Phase contrast MRI in intracranial aneurysms
van Ooij, P.

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Quantification and visualization of flow in the Circle of Willis: time-resolved three-dimensional phase contrast MRI at 7T compared with 3T

Pim van Ooij, Jaco Zwanenburg, Fredy Visser, Charles Majoe, Ed van Bavel, Jeroen Hendrikse, Aart Nederveen

Abstract

The assessment of both geometry and hemodynamics of the intracranial arteries has important diagnostic value in internal carotid occlusion, sickle cell disease and aneurysm development. Provided that SNR and resolution are high, these factors can be measured with time-resolved three-dimensional phase contrast MRI (PC-MRI). However, within a given scan time duration, an increase in resolution causes a decrease in SNR and vice-versa, hampering flow quantification and visualization. To study the benefits of higher SNR at 7T, PC-MRI in the Circle of Willis was performed at 3T and 7T in 5 volunteers. Results showed that the SNR at 7T was roughly 2.6 times higher than at 3T. Therefore, segmentation of small vessels such as the anterior and posterior communicating arteries (ACoA and PCoA) succeeded more frequently at 7T. Direction of flow and smoothness of streamlines in the ACoA and PCoA were more pronounced at 7T. Mean velocity magnitude values in the vessels of the Circle of Willis were higher at 3T due to noise compared to 7T. Likewise, areas of the vessels were lower at 3T. In conclusion, the gain in SNR at 7T compared to 3T allows for improved flow visualization and quantification in intracranial arteries.
6.1 Introduction

The Circle of Willis acts as an essential collateral pathway to maintain blood flow to the cerebral cortex in case of vessel occlusion due to thrombosis, atherosclerosis or vasospasm \[1\]. Knowledge of local hemodynamics and geometry of the Circle of Willis is beneficial in diagnosis, treatment, or screening of a number of pathologies. First, it may be beneficial in treatment planning in patients with symptomatic internal carotid artery occlusion \[2\]. Secondly, overt cerebral infarcts are common in children that suffer from sickle cell disease. The elevated blood flow in the internal carotid and cerebral arteries can be used as a screening method for infarct risk assessment \[3\]. Furthermore, intracranial aneurysms, a common cause for subarachnoid haemorrhage or cerebral stroke, often develop at arterial branch points in the Circle of Willis \[4-5\]. One risk factor for aneurysm development is the asymmetry and incompleteness of the Circle of Willis \[6-8\].

Studies attempting to understand the mentioned pathologies and its relation to the anatomy of and hemodynamics in the Circle of Willis are based on transcranial Doppler ultrasonography (TCD) \[9\], morphometric \[6, 10-11\], or computational analysis \[12-14\]. A disadvantage of TCD is its incapability to access the vessels higher in the brain and the acquisition of merely the maximum velocity magnitude in a vessel \[15\]. Morphometric studies lack flow information whereas computational studies suffer from various assumptions such as Newtonian fluid behaviour, rigid vessel walls and non-patient-specific boundary conditions \[16\].

A promising technique to non-invasively measure blood flow is time-resolved three-dimensional phase contrast MRI (PC-MRI) \[17-18\]. PC-MRI can combine anatomical information with spatial and temporal blood flow velocity information and the derived hemodynamic properties in the Circle of Willis, provided that the resolution of the measurement is high (voxel size < 1 x 1 x 1 mm) \[19-21\]. These flow measurements have successfully been applied to measure redistribution of blood flow after Internal Carotid Artery (ICA) occlusion \[22-25\] or to measure elevated blood flow in sickle cell disease \[3\]. The feasibility of characterizing blood flow patterns in intracranial aneurysms has been proven successful as well \[26-30\].

PC-MRI measurements may prove particularly valuable in the cases of the small communicating arteries in the Circle of Willis, since redistribution of flow in the Circle of Willis occurs through these arteries and since the instability and complexity of the flow may explain the incidence of aneurysms.
at these sites [31-32]. However, segmentation of these small vessels may be hampered by insufficient resolution of the PC-MRI measurement. Furthermore, image quality may be compromised in high resolution acquisitions since SNR decreases with increasing resolution. A decrease in SNR of PC-MRI data increases segmentation failure, blood-flow direction uncertainty and hampers flow quantification [21]. The application of parallel imaging [33] to decrease long scan times inherently related to PC-MRI, decreases SNR even further.

