned, cm/s</th>
<th>Difference, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>CCA</td>
<td>23.91±5.09</td>
<td>21.19±5.91</td>
<td>−12.89±25.51</td>
</tr>
<tr>
<td>ICA</td>
<td>19.39±4.77</td>
<td>17.49±5.90</td>
<td>−11.80±28.00</td>
</tr>
<tr>
<td>ECA</td>
<td>21.41±8.14</td>
<td>21.52±11.03</td>
<td>−2.93±36.02</td>
</tr>
<tr>
<td>CCA</td>
<td>25.18±4.89</td>
<td>24.74±3.73</td>
<td>−0.98±8.65</td>
</tr>
<tr>
<td>ICA</td>
<td>22.17±4.13</td>
<td>22.39±3.24</td>
<td>1.68±9.11</td>
</tr>
<tr>
<td>ECA</td>
<td>19.40±3.63</td>
<td>19.29±3.98</td>
<td>−1.16±8.84</td>
</tr>
</tbody>
</table>

Values are mean ± SD; \( n \) 9 subjects. The difference in maximum velocity (\( V_{\text{max}} \)) between the turned and straight scans is expressed as a percentage of the mean \( V_{\text{max}} \). The last three rows give the results when the same boundary conditions were imposed in the 2 head positions instead of using the flow rate calculated in each head position.

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Fig. 4. A: mean flow rate in the CCA was calculated using pulsed Doppler ultrasound. B–D: velocity parameter (\( V_{\text{max}} \)) in CCA, ICA, and ECA. The closed circle on the vertical line gives the sample-averaged mean ± SD. Each of the dotted lines corresponds to a single subject. The \( P \) value tests the hypothesis that the sample-averaged means in the two head positions are equal (paired Student’s \( t \)-test).
Fig. 5. Overview of time-averaged wall shear stress (WSS) for all subjects in both head positions. Captions refer to subject number and head position (i.e., 1: straight). Units are N/m² on a logarithmic scale.
conditions for both head positions to address this point more thoroughly. From Table 4, it can be seen that here, the $V_{\text{max}}$ RMSE did not exceed the 10% uncertainty. This was confirmed in Table 5, in which the change in $V_{\text{max}}$ proved $<10\%$ ($P > 0.9$ for all arteries, one-tailed paired Student’s $t$-test), which suggested that the difference in flow rate contributed more to the changes in velocity than the center line variation.

A Gauss-Newton nonlinear data fit also revealed more important dependence of the velocity on the flow rate than the center line.

It is clear from Figs. 5–7 that WSS and OSI patterns can change significantly with head rotation. In some cases, these changes were only quantitative: subject 6 had a high OSI zone in the straight position (Fig. 6K), and when the head was
turned, this high OSI zone was still present at more or less the same location (Fig. 6L). In other subjects, patches of high OSI either changed positions (subject 2, Fig. 6, C–D) or magnitude (subject 3, Fig. 6, E–F). The largest WSS differences were found in the daughter vessels (high variance in Fig. 7, B–C and E–F), whereas the inner walls of the ICA and ECA were the most affected by changes in OSI (Fig. 7, H–I). The first question arising from this finding was whether or not the change in hemodynamic parameter distributions could be correlated with head rotations. Subjects 2, 6, 7, and 9 all experienced a reduction in OSI in the vicinity of the apex, suggesting that turning the head to the left might produce a more advantageous hemodynamic environment in the right carotid arteries. The opposite was true for subjects 1, 3, 4, 5, and 8 whose right carotid OSI increased. Similar findings were made for the WSS. Therefore, there was no straightforward relationship between head position and the hemodynamic effect. Note that the vicinity of the apex is defined here as 10 mm above and 20 mm below the apex.

Correlations between geometric and hemodynamic changes. The second question arising from the differences in hemodynamic wall parameter distributions, was whether or not these
differences could be correlated with changes of flow rate, center line, or any other geometric property. Unsurprisingly, the change in the velocity parameter was directly related to the change in flow rate [linear correlation ($R$) of 0.97, 0.95, and 0.75 for the correlation between change in flow rate and change in velocity-parameter in CCA, ICA, and ECA, respectively]. This correlation can also be seen in Fig. 4 in which a rise in flow rate (Fig. 4A) is accompanied by an increase in $V_{\text{max}}$ (Fig. 4, B–D). The WSS correlated with the flow rate in the CCA ($R = 0.77$ and 0.87 for the CCA on the ICA and ECA sides, respectively) and inner part of the ICA ($R = 0.81$). Furthermore, the flow rate was correlated with the OSI (Fig. 8, A–B); with an increase in flow rate, the OSI decreased. This was not surprising, because high OSI occurs in regions of slow and recirculating flow. Thus slightly elevated flow rate is likely to discourage flow recirculation. This effect was highly noticeable in the part of the CCA from which the ECA emerges ($R = 0.81$, Fig. 8A) and in the outer wall of the carotid bulb ($R = 0.78$, Fig. 8B). The mean flow rate did not correlate with any of the described geometric parameters, i.e., cross-sectional areas and shapes, $z_{\text{DIST}}$, $S_{33}$, and $S_{11}$.

