Fast T1/B1 Mapping using multiple dual TR RF-spoiled Steady-State Gradient-Echo Sequences

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Introduction: Dynamic Contrast-Enhanced MRI (DCE-MRI) is widely applied to assess tissue perfusion and vascular permeability [1]. In most of these applications, a T_1 map, obtained prior to contrast agent (CA) administration, is used to convert the signal intensity from a series of T_1-weighted spoiled gradient-echo images into CA concentration. The accuracy of the approach is mainly affected by the flip angle dependency of the steady-state signal, which necessitates careful quantification of B_1 inhomogeneities in advance [2]. This study investigates a new approach called "Multiple TR B_1/T_1 Mapping" (MTM), capable of fast, simultaneous B_1 and T_1 mapping [3]. This approach is based on the "Actual Flip angle Imaging" (AFI) sequence [4], but uses multiple TR pairs instead of the standard AFI approach of a single TR pair. Recently, MTM has been investigated in terms of B_1 mapping performance [3]. In this work, MTM is analysed with respect to its T_1 mapping performance in comparison with an inversion recovery reference sequence and in due consideration of the limited time allowed in a clinical set-up.

$$S(TR_1, TR_2, TE, T_2^*; T_1, \alpha, \widetilde{M}_0) = M_0(T_2^*, TE, \widetilde{M}_0) \cdot F_i(TR_1, TR_2, \alpha, T_1) \quad (1)$$

$$F_1(\alpha, T_1) = ((1-E_2)\sin\alpha + \tfrac{1}{2}(1-E_1)E_2 \sin 2\alpha)/(1 - E_1 E_2 \cos^2\alpha) \quad (2)$$

Theory: The standard AFI employs a dual TR RF-spoiled steady-state gradient-echo sequence. The signal intensities S_1 and S_2 measured in the intervals TR_1 and TR_2, can be described analytically [4] (Eq. 1). The coefficient F_i is different for signals S_1 and S_2. For the first image S_1, it is given in Eq. (2), with $M_0 = \widetilde{M}_0 \cdot e^{-TE/T_2^*}$ and $E_{1,2} = e^{-TR_{1,2}/T_1}$. S_2 can be obtained from Eq. (2) by interchanging indices. MTM employs multiple dual TR sequences using different TR_1, TR_2 in subsequent measurements. Each dual TR sequence yields two data points $F_1(TR_1,TR_2,\alpha,T_1)$ and $F_2(TR_2,TR_1,\alpha,T_1)$ from signals S_1 and S_2 for each pixel. The parameters α, T_1, and M_0 can then be obtained by a numerical fit of the theoretical expressions for the signal intensities Eqs. (1,2) to the set of measured data points. Hereby, α and T_1 are obtained simultaneously and independently of each other. Thus, no further correction of either parameter has to be made, and no systematic B_1/T_1 errors are present in this method. Sequence parameters can be optimized to allow for optimal B_1 and/or T_1 mapping. In this study, the Cramer Rao Theorem (CRT) [5] was used to achieve best T_1 mapping performance.

Subjects and methods: (1) Sequence parameters for MTM were optimized with respect to maximum SNR in T_1 mapping using CRT. For optimization, a target T_1 = 900ms was chosen. SNR of MTM T_1 mapping was predicted as a function of T_1. (2) Experiments were conducted using a 1.5T clinical scanner (Philips Healthcare, Best, The Netherlands) on a calibrated phantom (Test Object 5, Eurospin II Test System, Diagnostic Sonar LTD). An in-plane resolution of 0.98 × 0.98 mm² and 10mm slice thickness (4 slices) was chosen. MTM T_1/B_1 mapping was applied using $TR_{11,12}$ = 115/750 ms, $TR_{21,22}$ = 105/202 ms, and $TR_{31,32}$ = 105/130 ms. An inversion recovery single shot TSE sequence (IR-TSE) served as an independent T_1 measurement (TR = 10s, IR delays [ms] = 5000, 2244, 1582, 1188, 906, 686, 506, 353, 221, 105). For better comparison, acquisition times were adjusted to approximately 5 min each. (3) The same MTM sequence was tested on a healthy volunteer using the same 1.5T scanner.

