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Chapter 5

Wall shear stress calculations based on 4D flow MRI and CFD: a comparison in healthy carotid bifurcations

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Wall shear stress calculations based on 3D cine phase contrast MRI and computational fluid dynamics: a comparison study in healthy carotid arteries
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5.1 Introduction

Flow induced wall shear stress (WSS) is an important biomechanical parameter, widely accepted to influence the endothelial function in the vasculature and thereby involved in many pathophysiological processes related to cardiovascular diseases [1]. The majority of studies have shown that, in the presence of risk factors, atherosclerotic plaques mainly form in the regions of low and oscillatory WSS [2-7]. Other studies have also suggested that high WSS has a pathogenic effect on the initiation of aneurysm formation, while low WSS facilitates aneurysm growth [8]. These findings suggest that knowledge of WSS may provide vital information about the initiation and progression of vascular diseases. Nevertheless, WSS assessment has not been integrated into clinical practice. This is mainly due to the difficulty of determining WSS in vivo.

WSS can be calculated by multiplying the blood viscosity with wall shear rate (WSR), the latter being the gradient of blood flow velocity in the normal direction of the vessel wall. The current gold standard method for determining 3D blood flow velocities and WSS is computational fluid dynamics (CFD). CFD has the advantage of solving velocities at a high spatial and temporal resolution. However, the disadvantage of CFD is that it requires non-clinical expertise and extensive computational power. Phase contrast (PC) MRI can also measure time resolved 3D velocities and several methods have been developed to quantify WSS using PC MRI velocities. In earlier studies, WSS calculation was based on parabolic fitting of the velocities measured by 2D PC MRI [9-16]. This approach was improved by using b-splines for the fitting of velocities measured by 3D cine PC MRI in more recent studies [17,20]. A drawback of these studies was that WSS calculations were limited to planar slices within an artery. Recently, a new algorithm for calculation of WSS throughout the 3D luminal surface using PC MRI velocities was introduced [21]. In the current study, our objective was to compare WSS distributions on the entire 3D luminal surface of carotid arteries calculated using 3D cine PC MRI velocities with those calculated using CFD. To our knowledge, this is the first study that compares the WSS distributions based on PC MRI and CFD on the entire luminal surface of the carotid arteries.

Figure 5.1: a The luminal surface of the carotid artery was segmented on the magnitude images and subsequently inspected in phase images using ITK-SNAP. b The segmented surface was split into CCA, ICA and ECA using VMTK.
5.2 Methods

5.2.1 MRI scans

Six healthy volunteers (25 – 30 years old) were scanned on a 3 T MRI system (Intera with software version 3.2.1 for the first volunteers and Ingenia with software version 4.1.3 for the last volunteer, Philips Healthcare, The Netherlands) using a dedicated 8-channel bilateral carotid coil (Chenguang Medical Technologies, Shanghai, China). A retrospectively gated 3D transient field echo (TFE) sequence with RF spoiling, a TFE factor of 8 and a parallel imaging factor (SENSE) of 2 with symmetric four-point velocity encoding was obtained with an isotropic non-interpolated resolution of 0.625 mm and a temporal resolution of 138 ± 11 ms (8 timepoints / heart cycle) as part of a carotid scan protocol (FOV/ TR/ TE/ flip angle/ bandwidth/ venc RLxAPxFH, 140x140x30 mm/ 6.7 ms/ 3.1 ms/ 15°/ 299 Hz/px/ 60x60x100 cm/s). This 3D cine PC MRI sequence took ~ 20 minutes per volunteer, depending on the heart rate. No velocity aliasing was found in the images. We corrected the PC MRI datasets for possible phase-offset errors and visually checked for artifacts.

5.2.2 Segmentation and meshing

Three carotid arteries were excluded due to motion and pulsation artifacts and thus nine were found suitable for the analysis. We manually segmented the luminal surface of the carotid arteries on the magnitude images using ITK-SNAP (22). We excluded small scale side branches. The segmentations were inspected on the phase images and corrected if necessary (Figure 5.1a). We smoothed the resulting surface with Taubin smoothing approach and split the carotid artery surface into three arteries (Figure 5.1b), the common carotid artery (CCA), internal carotid artery (ICA) and external carotid artery (ECA) using the open-source Vascular Modeling Toolkit (VMTK version 1.0.0) (23). We performed the lumen segmentation only on the PC MR images acquired during diastole (t = 742 ms in Figure 5.2a) because at this cardiac phase the blood flow fluctuations were minimal.

