Dependence of R2* bias on through-voxel frequency dispersion and gradient echo train in high-resolution 3D R2* mapping

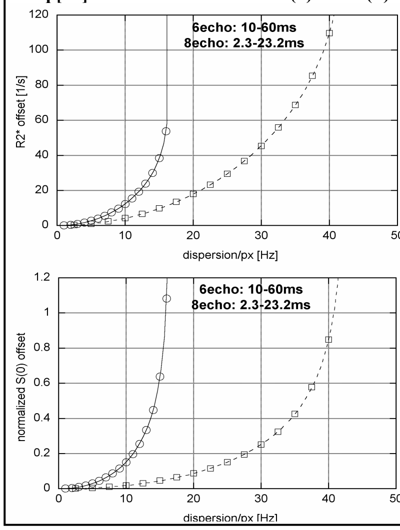
G. Helms[1], and P. Dechent[1]

[1]MR-Research in Neurology and Psychiatry, University Medical Center, Göttingen, Lower Saxony, Germany

Introduction:

The effective rate of decay of transverse magnetization, R2* = 1/T2*, is sensitive to B_0 distortions at a mesoscopic scale (1). It is thus highly correlated to tissue concentrations of non-heme iron (2) and myelinated axons (3), especially at high and ultra-high field strengths (2-4). R2* is commonly measured using long-TR multi-slice gradient echo sequences. This approach implies high SNR and long echo trains, but limited resolution and increased sensitivity to through-voxel gradients in the slice direction. The latter have been modelled by linear dispersion imposing a sinc modulation on the exponential T2* decay (5). However, many fiber tracts and iron-containing nuclei involved in motor disease (like substantia nigra and subthalamic nucleus (6)) have a double-oblique orientation, so R2* measurements may be compromised by a 2D approach. 3D measurements at isotropic high resolution is also desirable for voxel-based statistical approaches (7) requiring spatial normalization. Here, the time required for the additional phase-encoding contrains TR and thus the duration of the gradient-echo train. Therefore, the behaviour of sinc model was studied for constraints of short gradient-echo train to guide the implementation of 3D R2* mapping scheme.

Fig. 1: Simulated R2* offset Eq.[3a] and S0 offset Eq.[3b] for the schemes of refs 6 (○) and 7 (□)



Theory and simulations:

The modulation of the sinc function is parameterized by the frequency dispersion $\Delta\nu = \gamma G \Delta x$ (in Hz/pixel) of a through-plane gradient G along the voxel dimension Δx. Up to the first root (or "node"), it is excellently approximated by its Taylor expansion:

$$sinc(\Delta\nu TE) = sin(\pi\Delta\nu TE)/(\pi\Delta\nu TE) \approx 1-(\pi\Delta\nu TE)^2/6 + (\pi\Delta\nu TE)^4/6/20 - (\pi\Delta\nu TE)^6/6/20/42 \quad [1]$$

The log signals S(TE) = S0 *exp*(R2*TE) *sinc*(ΔνTE) are used, R2* and Δν appear at different order:

$$log(S(TE)) \approx log(S0) - R2^*TE - (\pi\Delta\nu TE)^2/6 - (\pi\Delta\nu TE)^4/180 - (\pi\Delta\nu TE)^6/2835; \quad [2]$$

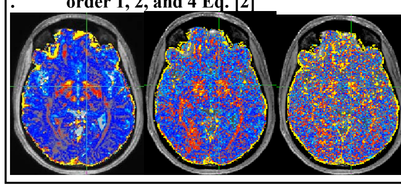
so the dephasing correction can be implemented as polynomial regression. For standard linear regression, Eq. [2] predicts an addititive bias ΔR2* = R2*(fitted) – R2* that is independent of R2* and increases with $\Delta\nu$ if $\Delta\nu^2$, if TEmax <<1. Log regression with different R2* and dispersion was simulation for TE sampling schemes reported for 3T (2-4,6,7) A steep increase of the ΔR2* offset was observed for if $\Delta\nu TE_{max} \rightarrow 1$, so ΔR2* was fitted empirically by:

$$\Delta R2^* = A\,\Delta\nu^2 / [1 - \Delta\nu TE_{max}]^B \quad [3a]$$

This was also used for the normalized S0 offset

$$\Delta S0/S0 = C\,\Delta\nu^2 / [1 - \Delta\nu TE_{max}]^D \quad [3b]$$

Fig. 2: R2* noise due to polynomial regression of order 1, 2, and 4 Eq. [2]

