Fast T2* mapping with SENSE acquisition and field inhomogeneity correction for lesion characterization and sensitive SPIO detection at 3.0T

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Introduction T_2* relaxometry is considered to be a sensitive tool for the characterization of lesions and the quantitative detection of small amounts of iron oxide contrast agents [1]. The accurate determination of T_2* can strongly be hampered by motion artifacts and large-scale field inhomogeneities ΔB_0. Motion artifacts occur especially in abdominal scans, e.g. in the liver. Parallel imaging (e.g. SENSE) allows the acquisition of multiple gradient echoes in a number of slices within a single breath-hold. Field inhomogeneities occur at air tissue interfaces, like lung/liver and sinus/brain, leading to signal losses and an overestimation of the relaxation rate R_2*. In order to obtain quantitatively precise R_2* values, a correction method is applied that corrects the influence of ΔB_0 inhomogeneities and provides an additional parameter for the characterization of tumors and the sensitive detection of iron oxide contrast agent distributions [2,3].

Materials and Methods Experiments have been performed on a 3.0 T whole-body scanner (Philips Intera). A multi gradient echo sequence obtains images at different echo times allowing the calculation of the relaxation time for each pixel. However, in the presence of a linear ΔB_0 inhomogeneity in slice-selection direction the exponential decay of the T_2* relaxation is multiplied by a sinc-function ($\sin(\omega t)/\omega t$) that induces an oscillation to the decay curve. For the correction of this sinc-function a two-step procedure is applied (see also Fig. 1). First, a frequency map is calculated from the phase of a multi gradient echo dataset, which is then utilized to determine the ΔB_0 inhomogeneity gradient over each voxel. Second, this gradient information is used to eliminate the sinc-function from the relaxation curve. The multislice scan does not require a very high resolution in slice direction, because the calculated ΔB_0 gradient is not directly used to correct the measured relaxation signal, but is employed as an initial value for an iterative algorithm. This algorithm searches for the value ΔB_0, which leads to a minimum standard deviation between a mono-exponential fit and the corrected relaxation data. This approach leads to a fast post-processing correction of the T_2* maps (30s for a 128^2 matrix). The corrected T_2* maps are independent of the local field variations.

The experiments were performed using a multi gradient echo sequence. For abdominal imaging of a patient with liver metastases a 6 element phased array coil and a SENSE factor of 2 was used to acquire T_2* maps of three subsequent slices in a single 14 s breath hold (TR=76ms, ΔTE=1.7ms, flip angle 28°, Matrix 224x160, 29 echoes). Phantom and brain imaging (healthy volunteer) was done with TR=153 ms, flip angle 30°, ΔTE=1.6 ms, and 34 echoes. In the phantom setup, glass tubes filled with different concentrations of super paramagnetic iron oxide (SPIO, Resovist, Schering AG) were used.

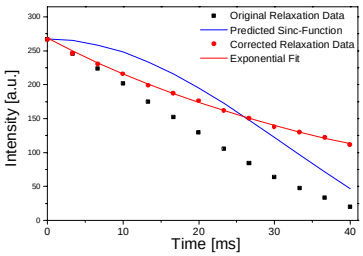

Fig.1: Relaxation signal from a phantom experiment. The original (black) are strongly influenced by the sinc-function The data, which are corrected (red), are well described by a mono exponential decay fit and show the part of the relaxation that is not distorted by ΔB_0 effects (blue).

Results Fig. 2 shows T_2* maps of the patient's abdomen computed with a mono-exponential fit and calculated with the described correction method. It reveals that some of the metastases that are superimposed by a strong susceptibility artifact in the uncorrected T_2* map are clearly visible in the corrected T_2* map (metastases T_2*: 12ms and liver T_2*: 20-30 ms). This opens up the possibility for quantitative tissue relaxometry, which may help to characterize liver lesions as well as monitor lesions during tumor treatment.