Figure 5.2: a A representative flow waveform recorded at 8 time points. WSSMRI was calculated using only the velocities recorded during diastole (t = 742 ms). b The generalized waveforms reported by Lee et al. (24). WSSCFD was calculated using only the velocities recorded during diastole (t = 573 ms). The generalized and the recorded flow waveforms have equal mean flow rates.
In the next step, we generated a tetrahedral volume mesh by using commercial mesh generation software package Gambit (Ansys). The element size was 0.12 mm at the vessel wall and increased inwardly. Each volume mesh contained approximately 1 million elements. For the computations of WSS based on PC MRI and CFD, we used the same volume mesh, enabling a one-to-one comparison between both WSS measures.

5.2.3 WSS calculations using 3D cine PC MRI velocities

WSS based on PC MRI (WSS\text{MRI}) was calculated using the 3D cine PC MRI velocities and the volume mesh, as described before (21). In summary, we first determined the inward normal vector of each mesh node on the luminal surface. Along the normal direction, we interpolated the velocities at two inward equidistant points, which were at 1.5 and 3 mm on the inward normal. We imposed the velocity to zero on the vessel wall. We subsequently fitted a curve to the interpolated velocities using a smoothing spline. We derived the slope of this curve at the vessel wall which gives the WSR. We finally multiplied WSR with the shear dependent viscosity (Carreau-Yasuda model) resulting in WSS vectors for all mesh nodes on the vessel wall. We calculated only the diastolic WSS\text{MRI}. This calculation took $\sim 15$ minutes for each carotid artery.

5.2.4 WSS calculations using CFD

We calculated the CCA, ICA and ECA flows using PC MRI velocities recorded at 8 time points (Figure 5.2 a) as inflow boundary condition. The sum of the ICA and ECA flows did not always match the CCA flow. We therefore corrected the ICA and ECA flows to match their sum with the CCA flow, while maintaining the ratio of the ICA and ECA flows constant. Due to the limited temporal resolution of the 3D cine PC MRI measurements, the recorded flow waveform was flattened (Figure 5.2 a). To account for the temporal flow changes in higher frequencies, we used the generalized flow waveforms of CCA and ICA reported by Lee et al. as substitute for the recorded flow waveforms (Figure 5.2 b) (24). We scaled the generalized flow waveforms so that they had mean flow rates equal to the measured mean flow rates.

As inflow profile, we used a single velocity profile, which was obtained by using both axial and in plane velocities and acquired during diastole ($t = 742$ ms in Figure 5.2 a). We scaled it at each time point in the cardiac cycle so that the flow was increased or decreased according to the flow waveform but the velocity profile shape did not change (Figure 5.3). ECA outlet was left as stress free. Fluid density was set at $1060$ kg/m\(^3\) and the Carreau-Yasuda model was used to mimic the non-Newtonian behavior of blood with the parameters used in (25).

We performed the CFD simulations on a standard desktop computer (Intel Xeon six core processor, 2.40 GHz CPU and 12 GB RAM) using the commercial finite element software FIDAP 8.7.4 (Ansys). We set the temporal resolution to 5 ms and performed the simulations for 2 cardiac cycles. The CFD simulations took $\sim 15$ hours for each carotid artery. Although CFD results were acquired for a complete cardiac cycle, we analyzed WSS\text{CFD} only in
diastole \((t = 573\ 	ext{ms}\) in Figure 5.2 b where CCA flow was equal to the measured diastolic CCA flow \((t = 742\ 	ext{ms}\) in Figure 5.2 a.

### 5.2.5 Analysis of WSS\text{MRI} and WSS\text{CFD}

#### WSS magnitudes

As shown in the previous studies, the spatial resolution of the velocities may affect the calculated WSS\text{MRI} values \((17, 21, 26)\). To study the effect of resolution, we down-sampled the CFD velocities into PC MRI resolution and calculated WSS (WSS\text{CFD,lowres}) based on the down-sampled velocity field. To down-sample the CFD velocity field, we first interpolated the CFD velocities to a cubic grid with an isotropic resolution of 0.1 mm. The isotropic PC MRI voxels of 0.625 mm contained \(\sim 216\) of these 0.1 mm isotropic voxels. We averaged these \(\sim 216\) velocity values and obtained a down-sampled CFD-based velocity field that mimicked the PC MRI velocity data. The down-sampling procedure is schematically depicted in Figure 5.4.