A straightforward option to increase SNR while maintaining sufficient spatial resolution would be to perform acquisitions at higher field strengths. To our knowledge, so far no blood flow velocity measurements with the use of time-resolved PC-MRI have been conducted at 7T. We hypothesize that at 7T the SNR is superior to 3T and that more detailed flow information can be gathered. In this study time-resolved PC-MRI is performed in the Circle of Willis of five volunteers at 3T and 7T at two different resolutions, to investigate the benefit of 7T in quantifying blood flow in the vessels of the Circle of Willis.

### 6.2 Materials & Methods

#### 6.2.1 Volunteers

MR examinations were performed on 5 subjects (2 males, 3 females) without any known history of cardiovascular abnormalities or neurological symptoms. Age varied between 21 and 55 years old. Written informed consent was given by all volunteers in accordance to the Institutional Review Board of the University Medical Center Utrecht.

#### 6.2.2 MR imaging procedure

Examinations were performed using PC-MRI based on a spoiled gradient echo with standard four point encoding [34] on a 3T MR system (Achieva, Philips Healthcare, Best, The Netherlands) in an 8-channel head coil (with a width of 220 mm diameter) and a 7T MR system (Achieva, Philips Healthcare, Cleveland, USA) in a volume transmit and 16-channel receive head coil (with a width of 180 mm diameter, Nova Medical, Wilmington, MA, USA). The examinations were retrospectively gated [35]. A specific $k$-line was repeatedly acquired during a complete RR-interval; this was repeated for the next $k$-line until all $k$-lines were acquired.
The temporal interpolation factor for the retrospective sharing between successive heart phases of acquired $k$-lines was set to 2. Gating information was acquired with the use of a peripheral pulse unit (PPU). PC-MRI was performed at two (non-interpolated) resolutions: 0.47 mm x 0.47 mm x 0.5 mm, referred to in the remainder of this paper as 0.5 mm, and 0.75 mm x 0.75 mm x 0.8 mm, referred to in the remainder of this paper as 0.8 mm. Imaging parameters for both sequences were: flip angle: 20°; field of view: 180 mm x 180 mm x 20 mm (Anterior-Posterior x Right-Left x Feet-Head); velocity encoding: 150 cm/s x 150 cm/s x 150 cm/s; SENSE: 3 (in RL direction); TE/TR at 0.5 mm: 4.1/8.6 ms; TE/TR at 0.8 mm: 3.9/7.8 ms. Number of slices at 0.5 mm: 40; Number of slices at 0.8 mm: 25. In each volunteer the temporal resolution was kept constant by means of adjusting the measured number of cardiac phases in the heart cycle. For acquisitions at 0.5 mm the number of reconstructed cardiac phases ranged from 5 to 11 cardiac phases at a heart rate of 85 to 41 beats per minute respectively, resulting in a temporal resolution of 147±7.0 ms. For acquisitions at 0.8 mm the number of reconstructed cardiac phases ranged from 10 to 24 cardiac phases at a heart rate of 86 to 40 beats per minute respectively, resulting in a temporal resolution of 69±3.6 ms. For both resolutions, scan times ranged from 17 to 9 minutes depending on the heart rate. Time in between the 3T and 7T scans was no longer than 60 minutes.

6.2.3 Data quantification and visualization

Phase images were corrected for background phase offset errors by subtraction of the average phase in a nearby static region of interest (amygdala). Phase correction was performed for every velocity encoding direction and cardiac phase individually [36]. The lumen of the Circle of Willis and its surrounding vessels was segmented semi-automatically with the use of a level set evolution algorithm [37] applied to the sum of the magnitude images derived from the complex phase contrast data, referred to in the remainder of this paper as the phase contrast magnitude images, and the complex difference reconstruction images [38]. The lumen was segmented at all cardiac phases and in every slice using the same algorithm. When the majority of cardiac phases showed a full segmentation of a certain vessel, the vessel was regarded as segmented. False segmentations due to cerebrospinal fluid or noise were manually removed. To remove outliers in the velocity vector fields, each velocity direction (RL,
AP and FH) of the phase contrast data was filtered with a custom-built filter, which used the maximum observed velocity difference between subsequent cardiac phases to define outliers. In pixels that showed differences higher than 30% of the maximum velocity difference, the filter prescribed the mean velocity of the non-zero surrounding pixels. This was repeated for pixels that now showed differences higher than 60% of the maximum velocity difference between subsequent cardiac phases.

