Modulated Repetition Time Look Locker (MORTLL): A Method for Rapid High Resolution Three Dimensional T1 Mapping

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Introduction: A modification of the Look-Locker (LL) technique (MORTLL for MOdulated Repetition Time LL) that enables high resolution T1 mapping over the physiologic range of intracranial T1 values is presented. Modifications include the use of a 3D balanced SSFP acquisition (for high SNR and resolution) and use of variable repetition time to allow effective full recovery of longitudinal magnetization. Using a simple fitting procedure, accurate T1 maps could be generated in a reasonable time. The technique is particularly well suited for imaging long T1 species.

Materials and Methods: *Sequence design:* Earlier segmented LL acquisition schemes have been used in the multi-slice 2D context. Typically, several LL phases are acquired to perform a three parameter fit. A correction is then applied to account for modulation of the longitudinal magnetization (M_z) as a result of the b-SSFP acquisition [1]. Through simulations [2], the modulation of M_z can be shown to become negligible with increasing excitation angle and increasing T1. In this limit, the two parameter IR equation $S = M_o(1 - 2e^{-TR/T1})$ can be used to obtain M_o and T1. Accordingly, our acquisition combines 3D slab acquisition (to provide higher SNR) with a low flip angle (10°) so that M_z recovery is minimally affected by flip angle. Each inversion pulse is followed by three phases of a 3D slice encoded (k_z step) single-shot (all k_y phases) b-SSFP acquisition. This is followed by a dead time during which M_z further recovers. To allow for full recovery of CSF M_z, a TR on the order of 18-20 seconds would be needed, resulting in a prohibitively long scan time. Therefore, we introduced a modulation of the dead-time, so that the TR of each shot varies from a shorter TR (e.g. 5s) at the edges of k_z-space to a very long TR (e.g. 18s) at the center of k_z space. Thus, the effective TR corresponds to full recovery of even long T1 species. Figure 1 shows the effect on slice profile of varying the TR from 5s to 18 s using a four term Blackman-Harris (B-H) window compared with constant TRs of 18s or 5s. *MRI experiments:* Phantoms with 2 different T1s and 6 healthy volunteers were imaged under an IRB approved protocol. A 3T Philips Achieva (release 2.5.3) scanner was used with all system and FDA related parameters calculated based on the minimum sequence time (i.e. dead time ≈ 0). Scan parameters were: FOV=23cm, $TI_{1,2,3}$ ≈ {300, 1050, 1800}ms; variable sequence TR with TR_{min}≈2.5s and TR_{max}=18s, b-SSFP acquisition TR/TE=3.7/1.59 ms, α=10° partial echo, partial Fourier encoding scan, resolution ≈ 0.9×0.9×1mm, #of slices=25, scan time≈5mins 20s. Only GM and WM T1 values were compared with IR-SE (TR=10s, scan time=8mins 50s) to keep scan time reasonable while a 8 phase low resolution (128×94 matrix) LL with constant TR=18s (CORTLL) and 3 parameter model with correction was also used for comparison [1].

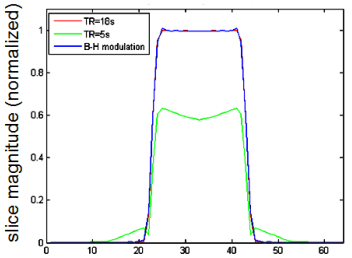

Fig. 1. Comparison of simulated image profile along Z using constant TR and modulated TR

Table 1. Measured T1 values of phantoms in ms

	MORTLL	CORTLL	IR-SE
long T1 phantom	2869 ± 64	2813 ± 116	2846 ± 9
short T1 phantom	969 ± 29	880 ± 13	963 ± 5

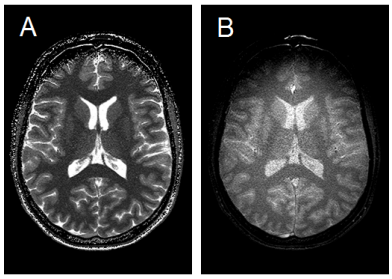

Fig. 2. Computed T1 ***(A)*** and proton density ***(B)*** maps

Results: In phantoms (Table 1), comparison of CORTLL and MORTLL to IR-SE (gold standard) showed comparable T1 values for the long T1 phantom but a considerably lower value (8.6%) for 8phase CORTLL with the short T1 phantom. In volunteers, the T1 map was robust to errors due to B1 inhomogeneity which are entirely captured in the M_0 map *(see Fig 2)*. Furthermore, the T1 maps using the three techniques are qualitatively similar *(Fig. 3)* Computed values of T1 for GM, WM and CSF across the six volunteers *(Fig.4)* are comparable, although there is a 7.8% difference in CSF T1 between CORTLL and MORTLL. T1 values for GM and WM measured by MORTLL was within 3% of that measured by IR-SE, while CORTLL underestimated these values by 8% and 10%. For two volunteers, T1 maps were also obtained with constant TR and the 3 phase acquisition scheme. The values were found to differ by a mere 1%, 0.95% and 0.5% in WM, GM and CSF, respectively. Average values for WM, GM and CSF were 755±10.1ms, 1202±9.4ms and 4482±71ms, respectively.

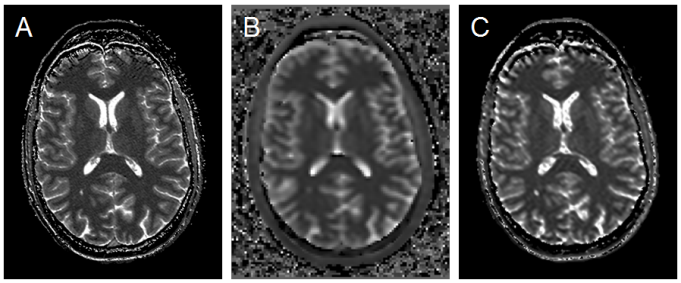

Figure 3. Computed T1 maps obtained with MORTLL ***(A)*** IR-SE ***(B)*** and CORTLL.

Discussion: A new technique for accurately mapping T1 values in the brain has been described here. The use of 3D b-SSFP and a low flip angle minimally perturbs the IR curve, thus a 3 phase LL with 2 parameter fit may be used. Simulations show that for α=10°, the 3 phase LL with 2 parameter fit for TIs used results in errors of 0.46%, 0.44% and 0% for WM, GM and CSF respectively. This obviates the need for a larger number of LL phases (which compromises resolution and/or scan time) as used in previous works [1]. In addition, the IR curve is not influenced by B1 inhomogeneity, so that the T1 maps are B1 independent, an advantage over dual flip angle T1 mapping techniques which require B1 mapping for accuracy [3]. Modulation of TR [4] increases scan efficiency by about 50% and allows the effective TR to be tailored to the maximum T1, making the technique suitable for measuring long T1 species. Worst case estimates on error (due to modulated TR) using simulations with acquired data show the error to be less than 6%. The T1 values obtained in WM, GM and CSF are comparable to those obtained using IR-spin echo imaging (gold standard) as well as 8 phase LL with a three parameter model and correction [1].

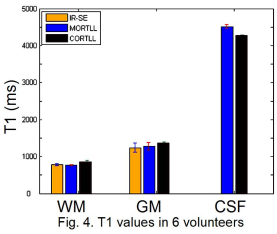

Fig. 4. T1 values in 6 volunteers

References: [1] P. Schmitt et al. MRM, 2004:51:661-667. [2] B. Hargreaves et al. MRM, 2001:46:149-158. [3] S. Deoni. JMRI, 2007:26:1106-1111. [4] J. Coleman et al. US Patent 5239266, 1993.