Phase contrast MRI in intracranial aneurysms
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Wall shear stress in an in vitro and in vivo intracranial aneurysm estimated with phase contrast MRI

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chapter 8

Abstract

Objective
The objective was to study the performance of an in-house developed wall shear stress (WSS) algorithm using PC-MRI in intracranial aneurysms.

Materials & Methods
First, the algorithm was applied to a high resolution in vitro PC-MRI measurement under steady and pulsatile flow conditions. A CFD simulation was performed with similar inflow boundary conditions. Second, WSS was estimated in steady PC-MRI data acquired at different resolutions. Third, the algorithm was applied to a pulsatile in vivo measurement and compared with CFD. The direction and magnitude of WSS vectors were computed and compared with Spearman’s correlation coefficient.

Results
Quantitative agreement was moderate for the phantom (Spearman ρ=0.69). The WSS magnitude of PC-MRI was lower than CFD for both the in vitro and in vivo case. However, there was qualitative agreement between PC-MRI and CFD, i.e. WSS vector direction was similar for both modalities.
Circular WSS patterns were found both in vitro and in vivo for PC-MRI and CFD. Increasing resolution uncovered complex WSS patterns with higher mean WSS magnitude.

**Conclusion**

The algorithm can robustly estimate wall shear stress patterns in aneurysm geometries with similar directions as CFD. PC-MRI based estimation of abnormal WSS patterns may aid in the identification of risk factors for aneurysm disease progression.
8.1 Introduction

Intracranial aneurysms occur in 5 percent of the population and lead to high morbidity and mortality when ruptured or during surgical or endovascular repair [1]. Since risk of complications during treatment may outweigh the risk of rupture of the aneurysm [2], the latter needs to be estimated as accurate as possible when making treatment decisions. Hemodynamic parameters can significantly contribute to the accuracy of rupture risk assessment of intracranial aneurysms [3].

It is widely believed that wall shear stress patterns strongly influence plaque and aneurysm formation, progression and rupture. Wall shear stress is the tangential force that flowing blood exerts on the vessel wall. High wall shear stress is believed to promote atheroprotective endothelial gene expression, whereas low and oscillating wall shear stress in regions with disturbed flow induce atherogenic behaviour of endothelial cells [4]. To derive wall shear stress, sufficiently detailed information on velocity gradients close to the vessel wall is needed, which up until recently could only be provided by computational fluid dynamics (CFD) [5–6]. However, mesh creation and prescription of boundary conditions can be strenuous and the complexity of CFD calculations poses high demands on CPU systems and requires long computational times. Moreover, the reliability of CFD depends on the accuracy of the geometry and inflow boundary conditions.

The last few years much effort has been put into quantifying wall shear stress on the basis of non-invasively measured velocity data acquired with phase contrast MRI (PC-MRI) [7–8], a technique that is becoming increasingly accurate with the recent technical advancements in MRI. As described in the literature, several methods for estimation of wall shear stress were developed and applied to a range of vessel structures.

Oshinski et al. [9] were the first to estimate wall shear stress in the aorta with through-plane PC-MRI data. Stokholm et al. [10] estimated 1D wall shear stress and oscillatory shear index by parabolic fitting to through-plane velocity profiles in the carotid artery. Papathanasopoulou et al. [11] accounted for secondary flow in the carotid artery and bifurcation by measuring velocities with three-dimensional PC-MRI and estimated three-dimensional wall shear stress vectors. Stalder et al. [12] and Markl et al. [13] improved this method in the aorta by using b-splines for fitting purposes. However, they needed to manually select slices perpendicular to the aorta, prior to their calculations. Bieging et al. used a similar method in ascending aorta
dilation [14]. Boussel et al. [15] were the first to estimate wall shear stress in intracranial aneurysms based on PC-MRI data acquired at a resolution of 1 mm$^3$, followed by Isoda et al. [16] using a similar resolution. Petersson et al. [17] showed with the use of numerical simulations that wall shear stress calculation by parabolic fitting produced the most accurate results. Recently, a wall shear stress estimation algorithm was developed that can be applied in 3D in any vessel geometry (Chapter 7).