To quantify differences between PC-MRI at 3T and 7T, the 7T phase contrast magnitude data was registered to the 3T phase contrast magnitude data using affine registration in FLIRT [39]. The means of the paired differences between the 3T and 7T PC-MRI acquisitions were determined by subtracting the velocity magnitude in corresponding voxels in every cardiac phase and subsequently averaging this difference over the number of voxels and over the number of cardiac phases. Standard deviations of the paired differences (SDp) and the median of the difference in angles of velocity vectors were calculated. Significance of the mean of the paired differences was tested with a t-test; p<0.05 was considered statistically significantly different.

Signal to noise ratios were calculated according to Price et al. [40]. Let $S_i$ and $S_j$ represent phase contrast magnitude signals in a region of interest selected in static tissue (amygdala) during different cardiac phases of similar mean velocity magnitude. By subtracting these images, an image containing minimum signal and maximum noise is obtained. SNR is estimated from [41]:

$$\text{SNR} = \frac{\text{mean}(S_i + S_j)}{\sqrt{2 \text{ std}(S_i - S_j)}}$$

Since the SNR of the phase images is proportional to the SNR of the phase contrast magnitude images [42], SNR estimations were not performed in the phase images separately. These post-processing and quantification steps were performed with custom-built software in Matlab (Mathworks, Natick, MA, USA). The postprocessed data were subsequently imported in GTFlow (Gyrotools, Zurich, Switzerland) to perform blood flow visualization and quantification in the arteries in the Circle of Willis. A perpendicular slice was manually positioned in the artery of interest and all segmented pixels were included in the mean velocity magnitude, area and flow calculation. Note that flow was calculated by multiplication of the area with the average through-plane velocity. The latter value is not given. For all vessels that were
found at both 3T and 7T, a t-test was performed to investigate whether the differences between mean velocity magnitude, area and flow at 3T and 7T were significant. Flow patterns were visualized by vector and streamline fields. Theoretically, streamlines are disrupted when divergence of the local velocities is high or smooth when divergence is low.

### 6.3 Results

In table 6.1 the segmentations of the Posterior Communicating Arteries (PCoA) and Anterior Communicating Arteries (ACoA) at the four different PC-MRI sequences are shown. Most communicating arteries were segmented at 0.5 mm resolution at 7T and none at 0.8 mm at 3T. Note that the left PCoA of volunteer 5 was segmented at 0.5 mm at 3T, but not at 7T. This was due to movement of the subject, who moved the PCoA outside the field of view at 7T after field of view positioning. The direction of the flow in the small vessels, deduced from the PC-MRI sequence at 0.5 mm at 7T, is added to table 6.1.

<table>
<thead>
<tr>
<th>ACoA</th>
<th>Left PCoA</th>
<th>Right PCoA</th>
</tr>
</thead>
<tbody>
<tr>
<td>7T</td>
<td>3T</td>
<td>7T</td>
</tr>
<tr>
<td>0.5</td>
<td>0.8</td>
<td>0.5</td>
</tr>
<tr>
<td>1</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>2</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>3</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>4</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>5</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 6.1 Circle of Willis configuration of the volunteers as measured by the four PC-MRI acquisitions. Direction of flow is derived from the 0.5 mm measurement at 7T except for volunteer 5.

Table 6.2 provides SNR values and differences between the measured velocities at 3T and 7T. Signal to noise ratios were higher at 7T than at 3T and at 0.8 mm than at 0.5 mm. Averaged over all volunteers, the increase in SNR at 7T was approximately a factor of 2.6 compared to 3T. It is shown that the means of the paired differences were small for both resolutions and all volunteers, albeit significant in most cases. Positive mean differences indicate that the velocities at 3T were higher than at 7T. The standard deviation of the paired difference and the median angle were higher at 0.5 mm than at 0.8 mm.
In figure 6.1a, stacks of segmented phase contrast magnitude images are shown of the Circle of Willis of volunteer 1 at 7T (left) and 3T (right). The ACoA (red) and right PCoA (green) were visible at both field strengths. Note that the neurological convention was used here. Some posterior vessels were more apparent at 7T, whereas the right ophthalmic artery was more apparent at 3T. The former was a result of an increase of SNR at 7T, the latter of field of view placement. In figure 6.1b, velocity vector visualization with a magnified inset of the flow in the ACoA at begin diastole is displayed. At 7T it can be seen that the ACoA had a blood flow direction from the right ACA to the left ACA. This was less obvious at 3T where noise levels were higher than at 7T. Increased noise hampered streamline visualization at 3T compared to 7T, as shown in figure 6.1c. Similar advantages of the higher field strength could be seen in the right PCoA, displayed in figure 6.1d and e. Blood at 7T flowed from PCA to ICA, which was less clear at 3T, and streamlines were slightly more disrupted at 3T.

