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Quantitative T_2 measurement of a single voxel with arbitrary shape using pinwheel excitation and CPMG acquisition

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Abstract

Objective—The aim of this study is to present a new approach for making quantitative single-voxel T_2 measurements from an arbitrarily shaped region of interest (ROI), where the advantage of the signal-to-noise ratio (SNR) per unit time of the single-voxel approach over conventional imaging approach can be achieved.

Materials and methods—Two-dimensional (2D) spatially selective radiofrequency (RF) pulses are proposed in this work for T_2 measurements based on using interleaved spiral trajectories in excitation k -space (pinwheel excitation pulses), combined with a summed Carr—Purcell Meiboom—Gill (CPMG) echo acquisition. The technique is described and compared to standard multi-echo imaging methods, on a two-compartment water phantom and an excised brain tissue.

Results—The studies show good agreement between imaging and our method. The measured improvement factors of SNR per unit time of our single-voxel approach over imaging approach are close to the predicted values.

Conclusion—Measuring T_2 relaxation times from a selected ROI of arbitrary shape using a single-voxel rather than an imaging approach can increase the SNR per unit time, which is critical for dynamic T_2 or multi-component T_2 measurements.

Keywords

T_2 measurement; Arbitrary shape localization; Interleaved spiral trajectory in k -space; CPMG

Introduction

Quantitative T_2 measurement traditionally involves acquisition of a conventional single-slice image with the Carr—Purcell [1] and Meiboom—Gill [2] (CPMG) scheme. A region of interest

(ROI) of arbitrary shape is then delineated from the obtained image, and the averaged image magnitude as a function of echo time is analyzed to compute T_2 or a spectrum of T_2 values. This imaging approach takes a long time and results in a relatively low signal-to-noise ratio (SNR) due to the necessarily small voxel sizes. More rapid single-slice T_2 imaging, with constant SNR per unit time, has been realized with interleaved echo-planar acquisition incorporated into a multi-echo sequence [3]. Spiral readout with T_2 preparation before excitation was proposed for more efficient multislice T_2 measurements [4], and three-dimensional (3D) coverage T_2 measurements in a clinically acceptable timeframe has also been presented [5]. In all of these cases, the objective was to accelerate acquisition when visualization of one or more slices is necessary. In cases where T_2 from a specific ROI is necessary and visualization is not, a single-voxel approach is advantageous.

Similar to in nuclear magnetic resonance (NMR) spectroscopy, SNR per unit time of the single-voxel T_2 measurement has a potential gain factor of the square root of N_{ROI} (the total number of pixels contained within the ROI) over that of the averaged pixels in the ROI with a conventional imaging approach [6]. This SNR per unit time advantage has been demonstrated numerically for multicomponent relaxation analysis [7]. Single-voxel localization can be achieved using either outer volume saturation (OVS) to isolate a rectangular or cylindrical ROI with unperturbed longitudinal magnetization [8], or using direct excitation of a rectangular ROI [9–11]. The T_2 decay curve from the selected ROI can usually be sampled by the subsequent CPMG sequence within one repetition time (TR), but these methods are currently limited to very simple shapes. Thus, the need for spatial specificity has limited single-voxel multicomponent T_2 measurements to tissues with large homogeneous volumes, mostly in body applications (skeletal muscle [12–14] and breast [15]). For accurate quantification and interpretation, single-voxel T_2 measurements should be obtained from well-defined regions of single tissues.

Spatial localization with multidimensional spatially selective radiofrequency (RF) pulses [16–18] offers a potential solution to this problem, as these pulses can in principle be tailored to excite any arbitrarily shaped ROI. The aim of this study is to present a new approach for making quantitative single-voxel T_2 measurements, by using pinwheel excitation pulses [19] (interleaved spiral k -space trajectories) combined with a summed CPMG echo acquisition to acquire a T_2 decay curve from an arbitrarily shaped ROI within several TRs. The construction of the pulse sequence is presented. Results from both phantom and squirrel monkey brain in vitro are compared with standard multi-echo imaging methods and show the feasibility and utility of this technique for single-voxel T_2 measurements.