The $V_{\text{max}}$, WSS, and OSI correlated with certain geometric parameters. Apart from the dependence on the flow rate, $V_{\text{max}}$ in the ECA correlated with the CCA nonplanarity ($R = 0.81$); with planarification of the CCA, the $V_{\text{max}}$ was lowered in the ECA. This means that the flow profile in the ECA was less peaked when the CCA became more planar. The change in overall nonplanarity influenced the WSS on the inner wall of the ICA ($R = 0.79$, Fig. 8C), i.e., the high shear rate region at the apex and inner wall of the bulb was reduced with overall planarification. Furthermore, $z_{\text{DIST}}$ correlated with $V_{\text{max}}$ in the ICA ($R = 0.77$), with the WSS on the inner wall of the ICA ($R = 0.83$), and more importantly with the WSS on the outer wall of the ICA ($R = 0.78$), which is the region prone to plaque formation (36). When the center line difference was large, values of WSS on the outer wall of the bulb were lowered. This meant that when the head was turned from a straight position to the left, a larger region of the bulb was exposed to a WSS lower than 0.4 $N/m^2$, the suggested threshold for atherogenic cellular behavior (24).

The number of studies examining the atherogenic geometric features of a carotid geometry is not overwhelming. Caro et al. (8) made the first suggestion saying that nonplanarity was an important factor influencing arterial flows. Figure 8C, which shows the correlation between nonplanarity and high shear regions, provides further evidence for this statement. Thomas et al. (37) investigated the effect of planarity and curvature on carotid hemodynamics and concluded that overall vessel curvature rather than planarity/nonplanarity had a more important impact as a potential geometric risk for atherogenesis. Note that they used a different definition for planarity. In this study, the finding that the overall $z_{\text{DIST}}$ correlated better to hemodynamic changes than the overall nonplanarity, agreed with Thomas’s results.

Summary and Implications

This study quantified the geometric changes and their hemodynamic consequences at the carotid bifurcation when the head is rotated. It was found that cross-sectional areas and shapes were only slightly influenced by head rotation, but the positions of the vessels changed. The only general observation was a planarification of the CCA. Other geometrical changes in nonplanarity, linearity, or overall center line agreement were subject dependent. Similar findings were made for the mean CCA flow rate; the sample-averaged mean CCA flow rate changed only slightly from one head position to another, but significant changes in individual flow rates were observed.

Changes in geometry and flow rate resulted in hemodynamic changes: flow profiles, WSS, and OSI distributions were affected in various ways. Reduction in flow rate resulted in lowered WSS in the CCA and the inner part of the ICA. Similarly, the change in OSI distribution appeared to be linked to the change in flow rate, e.g., on the outer wall of the bulb.

Fig. 8. Correlations between change in flow rate and difference in mean OSI in CCA part from which ECA emerges (A), change in flow rate and difference in mean OSI in outer part of ICA (B), and change in overall nonplanarity and difference in mean WSS in inner part of ICA (C). Δ, true difference between the straight and turned position. Ext, half of the CCA at the ECA side; out, outer part; inn, inner part.
Our data suggested the following relations between geometrical and hemodynamic changes: 1) with planarization of the CCA, the flow profile in the ECA became blunter; 2) when turning the head to the left made the bifurcation more planar, it resulted in lowered WSS in the apex region; and 3) the change in WSS was correlated with the change in vessel center line, i.e., the WSS on the outer wall of the bulb decreased when center line difference between head positions was large.

Considering that the center line difference is relatively large in most of the subjects and that WSS reduction on the outer wall of the bulb is usually an adverse hemodynamic phenomenon, one could conclude that the straight head position is hemodynamically preferable to the left-turned head position for right carotid arteries. However, the large variety of possible hemodynamic conditions after head rotation undermines such a general statement. The fact that hemodynamic situations can be very different in the studied head positions, sheds a new light on 3D US scan protocol. Until recently, the US operator was large.

The altered hemodynamic condition in different head positions has an effect on studies concerning the potential pitfalls of carotid implants such as carotid stents (18) or carotid emboli diverters (35). The carotid stent could work as planned in a particular patient, be subjected to low WSS values in a different head position, yielding a situation prone to restenosis. Similar lessons can be learned for studies on the vulnerability of carotid plaque. Apart from the composition of the carotid plaque (26), the hemodynamic forces acting on the plaque are the main parameters when predicting plaque rupture (13). Plaque can be harmless in one head position, but could be undergoing severe shear forces in another head position. Therefore, plaque can rupture in altered head positions, and this could be the cause of telephone strokes.

In conclusion, with the important incidence of atherosclerosis on one hand, and the correlation between regional hemodynamics and the atherosclerotic process on the other hand, carotid flow behavior becomes a topical subject. In this study, the right carotid arteries of nine subjects were investigated in the supine position with straight and left-turned head positions, respectively.

The 3D geometry was reconstructed by using 3D US and the carotid hemodynamics were calculated by combining 3D US with CFD. The vessel center line was the only geometrical feature that showed noticeable change, whereas cross-sectional shape and area were altered little by head rotation. The planarization of the CCA with head rotation was the only general feature. The overall center line difference proved to have a more powerful effect on the flow than nonplanarity on its own. It was shown that hemodynamic differences between two head positions were important: flow rate, WSS, and OSI distribution changed significantly. Therefore, unnatural, long-lasting head positions cannot be left out in the search for possible causes for telephone strokes. Correlations between geometrical changes and changes in flow properties were suggested. Based on this likely relationship between geometry and flow, defining atherogenic carotid geometries may become feasible. Furthermore, this study showed that there is a need for standardization of the choice of head position in the 3D US scan protocol. Finally, it was shown that carotid plaque and carotid stents need to be studied in several head positions before they can be considered safe.

REFERENCES