The purpose of the current study was to study the performance of this algorithm in intracranial aneurysms. The study consisted of three parts. First, wall shear stress in an intracranial aneurysm phantom under controlled steady and pulsatile flow [18] was estimated and compared to CFD estimates. Second, the effect of resolution of PC-MRI measurements on wall shear stress accuracy was studied. Third, wall shear stress estimates obtained from an in vivo PC-MRI measurement and CFD simulation were compared.

### 8.2 Materials & Methods

In the first experiment, wall shear stress estimated from high resolution PC-MRI was compared with wall shear stress calculated by CFD. In order to compare the two modalities, the wall shear stress was calculated with the use of velocity vectors obtained from PC-MRI and wall delineation obtained from 3D Rotational Angiography (3D-RA), which was the modality that was used for the creation of the CFD mesh. In the second experiment, wall shear stress estimated from PC-MRI acquired at multiple resolutions was compared. The wall segmentation was based on the PC-MRI magnitude data of the resolution under consideration. In the third experiment, wall shear stress was estimated from PC-MRI acquired in an in vivo aneurysm and compared with CFD. In this experiment the wall was segmented from the magnitude data of the PC-MRI dataset. These experiments are summarized in table 8.1. Throughout the article, the vessel and aneurysmal wall are considered as the outermost part of the segmentation of the lumen.

<table>
<thead>
<tr>
<th>Experiment</th>
<th>Resolution PC-MRI</th>
<th>Wall</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experiment 1</td>
<td>0.2 x 0.33 x 0.2</td>
<td>3D-RA</td>
</tr>
<tr>
<td>Experiment 2</td>
<td>0.19-0.94 isotropic</td>
<td>PC-MRI</td>
</tr>
<tr>
<td>Experiment 3</td>
<td>0.78 x 0.78 x 0.80</td>
<td>PC-MRI</td>
</tr>
</tbody>
</table>

Wall shear stress in an in vitro and in vivo intracranial aneurysm estimated with phase contrast MRI
8.2.1 Aneurysm phantom & flow loop set-up

A glass reproduction of an aneurysm located in the anterior communicating artery of a patient who supplied informed consent was manually created based on a 3D-RA dataset. The dimensions of the aneurysmal lumen were 6 mm x 4 mm x 9 mm in the x, y, and z-directions (see figure 8.1a) respectively. The phantom was submerged in agar gel and connected to a pump supplying constant flow and a computer-controlled piston pump supplying pulsatile flow. In figure 8.1b the flow loop setup is displayed, in figure 8.1c the velocity profile that served as input for the pulsatile measurement. For all scans and simulations water was used. Further details are described in [18].

8.2.2 Experiment 1. In vitro wall shear stress from steady and pulsatile PC-MRI compared with CFD

PC-MRI

All PC-MRI measurements were performed on a 3T MR system (Philips Healthcare, Best, the Netherlands) in a solenoid rat coil (Philips, Hamburg, Germany) with a diameter of 7 cm. Steady and retrospectively gated pulsatile flow measurements were performed at a spatial resolution of 0.2 mm x 0.33 mm x 0.2 mm and velocity encoding of 50 x 100 x 50 cm/s in x, y, and z-direction respectively (see figure 8.1a). Other imaging parameters: field of view: 25 mm x 16.5 mm x 25 mm; flip angle: 15°; TE/TR: 3.9 / 11.1 ms. Scan time of the steady measurement was approximately 15 minutes. The temporal resolution of the pulsatile PC-MRI measurement was 150 ms in a cardiac cycle of 3 s, resulting in 20 measured cardiac phases. Scan time was approximately 3 hours. More detail can be found in [18] or Chapter 2.
The geometric vascular model of the in vitro aneurysm phantom was obtained by filling the phantom with a contrast agent and performing 3D Rotational Angiography (3D-RA). The 3D-RA dataset was segmented and meshed in VMTK [19]. The wall of this 3D-RA mesh was used as wall delineation for experiment 1. The mesh consisted of 742,316 tetrahedral cells with a mesh density of 3119 elements per cubic millimeter. The simulations were performed in FLUENT (Ansys, Canonsburg, PA, USA) using boundary conditions derived from the PC-MRI measurements for both constant and pulsatile flow. More detail can be found in [18] or Chapter 2.