**Figure 5.4:** a CFD velocities interpolated to 0.1 mm cube grid. b The interpolated CFD velocities placed in PC MRI voxels. c Each voxel contained \(\pm 216\) of 0.1 mm isotropic velocities that were averaged. d The down-sampled CFD velocities into PC MRI resolution of 0.625 mm.

We present the WSS\text{CFD}, WSS\text{MRI} and WSS\text{CFD,lowres} magnitude maps of each artery. We report the mean of WSS\text{CFD}, WSS\text{MRI} and WSS\text{CFD,lowres} per artery and also within CCA, ICA and ECA. To check whether a systematic relation exists between WSS\text{CFD}, WSS\text{MRI} and WSS\text{CFD,lowres}, we plotted mean WSS\text{CFD} versus the difference of mean WSS\text{MRI} and WSS\text{CFD}, and also mean WSS\text{CFD} versus the difference of mean WSS\text{MRI} and WSS\text{CFD,lowres} within CCA, ICA and ECA. For the same reason, Bland-Altman analysis, which contained all data points of all arteries, was performed between WSS\text{MRI} and WSS\text{CFD}, and between WSS\text{MRI} and WSS\text{CFD,lowres}. Statistical significance was assessed using paired \(t\)-test.

We also compared the location of low, medium and high WSS regions. We labeled all WSS\text{MRI} and WSS\text{CFD} magnitudes according to three categories representing the low, medium and high tertiles. For each node, we compared the categorization of WSS\text{MRI} and WSS\text{CFD}. We report the percentage of the nodes having the WSS\text{MRI} and WSS\text{CFD} values labeled in the same tertile. Note
that the nodes on the luminal surface were uniformly distributed; each tertile therefore represented 33% of the total lumen surface area.

**WSS directions**

We calculated the angles between $\text{WSS}_{\text{MRI}}$ and $\text{WSS}_{\text{CFD}}$ vectors for all mesh nodes on the luminal surface. We generated an angle map for all carotid arteries. We report the mean angle of the luminal surface and within the low, medium and high WSS tertiles. Histogram analysis was also performed for the angles that were sorted in three tertiles.

### 5.3 Results

![Fig. 5.5](image)

**Figure 5.5:** Left: $\text{WSS}_{\text{CFD}}$, middle: $\text{WSS}_{\text{MRI}}$, right: $\text{WSS}_{\text{CFD, lowres}}$ magnitude [Pa] maps during diastole. The numbers 1 to 5 indicates the volunteers; while L and R represent left and right carotid arteries respectively.
Table 5.1: The mean WSS<sub>CFD</sub>, WSS<sub>MRI</sub> and WSS<sub>CFD,lowres</sub> [Pa] within CCA, ICA and ECA.

<table>
<thead>
<tr>
<th></th>
<th>CCA</th>
<th>ICA</th>
<th>ECA</th>
</tr>
</thead>
<tbody>
<tr>
<td>CFD</td>
<td>0.81 ± 0.28 Pa</td>
<td>1.07 ± 0.52 Pa</td>
<td>1.09 ± 0.46 Pa</td>
</tr>
<tr>
<td>MRI</td>
<td>0.60 ± 0.17 Pa</td>
<td>0.73 ± 0.26 Pa</td>
<td>0.56 ± 0.18 Pa</td>
</tr>
<tr>
<td>CFDlowres</td>
<td>0.57 ± 0.19 Pa</td>
<td>0.59 ± 0.24 Pa</td>
<td>0.52 ± 0.20 Pa</td>
</tr>
</tbody>
</table>

Figure 5.6: Bland-Altman plots a between WSS<sub>MRI</sub> and WSS<sub>CFD</sub> and b between WSS<sub>MRI</sub> and WSS<sub>CFD,lowres</sub>. The regression shown in black and the mean and the limits of agreement lines are shown in red. The colors indicate the density of the data points scaled with the colorbar on the right.

