Complex flow in a real-size intracranial aneurysm phantom: phase contrast MRI compared with CFD

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Purpose/Introduction: Fatality rates after rupture of intracranial aneurysms are high. The therapeutic decision is based on the result of rupture risk estimation and the risks related to treatment with 3-6% morbidity and 1.5% mortality rates for coiling [1]. Apart from the morphology and size of the aneurysm, hemodynamic information may contribute to rupture risk assessment. Flow information can be captured by performing high resolution time-resolved three-dimensional phase contrast (PC) MRI [2] but this technique is not yet well validated in small structures such as intracranial aneurysms. As a validation study for the use of PC-MRI in intracranial aneurysms, high resolution PC-MRI is performed in a real-size patient-specific intracranial aneurysm phantom and compared with a CFD simulation.

Materials & Methods: A glass reproduction of a high resolution 3D Rotational Angiograph (3DRA) of an aneurysm located in the anterior communicating artery of a patient who supplied informed consent was manually created and connected to a pump and a pulse generator. The phantom of inner size 6x4x9mm (length, width, height, no up-scaling) is shown in figure 1a; the experimental setup is shown in figure 1b. The pump input waveform, shown in figure 2a was previously measured with PC-MRI in the internal carotid artery to ensure sufficient resolution in its cross-section. A retrospectively gated (20 cardiac phases) PC-MRI measurement was performed on a 3T MR system (Philips Medical System, Best, The Netherlands) in a solenoid rat coil with a diameter of 7cm at a resolution of 0.2x0.2x0.33 mm^3. TE/TR = 3.86/11.13 ms, flip angle: 15°, velocity encoding: 50x100x50 cm/s in the x, y and z direction respectively (figure 1a). Scan time amounted to 3 hours. To reduce noise the data was filtered with a 3D median filter. The CFD geometry was measured with 3DRA and segmented using VMTK [3] and consisted of 742,316 tetrahedral cells with an average node spacing of 0.12 mm. The x, y and z components of the inflow as measured with MRI were applied as the inflow boundary conditions. The flow in three heart cycles was calculated to eliminate transient effects, the third of which was used in the results section. CFD was performed in FLUENT and simulation time was 3-5 days. Results of the CFD simulation were registered onto the results of PC-MRI. The difference between CFD and MRI was quantified by calculating the root mean square error (RMSE) and standard deviation of the paired difference (SDp). To minimize registration errors these values were based on the minimum of the difference between the velocity magnitude in an MRI voxel and the velocity magnitude in the same voxel and its surrounding voxels in CFD. Differences between velocity vector angles were determined for voxels where the velocity magnitude was higher than 20% of the maximum velocity magnitude to reduce discrepancies due to the lower SNR for small velocities.

Results: In figure 2a the mean velocities in the inflow vessel and in the phantom are displayed. Figure 3 displays cross-sections of two coronal slices whereas figure 4 displays cross-sections of two sagittal slices. The orientation of these slices is displayed in figure 2b. RMSE = 1.23 cm/s, SDp = 1.4 cm/s and median angle = 5.5°.

Discussion: From figure 2a it is clear that during systole the mean velocity of the CFD simulation in the total phantom and in the inflow vessel is slightly lower during systole. In figures 3 and 4, however, where flow information is displayed for four cross-sections, distinct similarities are clear between the two modalities. In C1 a small vortex in the tip (a) and a large vortex in the middle of the phantom (b) can be distinguished for both modalities, as well as a small vortex at the bottom of the phantom (c) in diastole. In C2 the vortex (d) and the inflow jet (e) are similar as well. In figure 4 similar flow details can be discerned such as the small vortex at the base of the inflow jet (f) and the complex flow pattern in the tip of the phantom (g) in S2. The RMSE and SDp averaged over time show that differences between the velocity magnitude of both modalities are small. Furthermore the median angle for the directional difference between CFD and MRI is fairly small. The remaining discrepancies between CFD and MRI can be a result of small geometrical differences between the 3DRA and the MRI measurement and segmentation. Different fluid properties e.g. viscosity and density between CFD and MRI may play a role as well.