8.2.3 Experiment 2. In vitro wall shear stress from steady PC-MRI at increasing resolutions

The PC-MRI measurements at varying spatial resolutions were performed in a different scanning session. The resolutions, TE/TR and scan times are listed in Table 8.2. Further imaging parameters were: field of view: 60 mm x 21 mm x 60 mm and velocity encoding of 30 x 60 x 30 cm/s in x, y and z-direction respectively.

<table>
<thead>
<tr>
<th>Voxel size (mm x mm x mm)</th>
<th>TE / TR (ms)</th>
<th>Scan time (min.s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.94 x 0.94 x 1</td>
<td>2.8 / 4.6</td>
<td>0.19</td>
</tr>
<tr>
<td>0.75 x 0.75 x 0.8</td>
<td>2.95 / 4.9</td>
<td>0.29</td>
</tr>
<tr>
<td>0.62 x 0.62 x 0.7</td>
<td>3.0 / 5.2</td>
<td>0.41</td>
</tr>
<tr>
<td>0.54 x 0.54 x 0.6</td>
<td>3.2 / 5.7</td>
<td>1.06</td>
</tr>
<tr>
<td>0.47 x 0.47 x 0.5</td>
<td>3.4 / 6.3</td>
<td>2.07</td>
</tr>
<tr>
<td>0.38 x 0.38 x 0.4</td>
<td>3.7 / 7.1</td>
<td>4.20</td>
</tr>
<tr>
<td>0.27 x 0.27 x 0.3</td>
<td>4.3 / 8.7</td>
<td>11.35</td>
</tr>
<tr>
<td>0.19 x 0.19 x 0.2</td>
<td>5.4 / 11.7</td>
<td>36.33</td>
</tr>
</tbody>
</table>

8.2.4 Experiment 3. In vivo wall shear stress from pulsatile PC-MRI compared with CFD

The protocol consisted of two MRI sequences that were conducted on a 3T scanner (Intera, Philips Healthcare, Best, The Netherlands) using an 8-channel head coil.

First, to acquire 2D PC-MRI data that served as inflow boundary conditions for CFD, velocity was measured in three directions in a slice placed perpendicular to the vessel proximal to the aneurysm. Scan resolution was
0.64 mm x 0.65 mm x 3 mm. Further imaging parameters: TE / TR / FA: 5.7 ms / 8.5 ms / 10°; Field of view: 200 mm x 200 mm x 3 mm in one slice; parallel imaging factor: 2; VENC: 100 cm/s in all directions. The number of measured cardiac phases was 36 cardiac phases, resulting in a temporal resolution of 26 ms. Scan time was approximately 3 minutes and 30 seconds. The view sharing factor for the retrospective sorting of acquired \( k \)-lines was set to 1.8 [20].

Second, PC-MRI was acquired at a resolution of 0.8 mm x 0.8 mm x 0.8 mm. Further imaging parameters were: TE / TR / FA: 3.0 ms / 5.8 ms / 15°; Field of view: 200 mm x 200 mm x 20 mm in 25 transversal slices; parallel imaging factor of 3; the velocity encoding was 100 cm/s in all directions; scan time was approximately 10 minutes and 20 seconds. The number of acquired cardiac phases was 10 resulting in a temporal resolution of 90 ms. The 2D and 3D PC-MRI acquisition were retrospectively gated using PPU.

