Is PRESTO advantageous for DSC-perfusion MRI using local AIF measurements due to crushing of intravascular signal?

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Introduction

DSC-MRI requires a correct AIF for the quantification of CBV, CBF and MTT. Commonly the AIF is selected in the vicinity of the MCA with the disadvantage of delay and dispersion effects. Delay can be compensated for in the deconvolution but dispersion compensation requires extensive knowledge of the vascular bed. AIF selection closer to the capillaries makes the measured AIF less sensitive to dispersion and this AIF is flow-territory specific. A previous study showed that voxels completely outside the vessel are optimal for AIF selection (1). It could therefore be argued that crushing of vascular signal would improve local AIF measurements. The current study investigated the possible advantage using PRESTO in local AIF selection. The large gradients used for echo-shifting in PRESTO crush the vascular signal and therefore reduce partial volume effects. A 3D numerical model was created to gain insight in the possible advantage of crushing the signal inside the vessel on the AIF selection.

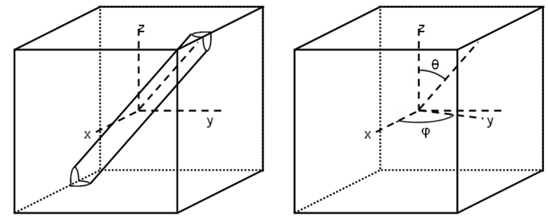

Figure 1: Schematic representation of the cylinder in the volume (left) the corresponding angles θ and φ (right).

Methods

PRESTO is an echo-shifted segmented EPI sequence that uses large gradients for echo-shifting. These large gradients crush the signal of flowing blood in the vessel and therefore annihilate the quadratic ΔR_2^* effect of Gd-DTPA in blood. A 3D numerical model with a cylinder representing the arteriole was created (see figure 1). Magnetic field changes in- and outside the cylinder were calculated for a high resolution 3D grid volume (grid spacing 250x250x250 μm^3) using the Maxwell equations corrected for the Lorentz-sphere (1). The high resolution grid volume was regridded to standard voxel sizes (3.5x3.5x3.5 mm^3) in order to include partial volume and dephasing effects. The total volume was 21x21x21 mm^3 and a box of 5x5x5 voxels was shifted in all three directions in steps of 500 μm, this to represent various locations of the vessel inside the voxels. Because both vessel size and orientation are unknown the model simulated 5 different vessel sizes (0.25; 0.5; 1; 1.5 ;2 mm) at 32 orientations (θ from 10º to 90º in steps of 10º, φ from 0º to 45º in steps of 15º) (see figure 1). In addition four echo times (11, 25, 40, 55 ms) were simulated. The main comparison was the in- and exclusion of the crushing effect in the model and therefore all other settings were kept the same. The number of voxels with a correlation with the ground truth of 0.995 or higher and relative signal strength larger than 50% for the 5x5x5 voxelvolume were counted.

Results

Table 1 shows for different vessel sizes and echo times the average number of voxels (and the standard deviation) with a correlation of 0.995 or higher and relative signal strength larger than 50% for the in- and exclusion of the crushing effect. The average number of voxels with a correct shape (correlation > 0.995, signal strength > 50%) for the in- and exclusion of crushing the vascular signal are plotted per unique combination of vessel size and orientation for the four different echo times in figure 2.

Table 1: Mean and standard deviation of each unique orientation for different vessel sizes, echo times and for crush and no crush

radius	TE1	11 ms	TE2	25 ms	TE3	40 ms	TE4	55 ms
[mm]	crush	no crush	crush	no crush	crush	no crush	crush	no crush
0.25	0	0	0	0	0	0	0	0
0.50	0.27 ± 0.42	0.42 ± 0.53*	0.38 ± 0.52	0.32 ± 0.40	0.20 ± 0.30*	0.16 ± 0.24	0.05 ± 0.11	0.04 ± 0.10
1.0	2.2 ± 2.6	3.6 ± 2.7***	1.6 ± 1.3***	1.4 ± 1.1	0.82 ± 0.72***	0.68 ± 0.68	0.34 ± 0.37***	0.31 ± 0.35
1.5	3.2 ± 3.8	5.7 ± 4.5***	3.9 ± 3.3	3.8 ± 3.5	2.3 ± 2.1***	2.1 ± 2.0	0.96 ± 1.0***	0.85 ± 0.97
2.0	3.5 ± 4.5	5.7 ± 4.7***	5.9 ± 4.9	6.3 ± 5.1	3.6 ± 3.0**	3.4 ± 3.1	1.8 ± 1.8***	1.7 ± 1.8

* larger (p<0.05)
** larger (p<0.01)
*** larger (p<0.005)

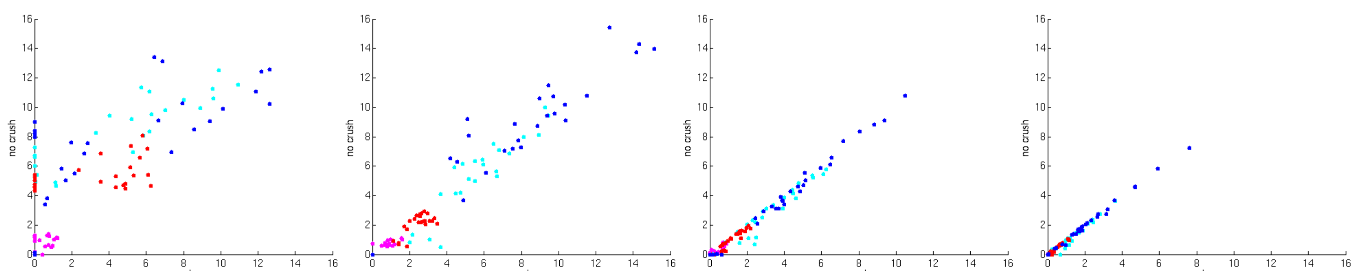

Figure 2: The average number of voxels, per unique combination of vessel size and orientation, for crushing the vascular signal and not crushing the vascular signal (0.25 mm (green), 0.5 mm (magenta), 1 mm (red), 1.5 mm (cyan) 2, mm (blue)). The four graphs show the data for an echo time of

Discussion and conclusions

The hypothesis was that crushing the vascular signal is beneficial for local AIF selection. The results of this simulation study show that crushing is not beneficial using echo times of 11 ms and 25 ms, for an echo time of 11 ms non-crushing is even significantly better. For longer echo times crushing is significantly better but the average number of optimal voxels is less. The data suggest that non-crushing sequences with short echo times are most optimal for local AIF measurements, but short echo times will degrade the quality of the measurement of the bolus passage through brain tissue. It is likely that for short echo times the local field changes around the vessel are not yet strong enough to enable the measurement of the concentration contrast agent outside the vessel. The effects of imaging such as EPI were not included in the model for computational purpose.

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

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