Another example is displayed in figure 6.2, where, apart from the ACoA (red) and the right PCoA (blue), the left PCoA (green) and both Anterior Choroidal Arteries (AChA, blue and green) of volunteer 2 were segmented. Again, the ACoA blood flow direction was from the right ACA to the left ACA, which was clearer at 7T compared to 3T, see figure 6.2b. At 3T the segmentation of the ACoA at displayed cardiac phase failed. Therefore, disrupted streamlines through the ACoA were seen at 3T in figure 6.2c. It is clear in figure 6.2d that in the left PCoA the blood flow direction was from the PCA to the ICA, which was more difficult to see at 3T. This was supported by the streamline visualization in figure 6.2e. Note in figure 6.2d and e that upward flow in the AChA could be visualized at 7T. In figure 6.2f it can be seen that, due to lower SNR, segmentation of the right PCoA and the right AChA failed.

<table>
<thead>
<tr>
<th></th>
<th>Volunteer 1</th>
<th>Volunteer 2</th>
<th>Volunteer 3</th>
<th>Volunteer 4</th>
<th>Volunteer 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resolution (mm)</td>
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<td>0.8</td>
<td>0.5</td>
<td>0.8</td>
<td>0.5</td>
</tr>
<tr>
<td>Field</td>
<td>3T</td>
<td>7T</td>
<td>3T</td>
<td>7T</td>
<td>3T</td>
</tr>
<tr>
<td>Phases</td>
<td>6</td>
<td>6</td>
<td>13</td>
<td>13</td>
<td>11</td>
</tr>
<tr>
<td>SNR</td>
<td>13</td>
<td>34</td>
<td>41</td>
<td>12</td>
<td>39</td>
</tr>
<tr>
<td>Mean (cm/s)</td>
<td>0.0</td>
<td>0.1*</td>
<td>1.4*</td>
<td>2.1*</td>
<td>0.7*</td>
</tr>
<tr>
<td>SDp (cm/s)</td>
<td>10.0</td>
<td>6.9</td>
<td>16.4</td>
<td>9.9</td>
<td>17.3</td>
</tr>
<tr>
<td>Angle (º)</td>
<td>21.6</td>
<td>13.0</td>
<td>27.6</td>
<td>13.5</td>
<td>26.0</td>
</tr>
</tbody>
</table>

Table 6.2 SNR, mean of the paired difference (3T minus 7T) where * indicates a significant difference (p<0.001), standard deviation of the paired difference (SDp) and the median of the angle distribution between the 3T and 7T PC-MRI measurements.
at 3T, whereas at 7T flow from ICA to PCA in the right PCoA and upward flow in the AChA could be visualized.

In figure 6.3a and b the flow in the PCoAs measured at 7T for 0.5 mm and for 0.8 mm, respectively, is plotted. The flow in the PCoAs measured at 0.8 mm, displayed in figure 6.3b, corresponded well with the flows in the same PCoAs measured at 0.5 mm in figure 6.3a. In two of the three measured flows in figure 6.3b, pulsatility was more obvious than in figure 6.3a, due to the higher temporal resolution. Note that the curves do not change sign, indicating that no backflow was observed.
Figure 6.2 The Circle of Willis of volunteer 2 obtained from a measurement with a resolution of 0.5 mm at 7T (left column) and 3T (right column). In row a the top view of the three-dimensional Circle of Willis after segmentation of the phase contrast magnitude images is shown at peak systole. The ACoA is shown in the red square, the left PCoA in the green square and the right PCoA in the blue square. The neurological convention is used here. Row b shows velocity vectors in the ACoA at begin diastole. In row c accompanying streamlines are presented. In row d and e velocity vectors and streamlines are shown in the left PCoA at end systole. Note the visualization of blood flow in the left AChA in these images. Row f displays velocity vectors in the right PCoA and right AChA at end systole.
In table 6.3, the mean velocity magnitude, area and flow averaged over all cardiac phases and all volunteers is given for the vessels in the Circle of Willis. At 0.5 mm, the mean velocity magnitude was higher (p<0.005) at 3T than at 7T for all vessels in all subjects; the area, however, was lower (p<0.005). Flow values were similar for both field strengths (p=0.4).

These effects were not seen for the mean velocity magnitude, area and flow at 0.8 mm, see table 6.4 (p=0.86, p=0.49, p=0.17 respectively).

