Arterial input function measurements in the MCA: A numerical model compared with phantom experiment results

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Introduction

Dynamic Susceptibility Contrast MRI (DSC-MRI) is the clinically most often used method to asses cerebral perfusion. This method suffers, however, from quantification errors especially due to errors in the determination of the arterial input function (AIF). Previously, Duhamel investigated the AIF selection for arteries oriented at arbitrary angles with respect to the main magnetic field (1). However, this study did not investigate vessels oriented perpendicular to the main magnetic field, like the MCA, which is the most commonly used location for the AIF selection. The aim of this study is to obtain a theoretical basis for the optimal location to pick the AIF in or near the MCA. The theoretical model is subsequently validated by means of phantom experiments.

Methods

The numerical model represents the MCA as a cylinder with the magnetic susceptibility dependent on the Gd-DTPA concentration. The magnetic field changes in- and outside the cylinder were calculated from the Maxwell equations and corrected for the Lorentz-sphere (2). The R2* within the cylinder was assumed to be linearly dependent on the contrast agent concentration. The simulations were performed at a much higher resolution than is usually used in DSC measurements. This high resolution representation yielded the complex signal for contrast agent concentrations between 0 and 15 mM. Hereafter, dephasing and partial volume effects were included by regridding the high resolution representation to a voxel size (2.2x4.4x8mm) comparable to clinical protocols. This model was subsequently compared with phantom experiments. In both the model and the phantom experiments a diameter of 5 mm was used.

Phantom experiments were performed on a 1.5 T Philips scanner in combination with a Sense surface coil. The phantom consisted of a tube of 5mm diameter through which $MnCl_2$-doped water was circulated. The tube was oriented perpendicular to the main magnetic field and the concentration of Gd-DTPA within the tube was increased in steps of 1 mM. Imaging was performed using dual gradient echo imaging (TR/TE1/TE2//flip angle = 600ms/12ms/35ms/35°) using voxels of 0.6x1x8mm. These results were regridded during post-processing to 2.1x4.7x8mm.

The modelling and imaging at a higher spatial resolution are subsequently regridded to yield the possibility to study the effect of changes in location of the vessel centre with respect to the imaging grid. During regridding of the theoretical and phantom results, the voxel was shifted in steps of 0.3 mm. For each shift the mean correlation of the measured/simulated AIF with respect to the true AIF was calculated and averaged over all shifts. To evaluate the amplitude of the measured/simulated AIF, the regression coefficient was also calculated.

Results

The quality of the quantification of the AIF measurement in a vessel perpendicular to the main magnetic field is presented in Figure 1&3 by showing the correlation and regression between the simulated (or measured) AIF and the ground truth. For quantitative comparisons the data through the middle of the vessel is shown in the graphs of figure 2 and 4, for which the data was averaged over different relative positions of the center of the vessel with respect to the imaging grid.

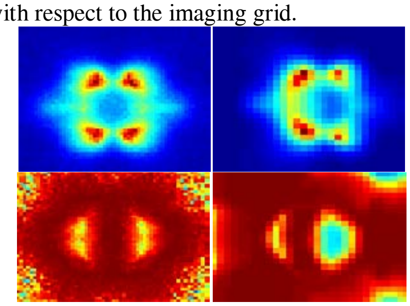

Figure 1. Amplitude (top) and correlation (bottom) for the simulation (left) and the experiment (right) using TE 12 ms.

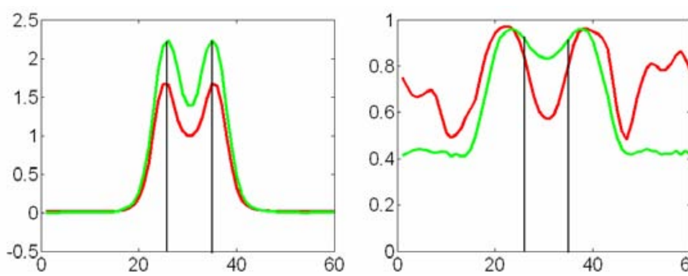

Figure 2. Regression (left) and correlation (right) for simulation (green) and experiment (red)using TE 12 ms. Gray lines indicate the vessel.

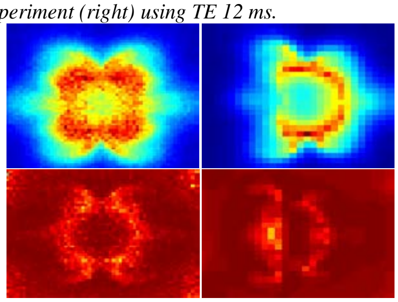

Figure 3. Amplitude (top) and correlation (bottom) for the simulation (left) and the experiment (right) using TE 35 ms.

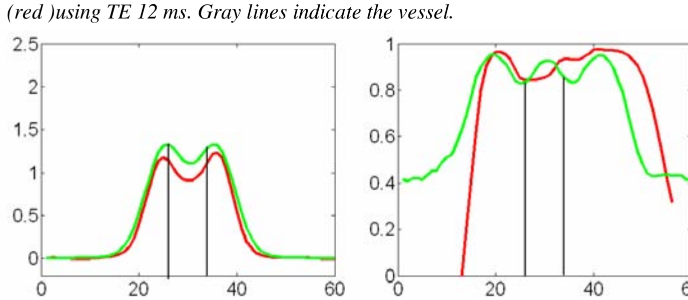

Figure 4. Regression (left) and correlation (right) simulation (green) and experiment (red)using TE 35 ms. Gray lines indicate the vessel.

Discussion and Conclusion

The most important finding of this study is that the optimal location for picking the AIF from the MCA is one or two voxels from the vessel wall. This is in agreement with previous findings of Duhamel for other vessels (1). The images of fig. 1&3 show that the numerical model agrees well with the phantom experiments. Because accurate quantification of cerebral blood flow requires both a correct shape(e.g. a high correlation), but also needs sufficient SNR (e.g. high regression coefficient) it can be concluded from fig. 2 and 4 that the best location is one or two voxels from the vessel wall.

References: 1. G. Duhamel et al. MRM 2006 2. E. M. Haacke Wiley 1999