Methods

Spiral trajectory in k -space

As described in our recent paper [20], the desired excitation profile is chosen as an $N \times N$ matrix with a resolution of $\Delta x = \text{FOE}/N$, where the field of excitation (FOE) is the size of the two-dimensional (2D) domain over which the spatially selective RF pulse is defined. A constant-angular-velocity spiral trajectory was utilized with number of cycles, $N_C = N/2$, to sample the radial direction. Within each cycle of the spiral, $N \times \pi$ rotational angles are sampled to remove aliasing from the FOE. With a step duration Δt determined by the maximum gradient amplitude, the RF pulse length, T_{EX} (ms), is the product of the duration per cycle, $\Delta t \times N \times \pi$, with the number of cycles.

Ideally, a single-shot spiral trajectory will excite all the desired signals from the selected ROI; however, the accumulated phase caused by B_0 inhomogeneity during T_{EX} manifests as blurring in the excited ROI shape [21]. The point-spread function of a spiral trajectory is a ring of radius $2T_{\text{EX}} \times f_{\text{off}} \times \Delta x$, where f_{off} is the frequency shift from resonance in Hz. Similarly, Rahmer et

al. [22] have shown that the point-spread function is broaden when the excitation pulse duration is significant compared to T_2 (full-width at half-maximum line width of $0.49T_{EX}/T_2 \times \Delta x$). For both of these reasons, it is desirable to segment the excitation N_i interleaved subpulses, each with N_C/N_i cycles. The observed signals from each of the N_i acquisitions are then summed to produce the signal from only the desired 2D excitation profile. Figure 1a depicts the interleaved spiral k -space trajectory consisting of four two-cycle spirals. Figure 1b shows the weighted RF pulses and the corresponding gradients of the x and y channels for one interleaf. The outward-in spiral trajectories end at the center of the k -space and require no refocusing gradient. The phase of the interleaved spiral gradients is offset by an integral multiple of $2\pi/N_i$.

Pulse sequence

After an ROI is defined from a scout image, an FOE that sufficiently contains this ROI is chosen in order to shorten the excitation pulse length. This excitation matrix is smaller than the acquisition matrix of the scout image ($FOE < FOV$) but has the same resolution. To minimize the contamination artifact caused by aliasing from outside the FOE or imperfect k -space sampling within the FOE, OVS is employed before excitation to suppress all unwanted signals outside a rectangular box encompassing the ROI. Two cycles of a four-sideband saturation were executed, each using 2 ms adiabatic full-passage sech/tanh pulses with a bandwidth times pulse length product of 20, followed by dephasing gradients. Immediately after the interleaved spiral excitation, slice selection was achieved with the first refocusing pulse (optimized sinc 180 [23]) at $TE_1 = 5.5$ ms. Subsequent nonselective refocusing pulses ($90_y 180_y 90_x$ composite pulses) were surrounded by crusher gradients of alternating polarity and decreasing magnitudes [24] to remove signal from unwanted coherence pathways. Each echo of the T_2 decay curve consisted of N_{read} samples and resulted from averaging corresponding echoes from each interleave. The acquisition window of each echo ($N_{read}/\text{bandwidth}$) should be chosen to cover as much of the plateau of each echo as possible to increase the average number, but also to avoid significant T_2^* decay. The pulse sequence diagram is shown in Fig. 1c. ROI localization was verified by replacing the CPMG sequence with conventional 2DFT imaging sequence.

Hardware

All experiments were performed at 400 MHz on a 21-cm-bore, horizontal 9.4-T magnet (Varian Inc., Palo Alto, CA, USA) at the Vanderbilt University Institute of Imaging Science. The system was equipped with imaging gradients capable of generating 40 G/cm with switching times to full amplitude of 135 μ s. A 63-mm-diameter volume coil was utilized for both RF transmission and signal reception.

Phantom experiments

Data were collected from an aqueous solution of $MnCl_2$ phantom which was composed of a convolved tubing (1 mm diameter) filled with a higher concentration of $MnCl_2$ than the surrounding bath solution (see Fig. 2). T_2 measurements were compared between a conventional single-slice imaging technique and the proposed single-voxel technique. The imaging approach had a matrix of $N_{read} \times N_{phase} = 128 \times 128$ with $FOV = 25.6$ mm and slice thickness = 1 mm. The total acquisition time was approximately 13 min ($NEX = 2$, $TR = 3,000$ ms). Twenty-four echoes were sampled with 8.5-ms echo spacing. Three ROIs were chosen: one including only the tubing (5.6 mm^2 , $N_{ROI} = 139$), another one including only the bath region (3.4 mm^2 , $N_{ROI} = 85$), and the third ROI combining the first two regions (8.8 mm^2 , $N_{ROI} = 219$).

