Clinically-Driven Fast and High-Resolution Mapping of T1, M0, and B1 with Whole Brain Coverage

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Introduction

Quantitative MR techniques have become increasingly on demand in human brain imaging. In particular, accurate mapping of the longitudinal relaxation time, T1, and water content, M0, is of prime interest pushing the development of techniques with high spatial resolution and short experimental durations. The current most efficient method for obtaining T1 maps is based on acquiring two spoiled gradient recalled echo (SPGR) images in steady states with variable flip angles [1,2]. Further acquisitions are usually required to obtain the brain tissue water content. Several factors, including inhomogeneities of the RF field, B1, particularly at higher filed strengths [3,4], and low SNR may negatively affect the accuracy of these methods and produce systematic errors in T1 and M0 estimation [4,5]. Correct values can still be achieved by applying suitable corrections [2-5]. However, the concomitant increase in scan time renders these techniques intractable in a clinical environment. In this study, we present a modified two-acquisition SPGR method for simultaneous B1, T1, and M0 mapping with an isotropic spatial resolution of 1×1×1 mm^3 that covers the entire human brain in a clinically acceptable time.

Materials and Methods: The steady state SPGR signal is given by: $S = M_0 \cdot \frac{1-e^{-TR/T_1}}{1-\cos(B_1\alpha)\cdot e^{-TR/T_1}} \cdot \sin(B_1\alpha) \cdot e^{-TE/T_2}$, where B1, the actual-to-nominal flip angle ratio, describes the flip angle inhomogeneity and M0 is the equilibrium magnetization of water. The variable flip angle SPGR approach [1,2] with optimal-SNR T1 mapping gives rise to a low flip angle (<6°) for relatively low TR's (<30 ms) for which the SPGR signal can be approximated with: $S \propto M_0 \cdot \sin(B_1\alpha) \cdot e^{-TE/T_2} \propto M_0 B_1 \alpha$. Since B1 is a spatially smooth function compared to M0 (which is of low contrast in brain tissues) and e^{-TE/T_2} is pixel independent, B1 map can be approximated by heavily smoothing the resultant low-flip angle SPGR signal; and then be reused for more accurate T1 and M0 estimation during the linear fitting. Five healthy subjects and four patients (three with ALS and one with mild-TBI) were scanned as part of larger clinical research protocols on a 3.0 T clinical MR scanner (Trio/TIM; Siemens Medical Solutions, Erlangen, Germany) using transmit body coil and an 8-ch phased-array receive-only head coil. The following acquisition parameters were used for the two SPGR scans: TR/TE=8.4/3.76 ms, α1=3° and α2=15° (optimal flip angles, *c.f.* Refs [1,2]), matrix=256×256, bandwidth=210 Hz/Px, FOV=256×256 mm^2, 160 slices, 1mm slice thickness, no gap between slices, iPAT acceleration factor = 2 with 24 auto-calibration lines, and total scanning time = 2 × (3 min and 14 s). Considering individual variations in T1 [6], T1 values in WM and GM were measured individually from the whole brain T1 histogram with two Gaussian distributing fittings [7]; and T1 value in CSF was limited to 3000 ms. The water content for each tissue type was normalized to CSF and calculated from ROI's randomly selected in WM and GM tissues.

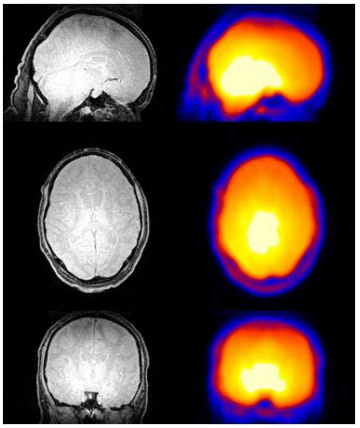

Fig 1: (left) Representative original SPGR images with low flip angle [1,2] linearly affected by the non-uniformity of the transmit B1 field. (right) 3D B1 map estimated by heavily smoothing, color map range: 0.32 – 1.27

Results and Discussion: Figure 1 shows the 3D low-flip angle SPGR image and the approximated 3D B1 map in three orthogonal planes. The color map represents a B1 range of 0.32 – 1.27 in the brain. Figure 2 shows an arbitrary slice (from the whole brain) of the estimated B1 and the corrected T1 and M0 maps from a healthy subject (top row) and an ALS patient. Uncorrected M0 maps with higher B1-nonlinearity manifestation are also shown for comparison. Figure 3 shows the 3-plane M0 maps from the TBI patient indicating the small lesions (arrows) which are in agreement with the clinical finding. Figure 3 also illustrates the patient T1 histogram and a representative T1 histogram of the healthy subjects. From the T1 histograms of five healthy subjects, average T1 values were measured as 890 ± 62 ms in WM and 1297 ± 108 ms in GM. The ROI analysis resulted in mean (WM, GM) water content of (0.73, 0.84), (0.69, 0.81), and (0.67, .78) for normal subjects, the ALS, and the TBI patients, respectively.

High spatial resolution with full brain coverage T1 and M0 mapping was achieved by the B1-adjuasted SPGR variable angle method that is suitable for neurological clinical research. Two welcome byproducts of the method are maps of B1 and M0 that can be used to correct other acquisitions and provide more information in the same session. Quantitative T1 maps together with water content maps provide larger range of contrast than conventional images, which may improve the accuracy of the segmentation. The proposed method can also be used to measure tissue-specific quantities in age-dependent groups and disease populations.

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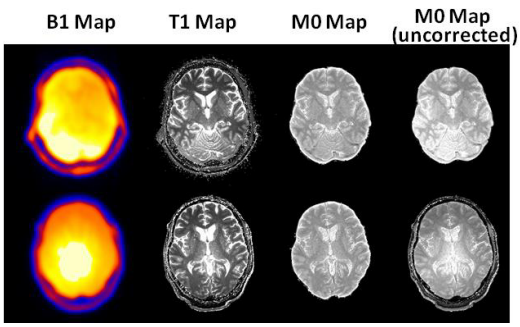

Fig 2: A representative high-resolution slice of the estimated B1 and the corrected T1 and water content M0 maps from a healthy subject (top row) and an ALS patient. Uncorrected water maps with more B1-nonlinearlty evident are also shown for comparison.

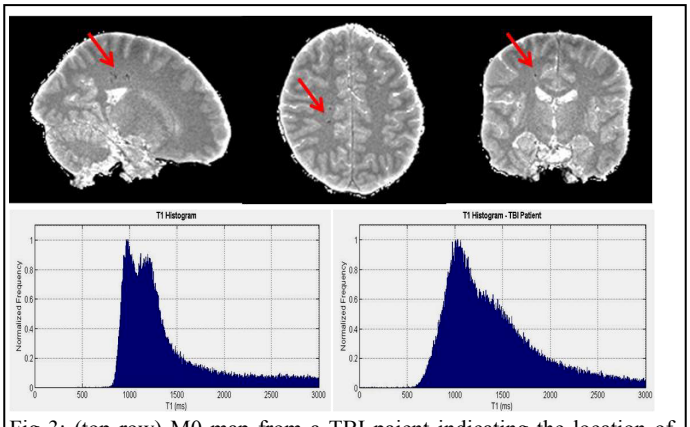

Fig 3: (top row) M0 map from a TBI paient indicating the location of small lesions consistent with the clinical finding. (bottom row) T1 histograms in a healthy subject (left) and the TBI patient of top row.