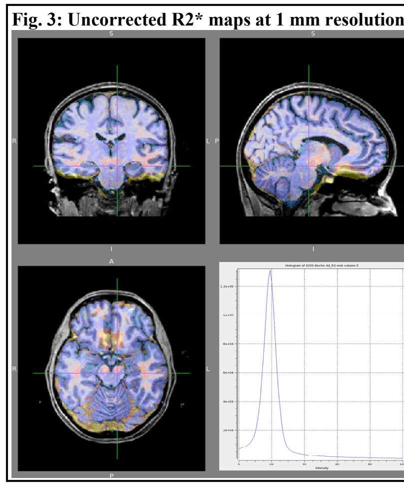


Experimental:

3D multi-echo FLASH imaging was performed on consenting healthy adults on a 3T whole-body system (Siemens Tim Trio) using an 8-channel receive-only head coil and the body coil for transmission. First, similar to (6), 8 bipolar echoes at multiples of TE = 4.92 ms with isotropic 1 mm resolution (non-selective excitation of 176 sagittal partitions, 256x176 mm field-of-view, TR/α=23 ms/6°). Measuring time using 6/8 partial Fourier (phase/partition) and 2x GRAPPA (24 reference lines, phase) was 6.5 minutes, respectively.

FSL 4.0 (/www.fmrib.ox.ac.uk/fsl) was used for image-processing. After brain extraction, regression of order 1, 2, and 4 was performed on the log signals. The frequency offset was calculated from unwrapped phase images. Local gradients were derived by a differentiation kernel and used to correct R2* with parameters A and B determined by simulation.

Fig. 3: Uncorrected R2* maps at 1 mm resolution



Results:

Simulation: The dispersion related offset of R2* and S0 were independent of R2* and excellently described by Eqs.[3a,b] (Fig.1). Thus, longer echo trains are more sensitive to frequency dispersion. The initial increase A was correlated to the shortest TE_{min}, whereas a fewer number of echoes increased the exponent B. *Experimental:* Polynomial regression introduced errors into the R2* maps (Fig. 2) probably due to a correlation between the terms. Correction of R2* by Eq.[3a] introduced noise or failed probably due to oblique gradients (not shown). Figure 3 shows R2* overlays and histogram of the 3D acquisition. The high-resolution/short TE scheme yielded reliable R2* mapping down to the level of the pons and the cerebellum SN, with excessive bias (R2* > 50 sec^{-1}) only in orbito-frontal and temporo-basal regions.

Discussion:

An empirical model for the influence of through-voxel gradients on R2* was derived from simulations, showing how the TE sampling scheme influences the sensitivity to frequency dispersion. 3D mapping of R2* advocates short gradient echo trains and high-resolution; thus confining the R2* bias to rather small sub-regions. This proved to be more feasible than correction. Non-selective excitation reduces the influence of small dispersion via a short TE_{min}. Despite the general trade-off between statistical error and sensitivity to bias, high-resolution R2* mapping of the brain seems feasible using 3D sequences at 3T.

References:

[1] Yablonskij, Haacke. *MRM* 32:749 (1994) [2] Yao et al.; *NeuroImage* 44:1259 (2009) [3] Li et al. *MRM* ePub (2009) [4] Peters et al. *MRI* 25:748 (2007) [5] Fernández-Seara, Wehrli. MRM 44:358 (2000) [6] Helms et al. Proc ISMRM 16 (2008) [7] Péran et al. JMRI 26:1413 (2007)