The single-voxel method was then used to measure T_2 from each of the three ROIs. An excitation matrix (48×48 , $FOE = 9.6$ mm) was defined within the scout images. The OVS pulses suppressed anything outside the square FOE box in the axial plain and a 1 mm slice was

selected by the first refocusing pulse. RF pulses and gradients of interleaved spirals pulses were generated with a step duration of $\Delta t = 8 \mu\text{s}$. $N_i = 8$ interleaves with three cycles per interleave were used and the excitation pulse length was 3.7 ms for each subpulse. The RF power was adjusted to give a roughly 30° nutation angle excitation to avoid significant nonlinearity in excitation. The maximum gradient strength required was 30 G/cm. The results of the excitation were imaged by using 2DFT encoding during acquisition (128×128 , FOV = 25.6 mm). The CPMG measurements kept the same parameters used by the imaging approach, except that $N_{\text{read}} = 64$ complexed points were recorded centered around each echo. One dummy scan was used before each interleaf to establish steady state. Accordingly it took 1.2 min for the T_2 measurement from each ROI.

In vitro experiments

An excised squirrel monkey brain, fixed in 4% paraformaldehyde, was used to compare the two techniques on a neural tissue sample (see Fig. 3). The imaging approach had a matrix of 128×128 with FOV = 40 mm and slice thickness = 1 mm. The total imaging time was approximately 26 min (NEX = 4, TR = 3,000 ms). Thirty-two echoes were sampled with 10-ms echo spacing. Again, three ROIs were chosen: white matter (18 mm^2 , $N_{\text{ROI}} = 184$), gray matter (45 mm^2 , $N_{\text{ROI}} = 461$), and a rectangular ROI which included the first two ROIs selected and some surrounding buffer solution (90 mm^2 , $N_{\text{ROI}} = 922$).

For the single-voxel approach, an excitation matrix (64×64 , FOE = 20 mm) was defined and the OVS box was a volume $10 \times 14 \times 1 \text{ mm}^3$, which in the axial plane was slightly bigger than the rectangular ROI. $N_i = 8$ interleaves with four cycles per interleave were used ($\Delta t = 4 \mu\text{s}$) and each sub-pulse's excitation pulse length was 3.3 ms with a roughly 30° nutation angle. The maximum gradient strength utilized was 29 G/cm. During the CPMG acquisition, $N_{\text{read}} = 64$ complex points were recorded for each echo and it took 2 min for each ROI (one dummy scan, NEX = 4, TR = 3,000 ms). As with the phantom measurements, the excitation profiles were imaged for validation (128×128 , FOV = 40 mm).

To further validate this method, an $8 \times 12 \text{ mm}^2$ ROI (approximately the same as the previous selected rectangular ROI), was localized by OVS alone. The interleaved spiral excitation pulse was replaced by a 2-ms adiabatic half-passage pulse to excite the rectangular ROI with 90° nutation. The T_2 measurement used NEX = 32, in order to match the total number of excitations [$N_i (= 8) \times \text{NEX}(= 4)$] used in the previous measurement.

Data analysis

All transverse magnetization decay curves were fitted with the sum of 128 decaying exponential functions with time constants ranging logarithmically between 10 and 300 ms [25]. Regularization was achieved with a minimum-energy constraint. Phantom and monkey brain data had different SNR, requiring different smoothing factors (phantom: $\mu = 0.004$; monkey brain: $\mu = 0.002$). The chi-square statistic (χ^2) of each T_2 spectral fit was calculated. Each component of every T_2 spectrum was characterized by its fractional area and mean T_2 value.