**CFD**

The geometric vascular model used for the in vivo CFD simulation was created from 3D-RA images. The mesh consisted of 1,422,476 tetrahedral elements with a mesh density of 5230 elements per cubic millimeter. The CFD simulations were performed in FLUENT 6.3 (Ansys, Canonsburg, PA, USA). Blood density was set to 1060 kg/m³, dynamic viscosity to 0.004 kg/m·s. The 2D PC-MRI data was positioned on the TOF data using rotation and translation matrices extracted from DICOM headers. A rigid registration of the TOF measurement on the CFD mesh was conducted in FLIRT [21]. The velocities measured with 2D PC-MRI were rotated and translated likewise and interpolated onto the nodes of the CFD inflow boundary. These steps were performed with custom-built software in Matlab (Mathworks, Natick, MA, USA). CFD iterations were continued until the residual of the continuity equation was below 0.001. The CFD estimates were resolved at fixed time intervals equal to the measured RR interval divided by the number of cardiac phases used for the 2D PC-MRI. Three heart cycles were simulated to eliminate transient effects. The third of these cycles was used to compare the calculated wall shear stress with the wall shear stress calculated from the PC-MRI results. Flow through the outflow vessels of the CFD model was prescribed according to outflow measurements at every cardiac phase of the...
PC-MRI data averaged over time. The simulation time was approximately 36 hours.

8.2.5 Postprocessing

PC-MRI background correction for phase offset errors was performed for every slice and, in case of pulsatile measurements, for each individual cardiac phase by subtracting the mean velocity in the stationary agar gel or brain tissue (amygdala). Wall delineation of the phantom and in vivo aneurysm was defined by a level set evolution algorithm [22], applied to the magnitude images of the PC-MRI measurements. SNR of the phase contrast magnitude images of the pulsatile in vitro and in vivo measurements were calculated according to Price et al. [23]. For these processes custom-built software was developed in Matlab (Mathworks, Natick, MA, USA).

The rigid registration between the PC-MRI data and 3D-RA wall delineation (experiment 1) was automatically performed in FLIRT (FMRIB’s Linear Image Registration Tool, FSL). After registration, PC-MRI voxels located outside the 3D-RA wall delineation were discarded.

8.2.6 Wall shear stress calculation

Wall shear stress vectors can be calculated by:

$$\tau = 2\mu (\dot{\varepsilon} \cdot \vec{n})$$

(8.1)

where $\tau$ is the wall shear stress vector, $\mu$ is the dynamic viscosity, $\dot{\varepsilon}$ is the rate of deformation tensor, and $\vec{n}$ is the normal vector.

By rotating the axes system such that the z-axis aligns with the normal vector of the vessel wall it holds that: $\vec{n} = (0,0,1)$. Combined with the assumption that no flow occurs through the wall, $\vec{n} \cdot \vec{v} = 0$ at the wall, the inner product of the rate of deformation tensor and the normal vector is reduced to:

$$2\dot{\varepsilon} \cdot \vec{n} = \left(\frac{\partial \dot{v}_x}{\partial x}, \frac{\partial \dot{v}_y}{\partial y}, 0\right)$$

(8.2)

The shear rates $\frac{\partial \dot{v}_x}{\partial x}$ and $\frac{\partial \dot{v}_y}{\partial y}$ are the spatial gradients at the wall of 1D smoothing splines [24] fitted through the rotated x-, and y-velocity values.
in the direction of the normal. The rotated wall shear stress vector $\vec{\tau}'$ is then defined as:

$$
\vec{\tau}' = \mu \frac{\partial v'}{\partial z}, \quad \vec{\tau}' = \mu \frac{\partial v'}{\partial z}, \quad \vec{\tau}' = 0
$$

(8.3)

The length of the inward normal vector was 0.6 mm. Measured velocity values surrounding the inward normal were interpolated such that the spline was fitted through 3 velocity values. To obtain a smooth surface of the aneurysm wall, the segmentation obtained by postprocessing is smoothed using a Laplacian filter [25]. See Chapter 7 for further detail.