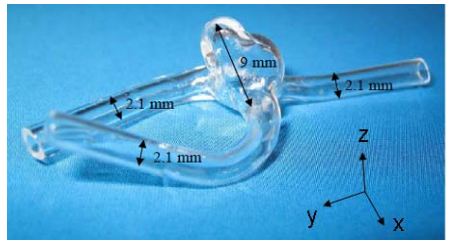
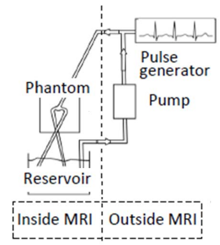


Fig 1a. Aneurysm phantom



b. Experimental set-up

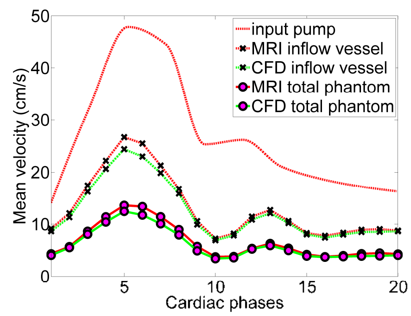
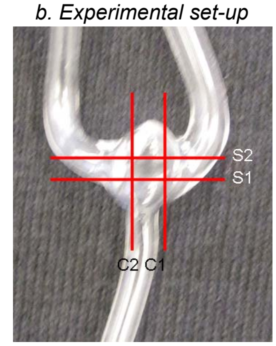


Fig 2a. Pump input and mean velocities of MRI and CFD in inflow vessel and phantom



b. Slices as displayed in figures 3 and 4

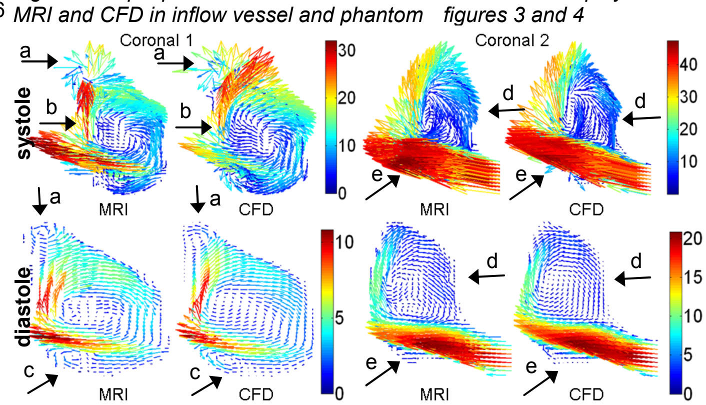


Fig 3. Coronal slices C1 (left) and C2 (right), velocities in cm/s

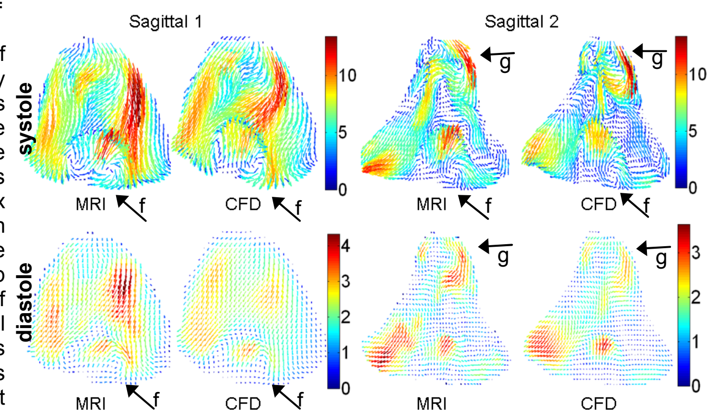


Fig 4. Sagittal slices S1 (left) and S2 (right), velocities in cm/s

Conclusion: Qualitatively, it is clear from figures 3 and 4 that flow patterns at peak systole and end diastole are very similar. Quantitatively, the differences between the two modalities are small. This study is a first step in the application of PC-MRI in aneurysms in a clinical setting by showing that time resolved 3D PC-MRI is capable of measuring complex flow patterns in small structures with an acceptable accuracy.

References:

[1] Spelle et al. J Neuroradiol, 35:116-120 (2008) [2] Markl et al.JMRI, 17(45): 499-506 (2006) [3] Antiga et al. MBEC 46(11): 1097-112 (2008)