Transversal Relaxation Effects on Arterial Spin Labeling Investigated by Dual Echo Pseudo Continuous ASL

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Introduction

Arterial Spin Labeling (ASL) is based on the difference between a control and a perfusion weighted image. Perfusion weighting is achieved by labeling the blood downstream from the imaging slices through inversion of longitudinal magnetization. Calculation of perfusion images is subsequently performed by physiological modeling and involves correction for arrival times and longitudinal relaxation of the tagged spins. However, signal intensity is also reduced due to T2* relaxation, which is frequently ignored or is assumed to be constant for the entire brain. In practice several phenomena can cause variations in T2*, like pathology, B_0 inhomogeneities and blood oxygenation changes due to (de-)activation of the brain (the BOLD effect).
In this study, a dual echo ASL imaging technique was applied and data from both echoes were used to estimate T2* voxelwise and to correct the perfusion images for transversal relaxation. Subsequently, errors due to spatial variation in T2* were quantified. The influence of the BOLD effect on perfusion fMRI was investigated by analysis of dual echo data from a block design finger tapping experiment.

Materials and methods

Eight healthy volunteers (6 male, 2 female) were scanned on a 3T, clinical scanner (Philips Medical Systems, The Netherlands). For ASL a pseudo continuous labeling scheme was applied that comprised 0.5 ms Hanning pulses with 1ms interval, flip angle 27º, gradient strength of 6 mT/m and a gradient duration asymmetry of 150 µs (1). Labeling duration was 1250 ms, with a delay before imaging of 1450 ms and background suppression pulses at 70 ms and 1070 ms after labeling ended. The imaging part consisted of a dual echo single shot EPI sequence with TR/TE1/TE2 of 1300/12/36 ms (7 slices, SENSE factor 2.5, in plane resolution of 3 x 3 mm^2). Acquisition was performed with slice thickness of 8 mm and 5 mm (gap of resp. 1.5 mm and 5mm). Two datasets were acquired, covering the upper and lower part of the brain. During the first acquisition, bilateral activation of the motor cortex was invoked through finger tapping with a block design paradigm of 1 min blocks and total scan-time of 6 min.
For each voxel T2* and $S_{0,DE}$ were calculated from both echoes, assuming a mono-exponential signal decay. Corresponding S_0 for single echo acquisition ($S_{0,SE}$) was calculated from the first time point by assuming a uniform and constant T2* of 50ms (2). Subsequently the relative error for the single echo approach was calculated: rel.error = 100% *($S_{0,DE}$-$S_{0,SE}$)/$S_{0,DE}$. On the control images regions of interest (ROI) were drawn adjacent to the nasal cavity, petrous bones and frontal sinus and in white matter of the internal capsule and in grey matter at the same level.
For the functional scans surround subtraction was used (3) and scans were analyzed with SPM99 (uncorrected threshold of 0.0005, minimal clusters size of 4 voxels). Voxels that showed activation for both TE1 and TE2 were averaged and changes in T2* and S_0 were determined.

Results

Differences for single versus dual echo extrapolation of S_0 are illustrated in fig 1, showing that single echo extrapolation underestimates perfusion in areas close to air cavities, although dual echo extrapolation can show some noisy voxels in these regions. The relative error in perfusion signal due to spatial variation in T2* is shown in table 1. Substantial errors are found in tissue close to air containing cavities and the petrous bones, whereas errors for grey and white matter ROIs were minimal. For imaging with a smaller slice thickness, errors were significantly reduced for frontal sinus and nasal cavity ROIs. Table 2 shows the results for functional, dual echo ASL. During activation, T2* of the labeled spins were significantly longer compared to rest, resulting in significant overestimation of CBF-changes when using single echo extrapolation of the first or second echo.

Discussion and conclusions

Ignoring T2* variations or assuming a constant T2* may lead to substantial errors in ASL calculations, especially in areas prone to magnetic field perturbations. Single echo acquisition can also severely overestimate CBF changes upon activation, due to relatively large changes in T2* upon activation due to the BOLD effect. This relatively large change in T2* upon activation supports theoretical models that predict a large capillary contribution to the ASL signal (2).
Dual echo acquisition in pseudo-continuous ASL enables voxel based correction for T2* variations that enables accurate estimation of quantification errors in perfusion MRI. The use of spiral readouts could circumvent these T2* errors.

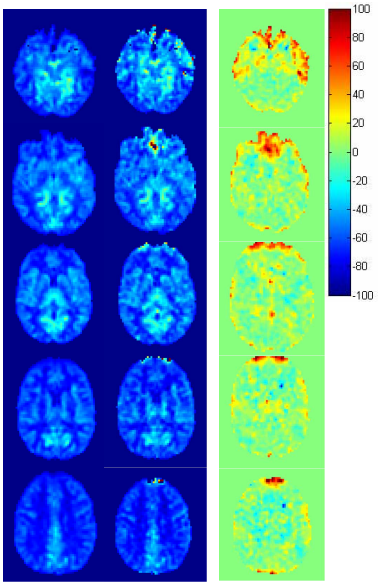

Figure 1. Example perfusion images with single echo (left) and dual echo (centre) extrapolation of S_0. Right column shows relative error in single echo $S_{0,se}$ as compared to dual echo, the color bar shows relative error in %. Note the substantial error in the area close to the frontal sinus and nasal cavity.

ROI	8mm error (%)	5mm error (%)
Grey matter	0 ± 3	0 ± 3
White matter	-1 ± 5	-1 ± 4
Frontal sinus	30 ± 8	22 ± 9*
Nasal cavity	38 ± 10	30 ± 13*
Petrous bone	33 ± 5	33 ± 8

*Table 1. Mean relative error ± stdev in S_0 when single echo extrapolation is applied for 8 and 5mm slice thickness. Data refer to specific regions of interest in eight volunteers. * p-value ≤ 0.01*

	Mean	p-value
T2* rest	52 ms	0.005†
T2* active	65 ms	
S_0 change	39%	
TE1 change	49%	0.017‡
TE2 change	68%	0.024‡

Table 2. Comparison of rest vs active periods during functional MRI (n=6). † Significant result for rest vs active. ‡ Significant change of signal at t = TE compared to change in S_0

References

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