For 3T the mean velocity magnitudes at 0.8 mm (table 6.4) were consistently lower than at 0.5 mm (table 6.3, mean difference: 6.9 cm/s, p<0.001), whereas areas were larger (mean difference: 3.4 mm², p<0.001). There was no significant difference in mean flow between 0.8 and 0.5 mm (p=0.22).

For 7T the mean velocity magnitudes at 0.8 mm (table 6.4) were lower than at 0.5 mm as well (mean difference: 3.0 cm/s, p<0.001), whereas the areas were larger (mean difference: 2.7 mm², p<0.001). There was no significant difference in mean flow between 0.8 and 0.5 mm (p=0.59).

Table 6.3 The mean velocity magnitude, area, and flow in vessels in the Circle of Willis averaged over all cardiac phases and volunteers for 0.5 mm resolution. Standard deviations are calculated over the 5 volunteers.

<table>
<thead>
<tr>
<th>Field strength</th>
<th>Mean velocity magnitude (cm/s)</th>
<th>Mean area (mm²)</th>
<th>Mean flow (ml/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Field strength</td>
<td>3T</td>
<td>7T</td>
<td>3T</td>
</tr>
<tr>
<td>ICA</td>
<td>34.6±5.9</td>
<td>29.7±6.3</td>
<td>11.8±2.0</td>
</tr>
<tr>
<td>MCA</td>
<td>30.9±7.2</td>
<td>28.6±5.8</td>
<td>9.7±2.5</td>
</tr>
<tr>
<td>BA</td>
<td>24.8±3.8</td>
<td>21.9±5.8</td>
<td>9.3±1.9</td>
</tr>
<tr>
<td>ACA1</td>
<td>30.7±8.0</td>
<td>24.9±5.9</td>
<td>4.9±0.8</td>
</tr>
<tr>
<td>ACA2</td>
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<td>24.0±6.7</td>
<td>4.9±1.2</td>
</tr>
<tr>
<td>ACoA</td>
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<td>12.0±8.7</td>
<td>2.1±0.9</td>
</tr>
<tr>
<td>PCA</td>
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<td>17.5±3.6</td>
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</tr>
<tr>
<td>PCoA left</td>
<td>19.7±7.6</td>
<td>8.9±5.4</td>
<td>2.6±0.3</td>
</tr>
<tr>
<td>PCoA right</td>
<td>8.3±5</td>
<td>8.4±2.4</td>
<td>2.2±1.0</td>
</tr>
</tbody>
</table>

Figure 6.3 Flow in the Posterior Communicating Arteries (PCoAs) of the volunteers measured at 0.5 mm (left) and 0.8 mm (right) at 7T. Flow from the ICA to the PCA is defined as positive flow whereas flow from the PCA to the ICA is defined as negative flow. The flow is measured at the same locations as in table 6.3. Note that the curves do not change sign, indicating that no backflow was observed.
6.4 Discussion

To our knowledge, this is the first study to perform time-resolved PC-MRI at 7T. The results undoubtedly support the hypothesis that SNR is superior at 7T to 3T. Apart from the higher calculated SNR at 7T, the gain in SNR at 7T resulted in an increased amount of segmentations of the small vessels in the Circle of Willis. Furthermore at 3T, the magnitude of velocity vectors, indicated by the mean velocity magnitude, is significantly higher due to noise in the data, whereas the areas of vessels are lower. A last indication of higher SNR at 7T is that the streamlines appear smoother at 7T than at 3T, due to the lower noise levels at 7T. The median angle indicates the difference in velocity vector direction between 3T and 7T. Since the SNR is higher at 7T, the direction of the velocity vectors is more accurate than at 3T. The median angle value is therefore mainly caused by velocity vector alterations due to noise at 3T. The SNR values found by Bammer et al. [21] at 3T ranged from 43-56, slightly higher values than presented in this study at 3T (28-41 at 0.8 mm). This is consistent, however, with the larger voxel volume used by Bammer et al., namely 0.96 mm³ compared with 0.51 mm³ used here. Between 3T and 7T, the SNR averaged over all volunteers and both resolutions increased by a factor of 2.6, roughly the expected gain. A small additional gain in SNR may have resulted from the use of a 16-channel coil at 7T, with a somewhat tighter fitting, compared to an 8-channel coil at 3T. Also, the 16-channel coil may have improved the parallel imaging performance compared to the 8-channel coil, improving SNR further.