5.3.1 WSS magnitudes

WSS<sub>CFD</sub>, WSS<sub>MRI</sub> and WSS<sub>CFD,lowres</sub> magnitude maps for nine carotid arteries are shown in Figure 5.5. The carotid bulb walls were mostly exposed to low WSS magnitude (blue) whereas the bifurcation apex and the inner walls of ICA and ECA had high WSS magnitude (red). Although these patterns were present in WSS<sub>CFD</sub>, WSS<sub>MRI</sub> and WSS<sub>CFD,lowres</sub> maps, the magnitude of WSS was different in each map. WSS<sub>MRI</sub> was lower than WSS<sub>CFD</sub> (0.62 ± 0.18 Pa versus 0.88 ± 0.30 Pa, \( P < 0.01 \)), but was only slightly higher than WSS<sub>CFD,lowres</sub> (0.56 ± 0.18 Pa, \( P < 0.01 \)). The mean WSS<sub>CFD</sub>, WSS<sub>MRI</sub> and WSS<sub>CFD,lowres</sub> are presented in Table 5.1. For higher WSS magnitudes, the difference between the WSS<sub>CFD</sub> and WSS<sub>MRI</sub> was larger (\( r^2 = 0.4 \)). Bland-Altman plot presented in Figure 5.6 a shows this trend. Such a trend was absent in the Bland-Altman comparison of WSS<sub>MRI</sub> and WSS<sub>CFD,lowres</sub> (\( r^2 = 0.01 \)), which is shown in Figure 5.6 b.

Figure 5.7 shows WSS maps labeled according to the three categories where dark blue, light blue and yellow represent the regions of low, medium and high WSS. The percentage of the WSS<sub>CFD</sub> and WSS<sub>MRI</sub> values labeled in the same tertile is presented in Table 5.2 for each artery. On average, 68.7 ± 4.4% of the low and 69.0 ± 8.9% of the high WSS<sub>CFD</sub> and WSS<sub>MRI</sub> values matched.

5.3.2 WSS directions

The maps of the angles between WSS<sub>CFD</sub> and WSS<sub>MRI</sub> vectors are shown in Figure 5.8. The angles were small inside CCA and at the distal part of the ECA but larger around the carotid bulb and
Table 5.2: The WSS_{CFD} and WSS_{MRI} match [%] based on division of WSS values into three categories.

<table>
<thead>
<tr>
<th></th>
<th>Low WSS tertile</th>
<th>Medium WSS tertile</th>
<th>High WSS tertile</th>
</tr>
</thead>
<tbody>
<tr>
<td>1L</td>
<td>65.3%</td>
<td>48.7%</td>
<td>69.5%</td>
</tr>
<tr>
<td>2L</td>
<td>64.1%</td>
<td>33.2%</td>
<td>57.9%</td>
</tr>
<tr>
<td>2R</td>
<td>66.5%</td>
<td>45.8%</td>
<td>70.9%</td>
</tr>
<tr>
<td>3L</td>
<td>67.9%</td>
<td>40.7%</td>
<td>54.2%</td>
</tr>
<tr>
<td>3R</td>
<td>74.7%</td>
<td>63.0%</td>
<td>84.7%</td>
</tr>
<tr>
<td>4L</td>
<td>78.0%</td>
<td>51.3%</td>
<td>67.2%</td>
</tr>
<tr>
<td>4R</td>
<td>65.3%</td>
<td>50.0%</td>
<td>78.7%</td>
</tr>
<tr>
<td>5L</td>
<td>67.4%</td>
<td>37.0%</td>
<td>65.8%</td>
</tr>
<tr>
<td>5R</td>
<td>68.9%</td>
<td>51.7%</td>
<td>72.0%</td>
</tr>
<tr>
<td>Mean</td>
<td>68.7%</td>
<td>46.8%</td>
<td>69.0%</td>
</tr>
</tbody>
</table>

Figure 5.7: Maps of WSS in the nine carotid arteries where the WSS magnitudes are divided into three categories. Dark blue: low WSS, light blue: medium WSS, yellow: high WSS regions. WSS_{CFD} (left) WSS_{MRI} (right).
the apex. The mean angle was $65.6 \pm 17.4^\circ$ at low, $28.9 \pm 10.0^\circ$ at medium and $20.3 \pm 8.2^\circ$ at high WSS tertiles. The histograms of the angles that were sorted according to three tertiles are shown in Figure 5.9. The angle between WSS\textsubscript{CFD} and WSS\textsubscript{MRI} vectors was smaller than $30^\circ$ in $35\%$ of nodes in the low WSS tertile, $67\%$ of the nodes in the medium WSS tertile and $80\%$ of the nodes in the high WSS tertile. Within the low WSS tertile, the angles were larger and varying. We found a particularly large angle of deviation in the proximal ICA of 1L. This seemed to be due to the helical flow pattern seen in the PC MRI data but not in the CFD model. Overall, the mean angle between WSS\textsubscript{CFD} and WSS\textsubscript{MRI} vectors was $38.2 \pm 9.0^\circ$.

