Accurate and precise T1 relaxometry with reduced data acquisition requirements

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Introduction: Most robustly, T1 is quantified by single-point sampling of inversion times (TI), because RF and static field offsets only scale the magnetization-time curve. Also, SNR is high and cardiac gating is feasible, though scan times are long. Multi-point methods (ie. Look-Locker), which follow an inversion pulse with a series of small-tip angle excitations of angle θ and imaging intervals of constant duration ΔT, accelerate T1 quantification, but SNR is low and cardiac gating is infeasible. In addition, the Look-Locker correction for RF sampling (Eq. 1: $1/T1 = 1/T1^* + \log(\cos(\theta))/\Delta T$) inherently sensitizes multi-point T1 to RF field effects. In this study, we identify data acquisition parameters which reduce both the precision (σT1) and accuracy of single and multi-point T1 to within target thresholds (ie. within 5% of truth), towards an overall goal of robust tissue characterization in vivo. Spiral imaging reduces the overall scan time for a given parameter set. We evaluate bias in multi-point T1 from RF tuning, associated with the slice profile of well-tuned systems and/or the RF sampling correction in mis-tuned systems (because knowledge of θ is assumed). Furthermore, a T1-based strategy for θ measurement and correction is presented, based on the equalization of Eq. 1 at variable ΔT.

Methods: The basic pulse sequence consists of an adiabatic inversion pulse, with subsequent spectral-spatial pulse and spiral imaging gradients (Foltz, JCMR, 2006). For multi-point acquisition, the small-tip and imaging sequence is applied repetitively (variable delay added to modulate ΔT). For single-point acquisition, variable delays are introduced between the sequence initiation signal and the inversion, and between the inversion and the adiabatic pulse. The summation of delays is constant across TI to ensure TI-dependent nuclear re-polarization. To further regulate the equilibrium polarization, three CHESS pulses (one 90° RF excitation and spoiler gradient per CHESS pulse) are applied sequentially in time following the final data acquisition. Every even-numbered sequence iteration is acquired without an adiabatic inversion, so that the addition of signals chops out the additive T1 recovery term. T1 is then quantified on a per region-of-interest (ROI) basis as the time constant of a monoexponential decay, via weighted least-squares fitting.

All measurements used MnCl2-doped water (500ml, 0.09mM, T1~1050ms) and the head or body coil of a 1.5 Tesla GE Signa. The relations between σT1, SNR, and the TI combination were evaluated via Monte Carlo simulation (to constant longest TI of 960ms). Sequential analyses of base single-point and multi-point data sets over the same range of TI combinations but variable ROI size provided model validation, because the per ROI SNR is the product of the per voxel SNR and the square root of the number of independent voxels. T1 bias associated with the slice profile were simulated by numerically solving the Bloch equations at variable θ, for both 'soft' (maximum through-slice θ of 21° for mean θ of 15°) and 'hard' (maximum through-slice θ of 17.6° for mean θ of 15°) spectral-spatial pulses. The mean θ was offset to simulate RF mis-tuning effects on the efficacy of the RF sampling correction. High SNR multi-point T1 maps were equalized for θ of 8, 15, and 22° and ΔT of 80 and 120ms, and maps of θ were extracted.

Results: σT1 is more strongly dependent on the per ROI SNR, rather than the number of TI (Fig a). All strategies reduced σT1 to 5% when the SNR is 60, and to 2.5% when the SNR is 120. More dense time sampling may alleviate SNR needs for high σT1. With well-tuned RF, simulated T1 varied by 3.5% and 2.3% for θ between 5 and 25° using 'soft' and 'hard' RF respectively, while experimental T1 varied by less than 2.2%. RF mis-tuning dominated multi-point T1 bias (Fig b,c). Preservation of the T1 bias to within 5% can be achieved for ΔB1 to approximately 15% by selection of $\theta \leq 15°$ and $\Delta T \geq 240$ms. Comparatively, single-point T1 varied by less than 1.5% for RF field offsets between ±30%. Fig. d demonstrates the feasibility of RF field mapping using spiral T1, including a slight positive bias which may reflect erroneous scaling between the tuning and slice-selective pulses.

Discussion: Single-point methods achieve σT1 targets readily because SNR is high, while multi-point methods achieve lower σT1 because of the flip angle scaling with other scan parameters held constant. Furthermore, RF effects can dominate multi-point T1 bias, unless slice profiles are 'harder' and θ and ΔT are selected appropriately, while single-point methods are inherently insensitive to the RF field. At constant averaging, single-point methods provide longer imaging times because scan duration scales with the number of TI, however multi-slice acquisition is feasible and without scan time penalty. Thus, multi-point methods accelerate imaging time, but only when the number of slices is greater than the number of TI. Furthermore, our single-point method can preserve a constant cardiac phase across TI (images not shown), while multi-point methods generate images at regularly-spaced cardiac phases, and thus cannot sensitize to pressure-sensitive tissue characteristics including the myocardial bood volume.

Conclusions: These results suggest that one may consider precise, accurate, and time-efficient T1 quantification in vivo. The methodology will facilitate tissue characterization in vivo, encompassing the quantification of intra-vascular and intra-cellular volumes (based on steady-state distributions of contrast agents) as well as multi-nuclear quantification of pO2. Furthermore, the equalization of T1 measurements at variable ΔT provides a novel method for B1 quantification, though its practicality is unclarified.

Figure: (a) T1 precision as a function of SNR and TI sampling; (b, c) multi-point T1 bias from RF mis-tuning at variable θ but constant ΔT of 120ms, and at constant θ of 15° but variable ΔT; and (d) T1-based maps of the B1 field for θ = 8, 15, and 22°.