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Optimization of sparse phase encodings for variable repetition-delay turbo-spin echo (TSE) T_1 measurements for preclinical applications ☆

DongKyu Lee, Sohyun Han, HyungJoon Cho *

Department of Biomedical Engineering, Ulsan National Institute of Science and Technology, Ulsan, Republic of Korea

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ABSTRACT

A variable repetition-delay (TR) spin echo sequence with repeated refocusing pulses, i.e., a variable TR turbo-spin echo (TSE), provides an attractive means of acquiring an accurate T_1 map information that is free from gradient echo based artifacts such as magnetic field inhomogeneities particularly for ultra-high field (at 7T and above) preclinical applications. However, the applicability of multi-slice TSE sequences is often limited by signal distortion from T_2 relaxation due to echo-train acquisitions for short T_2 tissues, inter-slice cross talks and magnetization transfer (MT) from repetitive slice-selective 180° pulse, and extended scan times with multiple TR excitations. These TSE shortcomings are difficult to remedy for preclinical applications, where small sizes of target organs usually limit the slice-gap control with restricted parallel imaging capabilities. In this study, compressed-sensing-assisted turbo-spin echo (CS-TSE) acquisitions for variable TR T_1 measurements at 7T preclinical scanner were implemented to reduce the echo-trains by sparse phase encodings. Following the sparse signal simulation and sampling scheme optimization, the measured T_1 values from CS-TSE and TSE were compared for phantoms, ex vivo, and in vivo subjects. The phantom T_1 values from CS-TSE and TSE were identical to those from the inversion recovery spin echo. For both ex vivo and in vivo multi-slice T_1 mapping, the shortened echo-trains of CS-TSE relieved the T_2 relaxation, reduced the inter-slice interferences of multi-slice acquisition, and made room for additional slice encodings while maintaining a shorter scan time than the conventional TSE at the expense of local image smoothness from CS regularizations.

1. Introduction

Mapping the absolute in vivo magnetic resonance (MR) T_1 information facilitates accurate and precise evaluations of tissue environments and provides non-invasive physiological surrogates [1–3]. Although inversion recovery [4] and saturation recovery [5] spin echo pulse sequences have been considered to be standard T_1 measurement methods, the lengthy scan times required to obtain them restrict their practical usage in in vivo applications.

To increase the acquisition speed of T_1 mapping, fast readout methods are commonly used. These methods include multiple excitations with single-inversion Look–Locker techniques [6–8] and multiple flip angles with steady-state pulse sequences [9–11]. However, these sequences typically suffer from image distortion artifacts and inaccuracy in absolute T_1 value estimation due to gradient echo based readouts and flip angle dependent multiple excitations, particularly as B_0 inhomogeneities and corresponding radio frequency (RF) increase [12] for ultra-high field (at 7T and above) preclinical scanners.

For ultra-high field preclinical applications, where the RF power deposition is noncritical and B_0 inhomogeneities are common, the spin echo sequence with repeated refocusing pulses, i.e., the turbo-spin echo (TSE) sequence [13], is used extensively for T_1 measurement with variable repetition-delay (TR) and robust signal behavior with a straightforward fitting model. Nevertheless, the applicability of the TSE sequence in the preclinical scanner is often hampered by (a) signal distortion from T_2 relaxation due to echo-train acquisitions for short T_2 tissues, (b) inter-slice cross talks and magnetization transfer from repetitive slice-selective 180° pulse, and (c) extended scan times with multiple TR excitations [14]. These particular TSE shortcomings are difficult to remedy for preclinical applications, where small sizes of target organs usually limit the slice-gap control. Parallel imaging capabilities can reduce these limitations; however, it may not be a suitable option

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* Corresponding author at: Unist-gil 50 (100 Banyeon-ri), Eonyang-eup, Uljugun, Ulsan Metropolitan City 689-798, Republic of Korea.
E-mail address: hjcho@unist.ac.kr (H. Cho).

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for preclinical applications owing to the relatively small field of view (FOV) and high hardware cost with limited channels.

Recently, the scan time of multidimensional MR imaging has been significantly reduced via compressed sensing (CS) assisted sparse sampling strategies [15–17]. Various studies have been undertaken to characterize the effects of accelerating the MR parameter mapping [18–22]. Consequently, the application of such sparse sampling strategies toward multi-slice T_1 mapping with variable TR TSE acquisitions could be a cost-effective solution for preclinical scanners, potentially synergistic with parallel imaging. However, to the best of our knowledge, systematic and quantitative characterizations of the effects and benefits of reducing echo-trains, i.e., a turbo factor (TF), with CS sampling schemes to multi-slice T_1 mapping have not yet been studied.

In this study, we investigated the benefits of replacing the TF with CS-assisted sparse sampling in variable TR TSE acquisitions for preclinical T_1 measurements. The effects of standard sparse sampling and CS reconstruction were examined and optimized for various concentrations of Gd-DOTA (DOTAREM) phantoms, ex vivo, and in vivo T_1 measurements at 7T.