Figure 5.8: The map of angles [$^\circ$] between WSS\textsubscript{CFD} and WSS\textsubscript{MRI} vectors throughout the luminal surface.

Figure 5.9: The histograms of the angles [$^\circ$] between WSS\textsubscript{CFD} and WSS\textsubscript{MRI} vectors within the low, medium and high WSS tertiles.
5.4 Discussion

In this study, the WSS vectors based on 3D cine PC MRI measurements and CFD simulations were compared in healthy carotid arteries. The strength of the current study with respect to the previous studies is that we calculated PC MRI-based WSS vectors on the entire 3D luminal surface instead of manually selected planar slices within an artery. Obtaining WSS vectors on the entire 3D luminal surface is essential, since WSS distributions in the arteries are heterogeneous and planar slices may not be representative for the entire luminal surface. In this respect, our study is the first to compare subject-specific WSS distributions based on 3D cine PC MRI velocities with those based on CFD on the 3D lumen surface of the carotid arteries. Our results showed that the spatial patterns of WSS\textsubscript{MRI} were in good agreement with those based on CFD. Regions of low WSS along the carotid bulb and regions of high WSS at the inner walls of ECA and ICA were found by both PC MRI- and CFD-based WSS calculations. However, WSS was lower when calculated using PC MRI velocities. By down-sampling the CFD velocity into PC MRI resolution, we were able to mimic the PC MRI velocity field, thereby demonstrating that this underestimation is caused by the limited resolution of PC MRI. We found that the difference between WSS\textsubscript{CFD} and WSS\textsubscript{MRI} increases in regions with higher WSS values. This finding indicates that PC MRI underestimates high WSS values more than low WSS values. This is because velocity gradients disappear within a voxel by the averaging effect of PC MRI, and the effect is more prominent for higher velocity gradients. Such a trend was not seen for the difference between WSS\textsubscript{CFD} and WSS\textsubscript{MRI,lowres}, demonstrating that the low resolution of the MRI was responsible for the large differences at high WSS.

The directions of the WSS\textsubscript{CFD} and WSS\textsubscript{MRI} vectors were also compared. The angles were mainly small in the CCA and at the distal part of ECA because the flows within these segments were mainly in the axial direction of the vessel. We also found larger angles also in regions of low WSS and disturbed flows. The large deviation in direction of WSS\textsubscript{CFD} and WSS\textsubscript{MRI} vectors in these regions could be due to partial voluming effects and the low signal-to-noise ratio (SNR) of the PC MRI measurements.

The mean WSS values reported in the current study are in agreement with those reported previously. The mean diastolic WSS values reported in the literature were in the range of 0.58 – 0.61 Pa for CCA (10, 11) and 0.55 – 0.70 Pa for ICA (12, 13), while we found the mean diastolic WSS values in the range of 0.60 ± 0.17 Pa for CCA and 0.73 ± 0.26 Pa for ICA. Although PC MRI- and CFD-based WSS calculations have never been compared for in vivo carotid arteries, such comparison studies were performed for realistic models of carotid arteries (16, 27). These previous studies on carotid phantoms also reported a good qualitative agreement between PC MRI- and CFD-based WSS results. Kohler et al. showed that PC MRI measurements with an in-plane resolution of 0.7 mm resulted in 40% lower mean WSS values than those predicted by CFD (27). Papathanasopoulou et al. obtained higher WSS values with PC MRI velocities, but they used different segmentations for PC MRI- and CFD-based calculations (16). Studies have also compared WSS\textsubscript{MRI} and WSS\textsubscript{CFD} for in vivo and in vitro intracranial aneurysms (19, 27–30) and found good qualitative agreement between them. Similar to our findings, WSS was underestimated for intracranial aneurysms when calculated using PC MRI, and this effect was more prominent for higher WSS values.