8.2.7 Data quantification and visualization

In experiment 1, mean and standard deviation of the wall shear stress magnitude values are calculated and plotted. The mean and standard deviations of the paired differences are given. Furthermore, linear regression is performed on the PC-MRI and CFD data and the Spearman correlation $\rho$ is calculated. Statistical comparison was done for paired groups (Wilcoxon-signed rank test) as differences were not normally distributed. The difference in direction between wall shear stress vectors is quantified in terms of the angle between corresponding wall shear stress vectors and expressed as the median of the angle distribution. Statistical analysis was performed in the total phantom and in the in-, and outflow vessels and phantom aneurysm separately.

In experiment 2 and 3, the wall shear stress magnitude values are expressed in terms of mean wall shear stress and the standard deviation.

All postprocessing and visualization was performed with in-house built software in Matlab (Mathworks, Natick, MA, USA).

8.3 Results

8.3.1 Experiment 1. In vitro wall shear stress from steady and pulsatile 3D PC-MRI compared with CFD

The SNR of the pulsatile in vivo PC-MRI measurements was 28. In figure 8.2, the wall shear stress patterns calculated from PC-MRI with 3D-RA wall delineation are shown for steady flow (8.2a) and for systole (8.2c) and diastole (8.2e) under pulsatile flow. Figures 8.2b,
d and f show the corresponding estimates from CFD using the same wall segmentation. For both PC-MRI and CFD, the wall shear stress patterns spreads upward and left and right (“star-like”) in the region where the flow impacts the wall (arrow 1). Fluctuations of wall shear stress of PC-MRI are visible. The magnitude of the wall shear stress vectors in the dome and bleb of the phantom was approximately twice as low for PC-MRI than CFD. In the inflow and outflow vessels the magnitude of the wall shear stress vectors was similar for PC-MRI and CFD. Note that the small circular wall shear stress pattern in the tip of the aneurysm was resolved for both PC-MRI and CFD (arrow 2).

In figure 8.3 the spatially averaged wall shear stress and standard deviation over time of the PC-MRI measurements and CFD simulations is shown. The mean and standard deviation of the wall shear stress are lower for the PC-MRI measurement than for the CFD simulation, specifically in systole.
Differences between the wall shear stress obtained from PC-MRI with 3D-RA wall delineation and CFD are quantified and summarized in Table 8.3. Due to the complexity of flow in the aneurysm, for the steady and systolic wall shear stress, the estimation of wall shear stress in the aneurysm is worse than in the inflow and outflow vessels, i.e. the slope of the regression line is closer to 1 if only the inflow and outflow vessels are considered. Due to the lower wall shear stress for PC-MRI compared to CFD, moderate quantitative agreement was found, as can be appreciated from the Spearman correlation in Table 8.3 and the correlation and Bland-Altman plots for the total phantom in Figure 8.4.

Figure 8.3 Mean and standard deviation of wall shear stress for the in vitro pulsatile PC-MRI measurement and CFD simulation.

Table 8.3 Mean paired difference ± standard deviation of the paired difference of the wall shear stress magnitude as calculated from the steady and pulsatile PC-MRI measurements and CFD simulations for the inlets and outlets, the aneurysm and the total phantom. The median angle describes the difference in velocity vector direction and $p_1$ and $p_2$ represent the slope and intercept of the linear regression analysis. The Spearman correlation coefficient is given as well.