Results/Discussion: Predicted SNR via CRT and measured SNR for MTM T_1 mapping are shown in Fig. 1. IR-TSE T_1 map is shown in Fig 2a. MTM T_1 and B_1 maps are shown in Fig. 2b and 2c. A quantitative comparison is shown in Fig. 3. MTM T_1 phantom results were found in agreement with manufacturer specifications. The maximum deviation from the true value was 126 ms for MTM and 257 ms for IR-TSE. In comparison with IR-TSE, MTM results were superimposed by stronger noise. Maximum noise (1/SNR) was 11.5% for MTM and 1.4% for IR-TSE. *In vivo* MTM T_1 and B_1 results are shown in Fig. 3.

Conclusion: Efficient and accurate baseline T_1 and B_1 quantification is a pre-requisite for standardized and clinical DCE-MRI. This work presents an approach to fast and simultaneous T_1 and B1 mapping. Fitting of a theoretical model to a steady-state gradient-echo signal delivers independent T_1 and B_1 values. Adjusted in scan time (5min), MTM T_1 mapping was found to be more accurate than IR-TSE in calibrated phantom measurements at the expense of the SNR of the delivered T_1 maps. It is expected that the SNR of MTM can be enhanced by combining MTM with an EPI read out scheme.

References: [1] Padhani AR, JMRI 16 (2002), 407-422 [2] Treier R, et al., MRM 57 (2007), 568-576 [3] Voigt T. et al., ISMRM (2009), 4543. [4] Yarnykh VL., MRM 57 (2007), 192-200. [5] Kay SM. Fundamentals of Statistical Signal Processing: Estimation Theory. Englewood Cliffs, NJ: Prentice Hall; 1993.

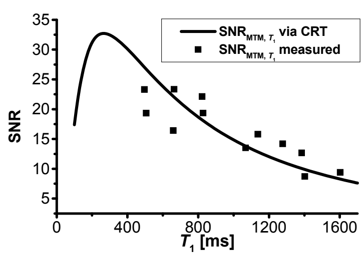

Fig. 1: Predicted and measured SNR of MTM T_1 mapping.

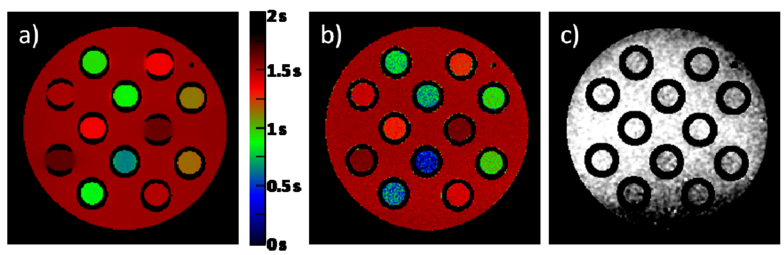

Fig. 2: IR-TSE T_1 map (a) and MTM T_1 map (b). MTM does also provide a B_1 map (c). T_1 values provided by manufacturer (left to right, top to bottom, T=23°C, units [ms]): 1278, 822, 664, 1385, 1137, 828, 496, 506, 1603, 1063, 1403, 659.

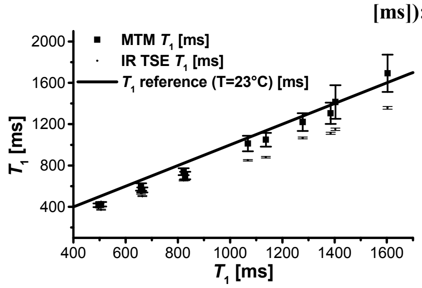

Fig. 3: quantitative comparison of T_1 mapping results from MTM and IR-TSE.

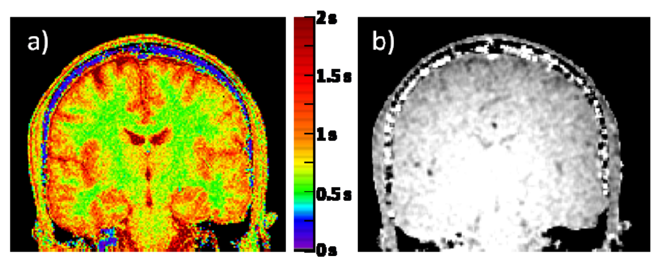

Fig. 4: *In vivo* MTM T_1 map (a) and B_1 map (b).