2. Methods

2.1. T_1 fitting procedures

By adopting a centric echo order that minimizes the T_2 decaying effects, the signal expression of TSE acquisition [23] can be generalized at a specific TF:

$$S_{TSE} = S_0\left\{1 + \sum_{n=1}^{TF} 2(-1)^n \exp((-TR + TE(2TF - n + 1)/2)/T_1) + 2(-1)^{TF-1}\exp(-TR/T_1)\right\}, \quad (1)$$

where S_0 is the proton spin density, and TE is the echo time (echo spacing) between the initial 90° pulse and the first echo. The summation term in the signal equation represents the repeated refocusing pulses that continuously reverse the longitudinal magnetization, which effectively reduce the recovery time of the longitudinal magnetization [24].

The inversion recovery spin echo (IRSE) data were fitted using a signal equation [23].

$$S_{IRSE} = S_0\{1 - 2\exp(-TI/T_1) + 2\exp(-(TR - TE/2)/T_1) - \exp(-TR/T_1)\}, \quad (2)$$

where TI is the inversion time. All of the T_1 fitting procedures were performed using the nonlinear least squares method in the MATLAB curve-fitting toolbox.

2.2. Sampling scheme optimization

Eight Gaussian-weighted random undersampling schemes were generated while maintaining incoherency along the TR direction by varying the fixed number of phase encodings in the low-frequency region of the k-space. The length of the fixed center region was set at 59, 49, 39, 31, 25, 19, 11, and 5 for each scheme. Moreover, the maximum priority was assigned to the fixed regions during the generation process of weighted random sampling, which is described by Efraimidis and Spirakis [25]. The single Gaussian-weighted ($\sigma^2 = 0.125$) approach was used for the weighted random distribution of the remaining part.

Because undersampled k-space signals of CS-TSE acquisitions cannot be obtained from retrospective subsampling of fully sampled TSE signals, the k-space signals for the eight sampling schemes were calculated using the following equation:

$$S(k_x, k_y) = \sum_{y=1}^{FOV_y}\sum_{x=1}^{FOV_x} \rho(x,y) f_{T1}(x,y) f_{T2}(x,y) \exp(-i2\pi(k_x x/FOV_x + k_y y/FOV_y)), \quad (3)$$

where $f_{T1}(x,y)$ is a function corresponding to the T_1 recovery described by Eq. (1) and $f_{T2}(x,y) = \exp(-TF \times TE/T_2(x,y))$ is used for the T_2 decay of echo-trains. The T_1 and proton density values used for the simulation were obtained from various inversion time IRSE acquisitions with the same imaging parameters used in the ex vivo kidney experiment, and the T_2 map was evaluated by performing a multi-slice multi-echo experiment (TR = 7500 ms, echo spacing = 8 ms, and number of echo images = 32). With these maps, k-spaces of TR-varying CS-TSE acquisitions were simulated under the same imaging conditions used in the ensuing ex vivo and in vivo experiments. After the image reconstruction of the simulated k-spaces, the T_1 maps of each scheme were generated. The quality of the simulated T_1 maps was evaluated by assessing their concordance correlation coefficient (CCC) values [26] based on the reference T_1 map from the IRSE. By comparing the normalized CCC values, the optimized sampling scheme for each TF was determined. Following the acquisition of the optimal sampling scheme with the maximum CCC values, the number of necessary TR steps was investigated with logarithmic time spacing for all cases.

For CS reconstruction, the following constrained optimization problem was solved with the sequence of images in the TR direction (s):

$$\begin{aligned} &\text{minimize}\|\Psi s\|_1 \\ &\text{subject to}\|Fs - k\|_2 < \varepsilon, \end{aligned} \quad (4)$$

where Ψ denotes the linear operator that transforms s into a sparse representation (2D wavelet transform and 1D Fourier transform in the x–y and TR directions, respectively). F is the undersampled Fourier operator, k is the acquired k-space data, and ε controls the fidelity of the CS reconstruction to the measured data, which is usually set below the noise level of the k-space [27,28].

The reconstruction was performed using a software developed in-house with MATLAB R2015a (MathWorks, Natick, MA, USA). Two external packages were also used: spgl1 v.1.8 [29] and Wavelab v.8.02 [30] for the l1-norm minimization and application of the wavelet transform into images, respectively. The software was executed on a personal computer with an Intel Core i7-5960X 3.00 GHz central processing unit and 64 GB of memory running on Windows 7 operating system.

2.3. T_1 map reconstruction simulation at multiple T_2 values and acceleration factors

To differentiate the degrees of T_1 map quality degradation between sparse sampling and T_2 signal decaying of the echo-trains in the CS-TSE acquisition, the k-space signals for TSE and CS-TSE were simulated with variously scaled T_2 maps (original T_2 map × 1/4, 1/3, 1/2, 1, 2, 3, and 4) and the same simulation conditions for the sampling scheme optimization. After the CS image reconstruction of the simulated k-spaces, the T_1 maps of TSE and CS-TSE were evaluated. Then, the corresponding CCC values were calculated and compared with varying T_2 values to investigate the benefits of replacing echo-trains with sparse samplings to minimize uncompensated T_2 multi-echo decays.

2.4. Phantom experiment

All of the experiments were performed on a Bruker 7T MRI scanner. A 40 mm transmit/receive volume coil was used in the phantom experiment.