<table>
<thead>
<tr>
<th>Inlet/outlet</th>
<th>Steady</th>
<th>Systole</th>
<th>Diastole</th>
<th>Steady</th>
<th>Systole</th>
<th>Diastole</th>
<th>Steady</th>
<th>Systole</th>
<th>Diastole</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean/SD WSS (Pa)</td>
<td>0.21±0.30</td>
<td>0.28±0.48</td>
<td>0.03±0.11</td>
<td>0.38±0.74</td>
<td>0.66±1.15</td>
<td>0.03±0.18</td>
<td>0.29±0.59</td>
<td>0.66±1.15</td>
<td>0.03±0.14</td>
</tr>
<tr>
<td>Median angle (°)</td>
<td>11.9</td>
<td>11.1</td>
<td>9.1</td>
<td>24.0</td>
<td>25.8</td>
<td>20.9</td>
<td>16.5</td>
<td>15.8</td>
<td>13.3</td>
</tr>
<tr>
<td>$p_1/p_2$</td>
<td>0.93 / 0.25</td>
<td>0.92 / 0.35</td>
<td>0.74 / 0.09</td>
<td>1.98 / 0.13</td>
<td>1.58 / 0.30</td>
<td>1.19 / 0.00</td>
<td>1.14 / 0.21</td>
<td>1.10 / 0.38</td>
<td>0.92 / 0.05</td>
</tr>
<tr>
<td>Spearman ρ</td>
<td>0.68</td>
<td>0.64</td>
<td>0.70</td>
<td>0.67</td>
<td>0.66</td>
<td>0.61</td>
<td>0.69</td>
<td>0.65</td>
<td>0.74</td>
</tr>
</tbody>
</table>
8.3.2 Experiment 2. In vitro wall shear stress from steady 3D PC-MRI at increasing resolutions

In figure 8.5 the wall shear stress vectors calculated for the steady PC-MRI measurements at four different resolutions are shown. The wall delineation was obtained from the individual measurements, becoming increasingly coarse. The complexity of wall shear stress vectors diminished with coarser resolution in the region of impact of the flow with the wall, as pointed out by arrow 1. Furthermore, the maximum wall shear stress diminished with decreasing resolution. Note, however, that the circular wall shear stress patterns in the tip of the phantom could be resolved at all resolutions, even with decreased segmentation accuracy at lower resolutions. Higher mean wall shear stress at higher resolution was found. This is graphically displayed in figure 8.6.
8.3.3 Experiment 3. In vivo wall shear stress from pulsatile PC-MRI compared with CFD

The SNR of the in vivo PC-MRI measurements was 15. Figure 8.7 depicts wall shear stress for the in vivo aneurysm. Arrow 1 indicates circular wall shear stress at the side of the aneurysm which was resolved for PC-MRI as well as CFD. Elevated wall shear stress at the top of the aneurysm (arrow 2) was also seen in both methods. Wall shear stress was around two to three times lower for the PC-MRI measurement than in the CFD simula-
tion, while measured and simulated velocities were similar (34.7±19.1 cm/s and 16.2±9.5 cm/s at systole and diastole for PC-MRI respectively versus 35.2±25.0 cm/s and 15.5±11.5 cm/s at systole and diastole for CFD respectively). Mean wall shear stress over time is displayed in figure 8.8.

Figure 8.7 Wall shear stress vectors calculated in the in vivo PC-MRI data (left column) and CFD (right column) at systole (top row) and diastole (bottom row).

Figure 8.8 Mean and standard deviation of wall shear stress for the in vivo PC-MRI measurement and CFD simulation.
8.4 Discussion

In this study a wall shear stress calculation algorithm was applied to PC-MRI data measured in an in vitro and an in vivo aneurysm. In this work we considered wall shear stress vectors, considering both direction and magnitude of the vectors. We demonstrated that in both the phantom and in vivo, wall shear stress patterns are qualitatively similar to their CFD predictions. It has been discussed in the literature \cite{12, 26} that wall shear stress estimations improve with increasing resolution (Chapter 7). This is the first study that shows that wall shear stress estimations show higher values and more detail in an intracranial aneurysm phantom by measuring velocity at increasing resolutions.

