Evaluation of correction methods for errors in T2* quantification caused by background gradients

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Introduction: Accurate measurement of T_2* is of importance for many basic, preclinical, and clinical MR applications. Examples include blood oxygenation level dependent functional (BOLD) imaging [1], super-paramagnetic iron oxide contrast imaging [2], and tissue iron quantification in the brain [3], heart [4], and liver [5]. However, T_2*-measurement accuracy can be influenced by macroscopic field inhomogeneities (i.e., background gradients) that are introduced by susceptibility changes, e.g. at air-tissue boundaries. Two major approaches have been proposed to correct for the errors in T_2*-quantification that arise from background gradients: (a) *sinc*-weighted fitting of the signal decay (FIT) [6], and (b) direct measurement of the field ΔB_0 (DMF) [7]. As no formal comparison of both strategies has been performed yet, the pros and cons associated with each technique are unclear as well as which technique should be used for a certain application. Purpose of this work is to compare and evaluate these two methods in phantoms and *in vivo* with specific focus on hepatic imaging.

Materials and Methods: The MR signal M acquired by a multi-echo spoiled gradient echo (mGRE) sequence can be expressed as: $M(TE)=d\cdot M_o\cdot exp(-TE/T_2^*)\cdot sinc(\gamma\ TE\ G_b\ d/2)$ [Eq.1] where TE is the echo time, d the slice thickness, and G_b the background gradient caused by field inhomogeneities. When using FIT, G_b is treated as an unknown parameter and T_2* is obtained by a three parameter fit (T_2*, M_0, and G_b) [6]. For DMF, G_b is estimated from B_0 field maps calculated by phase images obtained from the mGRE acquisition; after G_b has been determined, T_2* is calculated by a two parameter fit (T_2* and M_0) [7]. The gradient calculation relies on the field information of two neighboring slices. Therefore, no gradient can be determined for the two outmost slices of an image stack [7]. In this study, Autoregressing Moving Average Modeling (ARMA) [8] is used to generate field maps and the resulting G_b maps. All algorithms were implemented in Matlab (MathWorks), and fitting was performed on a voxel-by-voxel basis. *Phantom Scans:* to generate field inhomogeneities caused by air/water boundaries, an empty cuboid bottle (volume $7.5\times7.5\times18cm^3$) was sealed and glued into a square container (volume $20\times20\times19cm^3$) filled with 8.33mM Gd-doped water (Fig. 1). Images were acquired on a 1.5T scanner (Siemens Avanto) using an mGRE sequence with 20 echoes. Parameters were as follows: TE_{min} 0.98ms, echo spacing 1.38ms, TR 200ms, flip angle 25°, FOV 24×30cm, and matrix size 128×104. Seven coronal slices were acquired with a slice thickness and inter slice gap of 10mm. *In Vivo Scans:* hepatic imaging of healthy volunteers was performed on a 3T scanner (Siemens Trio) using the mGRE sequence with 20 echoes in a single breathhold. Acquisition parameters were TE_{min} 1.73ms, echo spacing 1.29ms, TR 200ms, flip angle 35°, FOV 28.4×35cm, and matrix size 128×104. Nine transverse slices were acquired, starting from the diaphragm and covering the liver with a slice thickness of 10mm and no gap. Only the 7 center slices were used for the calculations.

Results and Discussion: Table 1 summarizes the results obtained for the phantom at specific locations ROI 1 & 2 (Fig. 2) and a healthy volunteer (male, age 23y). *Phantom Scans:* The effect of background gradients was not observed when the ROI was far away from the air/water boundary. In that case (slices 1 & 2 for ROI 1 and slices 1-4 for ROI 2) the uncorrected T_2* values were around 20ms. For slice 1, uncorrected and FIT T_2* values are very close. Approaching the air/water boundary, uncorrected T_2* values decreased (bold), indicating that increasingly stronger background gradients impact the decay. FIT-corrected T_2* values were around 20ms (range 18.2-20.7ms) for both ROIs. This indicates that the FIT correction works properly. The DMF algorithm corrected slice 2 pretty well, but showed over-correction for both ROIs for slice indices approaching the air/water boundary. Slice 7 was placed too close to the empty bottle, so that no T_2* map could be calculated with either approach. *In Vivo:* The last column of Table 1 shows the results obtained for a healthy volunteer. Here, the first slice is close to the lung/liver interface. Steep drops in uncorrected T_2* can be seen for the first few slices (bold) because of background gradients caused by the lung/liver boundary. At the center of the liver (slice 7, portal vein level), the uncorrected liver T_2* reaches a maximum of 20.3ms. As this is a healthy volunteer, the T2* of liver tissue should not vary strongly within the organ. It therefore seems reasonable to attribute the observed deviation of approximately 50% in liver T_2* between slices 1 & 2 and slice 7 to stronger background gradients near the lung/air interface. FIT recovered T_2* to approx. 24–25ms. However, for slice 1 the value could still not be completely corrected. DMF corrected slices 5-7 in the center of the liver well where background gradients are small. However, the DMF correction seemed to deteriorate the closer a slice gets to the lung/liver interface.

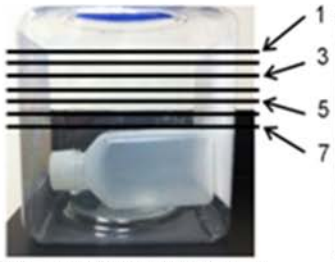

Fig 1. Phantom with slice positions.

Fig 2. ROI locations.

Table 1. Summary of T_2* corrections for the phantom and a volunteer

slice	ROI 1 T2* [ms]			ROI 2 T2* [ms]			Volunteer T2* [ms]		
	Uncorr	FIT	DMF	Uncorr	FIT	DMF	Uncorr	FIT	DMF
1	20.0	20.5	--	20.1	20.5	--	**9.9**	16.9	10.5
2	19.4	20.2	22.5	20.4	20.6	20.8	**9.3**	22.5	13.6
3	18.8	20.2	26.6	20.4	20.8	20.5	**12.5**	24.9	16.5
4	**16.9**	20.0	90.9	20.2	20.7	23.5	**14.0**	25.8	17.0
5	**11.6**	18.4	32.7	**14.5**	19.1	70.0	**17.5**	24.4	20.1
6	**15.0**	19.3	--	**9.9**	18.2	--	18.3	25.3	20.1
7	--	--	--	--	--	--	20.3	28.4	20.7

Conclusion: There is a need for correction of T_2* values in areas of strong susceptibility changes. By comparing FIT and DMF, we found that FIT worked better in correcting T_2*. DMF is not very accurate and prone to over- or underestimation. The reason is that the field maps (and the resulting gradient) calculated from phase images represent the magnitude of all three (x,y,z) field components. However, G_b in Eq.[1] is only the gradient component along the z direction. This can cause over- or undercorrection of T_2*. Our results indicate, that FIT is preferred unless the z component of the magnetic field dominates among the three spatial components.

References: [1] Sadowski EA et al. MRI 2010;28:56-64. [2] Dahnke H et al. MRM 2005;53:1202-6. [3] Gelman N et al. Radiology 1999;210:759–67. [4] Hankins J et al. Ped Blood&Cancer 2010;55;495-500. [5] Hankins J et al. Blood 2009;113:4853-5 [6] Fernandez-Seara M et al. MRM 2000;44:358-66. [7] Hernando D et al. MRM 2012;68:830-40. [8] Taylor BA et al. JMRI 2012;35:1125-32.