<table>
<thead>
<tr>
<th>Field strength</th>
<th>Mean velocity magnitude (cm/s)</th>
<th>Mean area (mm²)</th>
<th>Mean flow (ml/s)</th>
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</thead>
<tbody>
<tr>
<td>Field strength</td>
<td>3T</td>
<td>7T</td>
<td>3T</td>
</tr>
<tr>
<td>ICA</td>
<td>27.2±2.5</td>
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<td>17.1±1.9</td>
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<td>MCA</td>
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<td>6.0±1.4</td>
<td>-</td>
</tr>
<tr>
<td>PCA</td>
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<td>15.1±3.1</td>
<td>8.0±2.4</td>
</tr>
<tr>
<td>PCAoA left</td>
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<td>4.5</td>
<td>-</td>
</tr>
<tr>
<td>PCAoA right</td>
<td>-</td>
<td>5.7±3.0</td>
<td>-</td>
</tr>
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</table>

Table 6.4 The mean velocity magnitude, area, and flow in vessels in the Circle of Willis averaged over all cardiac phases and volunteers for 0.8 mm resolution. Standard deviations are calculated over the 5 volunteers.

*Measured in 2 volunteers: hypoplastic, not present or not segmented in others
*Measured in 3 volunteers: hypoplastic, not present or not segmented in others
*Measured in 1 volunteer: hypoplastic, not present or not segmented in others
*Measured in 4 volunteers: hypoplastic, not present or not segmented in the other

Table 6.4 Improvements
A few other studies have attempted time-resolved PC-MRI in the intracranial arteries [3, 19-22]. Flow quantification in this study corresponded well with the values presented by these groups. The flow quantification in the PCoAs visible at both 0.5 and 0.8 mm showed similar results per individual vessel at 7T. Except for Bammer et al. [21], who found an incidental type 15 variation (De Almeida classification) of the ACAs and ACoA using PC-MRI at 3T in one volunteer, none of these studies were able to visualize or quantify flow in the ACoA or PCoA.

Other studies focusing on carotid occlusion used non-gated PC-MRI in a single slice capturing the ACoA and PCoAs [23-25]. With these data the direction of the flow could be established, but no quantification was conducted. The visualization of the flow in the PCoAs may provide new insights in collateral flow pathways. From embryology it is known that the PCAs are fed with blood from the ICA via the PCoAs [43]. In our study, however, two of the six visualized PCoAs showed flow from PCA to ICA. The direction of flow in the ACoAs and PCoAs never altered over the cardiac cycle; only unilateral flow directions were found. These findings encourage further research concerning flow directions in the Circle of Willis in patients with collateral flow for instance related to internal carotid artery occlusions or severe stenosis. Furthermore, detailed flow analysis of the communicating segments of the Circle of Willis may also result in a better understanding of the preferential locations of intracranial aneurysm development e.g. close to the origin of the ACoA or PCoAs.

Several limitations of the study should be mentioned. Due to the need for an added flow-compensated acquisition alongside the flow-encoded one, PC-MRI inherently suffers from long scan times. In this study, the increased scan time due to the decrease in voxel size from 0.8 mm to 0.5 mm, was compensated by a decrease in temporal resolution. A decrease in temporal resolution results in underestimated flow measurements in the systolic phase.

Another method to save scan time was by using parallel imaging with a fairly high factor of 3. A well-known disadvantage of the use of high parallel imaging acceleration factors is the decrease in SNR [33]. Acceleration techniques that preserve SNR such as k-t BLAST [44] or compressed sensing [45] may be applied to time-resolved PC-MRI, although in these techniques impairments exist such as temporal filtering in the case of the former [46] or long reconstruction times in the case of the latter. Another limitation is the...
use of two separate scanning sessions which may have introduced physiologic variations. However, variations were minimized by performing the scans as close to each other as possible. Unfortunately, one volunteer displaced the PCoAs outside the field of view in between a localizing scan and the PC-MRI sequence.

In conclusion, due to the gain in SNR, time-resolved PC-MRI at 7T yields better visualization and quantification of flow patterns in small vessel structures that require high resolution, potentially significantly contributing to diagnostic value in patients with carotid occlusion, sickle cell disease and intracranial aneurysms.

6.5 Acknowledgments
The author would like to thank Gérard Crelier of Gyrotools, Zürich for his GTFlow updates and advice.

6.6 Disclosures
Fredy Visser is an employee of Philips Healthcare, Best, the Netherlands.

6.7 References


Quantification and visualization of flow in the Circle of Willis: time-resolved three-dimensional phase contrast MRI at 7T compared with 3T