The wall shear stress magnitude estimated by PC-MRI was lower than the CFD based values, which can be attributed to the lower spatial resolution for PC-MRI than CFD. The difference in mean wall shear stress found for in vivo PC-MRI and CFD was similar to the mean wall shear stress difference found for in vitro PC-MRI at 0.75 mm and 0.19 mm. Imperfect registration of the high resolution 3D-RA wall delineation to the lower and anisotropic resolution PC-MRI grid may have introduced further discrepancies between wall shear stress profiles in experiment 1.

The quantitative agreement found in the in vivo aneurysm in this study was better than in Boussel et al. \cite{15}, who remarkably found 6-12 fold higher maximum wall shear stress for PC-MRI than CFD in three in vivo aneurysms. Apart from higher spatial resolution in our in vivo case, the spline-fitting nature of our wall shear stress algorithm may provide more robust results than their method. One other study comparing heavily interpolated PC-MRI data with CFD \cite{27} found a low to moderate degree of correlation. Neither study visualized the wall shear stress vectors nor provided any information on the agreement in vector direction.

Disturbed and laminar flow provoke opposite biological reactions from endothelial cells \cite{28}. Wall shear stress patterns therefore play a possible role in rupture, forming the rationale for the current study. The exact link between spatial and temporal shear patterns and rupture remains subject of discussion. Identifying such a link in any case requires proper wall shear stress estimation. However, it is reassuring to see that presumably important wall shear stress patterns, such as elevated or circular wall shear stress, were similar for low and high resolution PC-MRI and CFD. This hints to the possibility of clinical use of wall shear stress estimation from PC-MRI.
data. Wall shear stress vectors calculated from PC-MRI have been visualized in the aorta [12] and the carotid bifurcation [11]. While those studies show complex spatial and temporal differences in direction, in the current intracranial aneurysm circular wall stress patterns are observed, resulting from ‘tornado-like’ vortices, i.e. with the vortex axis perpendicular to the wall. To our knowledge, this is the first study to show such circular wall shear stress behavior in the cardiovascular system. It remains to be addressed what the consequences are of such streaming for endothelial biology and rupture risk. These structures can only be detected if wall shear stress vectors are derived and will go undetected if merely wall shear stress magnitudes are visualized. Circular wall shear stress behavior may be added to existing factors in the statistical analysis of ruptured versus unruptured aneurysms such as disturbed or stable flow patterns, small or large impingement regions and narrow or wide inflow jets [3].

This study underlines that the main concern of wall shear stress estimations based on PC-MRI is the limited spatial resolution of PC-MRI. A higher SNR may be beneficial for wall shear stress estimations as well. With recent technical advancements, scanners with higher field strengths are becoming rapidly available. The first studies that present PC-MRI data with higher SNR and resolution are now being published [29]. It is clear that this approach will be beneficial for the accuracy of wall shear stress estimations from PC-MRI. One recently presented new approach to improve measured PC-MRI data is divergence reduction [30]. A requirement in wall shear stress estimations is the accurate acquisition of the velocity close to the wall. Due to the nature of blood flow, this velocity is low. In PC-MRI low velocities are difficult to resolve and noise on low velocity values will be more prominent. The divergence reduction technique may therefore be able to improve wall shear stress estimations significantly.

8.5 Conclusion

In conclusion, this study shows that the wall shear stress algorithm is capable of calculating wall shear stress vectors from PC-MRI data in an in vitro and in vivo intracranial aneurysm. The direction of the wall shear stress vectors was similar to the wall shear stress vectors simulated with CFD, both in vitro and in vivo. Qualitative agreement between PC-MRI and CFD was good, whereas quantitative agreement was moderate. Further-
more, in order to increase the accuracy of estimated wall shear stress values, the spatial resolution of PC-MRI measurements must be as high as possible. However, important wall shear stress vector patterns, such as circular wall shear stress and regions of high versus low wall shear stress can be resolved at lower resolutions.

8.6 Acknowledgments

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