Our study showed the validity of the assessment of WSS distribution based on PC MRI by comparing it with CFD-based WSS. However, CFD results are also dependent on a number of assumptions and the chosen boundary conditions, which may not reflect the in vivo situation. In
that respect, by showing a good agreement between PC MRI-based WSS and WSS based on the
down-sampled CFD velocities, we can also interpret our results as the validation of subject-specific
CFD-based WSS calculation using PC MRI measurements. Thus, with the high correlation between
the two methods, it can safely be stated that the chosen boundary conditions and assumptions of
CFD result in in vivo WSS distribution at diastole.

There were three main limitations of our study design. First, we compared WSS$_{\text{CFD}}$ and WSS$_{\text{MRI}}$
values during diastole only. This is a direct result of the choice for increased spatial resolution of
the PC MRI measurements at the expense of temporal resolution. During our PC MRI scans, we
recorded the velocities with a temporal resolution of $138 \pm 11$ ms, which is low compared with
literature values (20, 31). Due to the limited temporal resolution, we were unable to capture the
large fluctuations in flow that are known to occur particularly during systole. For this reason, we
decided not to compare the systolic data. Since the analysis was performed at a single time step,
time dependent hemodynamic parameters such as oscillatory shear index were not computed. To
analyze PC MRI-based WSS calculations during systole and to obtain time dependent parameters,
higher temporal resolution will be required in future studies. Second, by assuming rigid walls, we
neglected the pulsatility of carotid arterial walls in our CFD simulations. However, we segmented
the vessel walls on the diastolic images and we compared the WSS$_{\text{CFD}}$ and WSS$_{\text{MRI}}$ values only
during diastole. We assume that the use of rigid walls therefore had a limited impact on our
calculations. Finally, we used the diastolic spatial velocity profile shapes within the CCA and ICA
throughout the cardiac cycle. As a result, we were unable to capture the helical flow that we
observed in PC MRI data of carotid artery 1L in our CFD model. Acquisition of PC MRI data in
higher temporal resolution and applying time varying velocity profiles as boundary conditions in the
CFD simulations may prevent this limitation in future studies.

5.4.1 Clinical relevance
Spatial WSS patterns strongly affect endothelial cell signaling and early events in atherosclerosis (1–7). These patterns were estimated similarly by PC MRI- and CFD-based calculations. For
this reason, both WSS measures can be used when knowledge of low and high WSS patterns is
sufficient. WSS was however underestimated when calculated using PC MRI due to the limited
spatial resolution of PC MRI. Increasing the spatial resolution of PC MRI measurements may have
been a remedy for reducing underestimation of WSS. However, it would have also decreased SNR
and increased the scan duration dramatically. The future developments in MRI technology may
allow PC MRI measurements in higher spatial resolution within reasonable time frames and, as a
result of this, underestimation of PC MRI-based WSS may reduce in future studies. Despite this
limitation, PC MRI-based WSS calculations were completed within shorter time frames ($\sim 15$ min
versus $\sim 15$ h per artery) by using a simpler method that is easily applicable to the acquired MR
images. These advantages may make 3D cine PC MRI data an attractive candidate for calculating
WSS magnitudes in clinical practice in future. It should be noted that in the low WSS regions
we found large angles between the CFD- and PC MRI-based WSS vectors. In the carotid arteries,
particular locations such as the carotid bulb are it should be noted that in the low WSS regions
we found large angles between the CFD and PC MRI-based WSS vectors. In the carotid arteries,
particular locations such as carotid bulb are exposed to low and also oscillatory WSS. Some studies
have suggested that oscillatory shear can induce proatherogenic effects on endothelial cells (32). It
may therefore also be necessary to obtain accurate WSS directions in the regions of low WSS, and
this may limit the use of PC MRI-based WSS calculations. The use of a variable encoding velocity over the heart cycle (33) or the use of a dual encoding velocity in combination with accelerated imaging (34) may also improve PC MRI measurements in regions of low velocity, such that the direction of WSS vectors is decently monitored and can also be used to estimate the degree of changes over the cardiac cycle.

5.5 Conclusion

We showed that PC MRI-based WSS magnitudes are lower than those based on CFD. This is mainly due to the limited spatial resolution of PC MRI measurements. However, we observed good agreement between high and low WSS\textsubscript{CFD} and WSS\textsubscript{MRI} patterns and also between the directions of the WSS vectors in the high WSS regions. Although the PC MRI-based WSS calculation method has some limitations, it has the potential to be applied in the clinical assessment of WSS in carotid arteries since it is simpler and easily applicable to the acquired images compared to the current reference standard CFD.

Bibliography