Fig. 3 shows an uncorrected and corrected T_2* map of a phantom setup that was exposed to strong ΔB_0 inhomogeneity. The difference in SPIO concentration for the right glass tube is 5 times higher than the measured standard deviation of T_2* values in the surrounding solution, this difference is only detectable in the corrected T_2* map. This shows that a difference of R_2* with a SNR of 5 is well detectable by this method. Fig. 4 shows that T_2* maps of brain are also influenced by ΔB_0 inhomogeneities. Calculating the mean R_2* over the whole brain, except the ventricle, leads to an uncorrected R_2* of 26 ± 4 s^{-1} and a corrected R_2* of 17 ± 2.5 s^{-1}. These results reveal a strong overestimation of the uncorrected data and a smaller standard deviation after the correction. These results were used to derive an upper detection limit for SPIO concentrations in different tissues. The amount of SPIO necessary to be detected in these organs can be estimated from the standard deviation of the mean R_2* and the previously measured relaxivity to be $120x10^3$ cells/ml (2.4 µg Fe/ml) in brain, under the assumption that cells are labeled with 20 pg Fe/cell [4]. Taking into account the voxel size, the predicted sensitivity is approximately is 600 cells per voxel for brain imaging at a resolution of 1x1x5 mm.

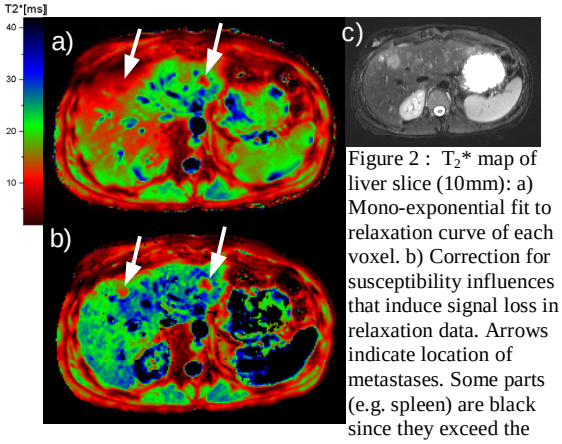

Figure 2 : T_2* map of liver slice (10mm): a) Mono-exponential fit to relaxation curve of each voxel. b) Correction for susceptibility influences that induce signal loss in relaxation data. Arrows indicate location of metastases. Some parts (e.g. spleen) are black since they exceed the color scale.
c) T_2 weighted image

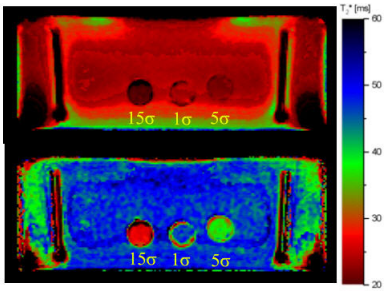

Figure 3: SPIO-Phantom. Top: Exponential fit Bottom: Corrected for susceptibility influences. The difference in SPIO concentration between the right tube and the surrounding liquid is equivalent to 5 times the mean standard deviation of the surrounding liquid.

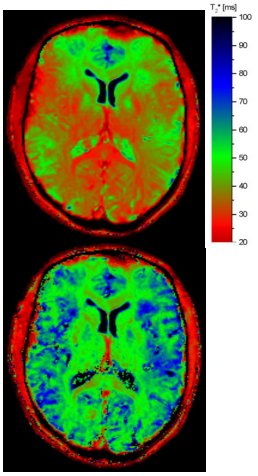

Figure 4: T_2* map of brain slice (5mm). Top: Exponential fit to relaxation curve of each voxel. Bottom: Corrected for susceptibility influences that induce signal loss in relaxation data.

Conclusion The proposed approach increases the accuracy of T_2* relaxometry by eliminating the overestimation of the relaxation rate due to the signal loss induced by ΔB_0 inhomogeneities. Motion artifacts are suppressed by single breath-hold abdominal T_2* map acquisition utilizing parallel imaging methods. No extra measurement time is needed, since the values of ΔB_0 are calculated from the phase of the multislice data. The parallel imaging acquisition and post-processing correction method allows fast quantitative MR imaging.

References [1] Bowen et al. Magn Reson Med 2002;48:52-61. [2] Fernandez-Seara et al. Magn Reson Med 2000;44:358–366 [3] An et al. Magn Reson Med 2002;47:958–966. [4] Frank et al. 2003 Radiology 228:480-487.