Both the phantom and monkey brain measurements (imaging, IM; single-voxel, SV) were performed with the same acquisition bandwidth (100 kHz) and receiver gain. Within one TR period, the standard deviation (SD) of the measured noise (before Fourier transform) was $\sigma = 6.7$ (arbitrary units). The SD of the total acquisition noise can be derived as

$$\sigma^{\text{IM}} = \sigma \cdot \sqrt{\text{NEX}^{\text{IM}}} \cdot \sqrt{N_{\text{read}}^{\text{IM}} \cdot N_{\text{phase}}^{\text{IM}}} \text{ for a single pixel of the imaging approach and}$$

$$\sigma^{\text{SV}} = \sigma \cdot \sqrt{\text{NEX}^{\text{SV}}} \cdot \sqrt{N_i^{\text{SV}}} \text{ for a single sample point of the acquired FID.}$$

The achieved SNR was calculated as the magnitude of the first echo $s(1)$ of the T_2 decay curve divided by the SD of the total acquisition noise and times the factor $\sqrt{N_{\text{ROI}}}$ for the imaging approach as $(S/N)^{\text{IM}} = s(1)^{\text{IM}} / \sigma^{\text{IM}} \cdot \sqrt{N_{\text{ROI}}}$, or times the factor $\sqrt{N_{\text{read}}^{\text{SV}}}$ for the single-voxel approach as $(S/N)^{\text{SV}} = s(1)^{\text{SV}} / \sigma^{\text{SV}} \cdot \sqrt{N_{\text{read}}^{\text{SV}}}$.

The SNR per unit time is the achieved SNR divided by $\sqrt{T_{\text{total}}}$ as $(S/N)_{\text{unit}} = (S/N) / \sqrt{T_{\text{total}}}$. It is proportional to $\sqrt{N_{\text{ROI}}} \cdot \sqrt{N_{\text{read}}^{\text{I}}}$ with the imaging approach and $N_{\text{ROI}} \cdot \sqrt{N_{\text{read}}^{\text{SV}}}$ with the single-voxel approach, a potential gain factor of $\sqrt{N_{\text{ROI}}}$ for the latter technique (assuming all other factors to be the same).

Results

Phantom experiments

The images in Fig. 2a, c, e are zoomed views of the phantom from the multi-echo image set (TE = 17 ms). The darker convolved tubing area has a shorter T_2 relaxation than the surrounding solution. The dashed square lines demarcate the FOE, and the solid lines demarcate the three selected ROIs (tubing, bath, and the combined region). The localization results from all three ROIs are shown in Fig. 2b, d, f. Adjacent to each image is the measured T_2 decay and the corresponding T_2 spectrum. The mean T_2 and fractional amplitude of each T_2 component are also shown. In all cases, the T_2 decay curves and corresponding spectra obtained using the SV technique (Fig. 2b, d, f) are similar to the corresponding curves and spectra obtained from the IM approach (Fig. 2a, c, e). The sampled amplitudes at the first echo (IM: 8.5 ms; SV: TE₁ + TE = 5.5 + 8.5 = 14ms) of the T_2 decay curves are listed in Table 1 for the three ROIs. The subsequent echo spacing was at 8.5 ms. Thus, the second echo was at 22.5 ms, so signal with a $T_2 = 10$ ms will have decayed to about 10% of its full amplitude (at TE = 0), which is well above the noise floor.

Table 1 also compares the achieved SNR per unit time using the two approaches for each ROI. The measured improvement factor of SNR per unit time is simply $\text{IF}_m = (S/N)_{\text{unit}}^{\text{SV}} / (S/N)_{\text{unit}}^{\text{IM}}$. When considering the additional signal relaxation of SV during the slice-selecting part of TE₁ (5.5 ms) before the first echo of the CPMG echo train, the smaller nutation angle of SV (30°), and the difference of N_{read} (IM: 128; SV: 64), the theoretical improvement factor of SNR per unit time of the employed single-voxel approach over the imaging approach is

$$\text{IF}_t = \exp(-\text{TE}_1/T_2) \sin(\theta^{\text{SV}}) \cdot \sqrt{N_{\text{read}}^{\text{SV}}/N_{\text{read}}^{\text{IM}}} \cdot \sqrt{N_{\text{ROI}}}.$$

In vitro experiments

The results of the squirrel monkey brain in vitro are shown in Fig. 3, presented similarly to the phantom data in Fig. 2. The white matter is dark compared to the surrounding gray-matter area, and the buffer surrounding the brain is the brightest signal. The dash-dot rectangular boxes demarcate the OVS used in the single-voxel measurements and the solid lines demarcate the ROIs. Analysis of the obtained T_2 spectra with the mean T_2 peaks and associated fractional distribution showed general agreement between IM and SV for each ROI. The observed values are also consistent with other studies of fixed brain tissue [26]. Table 2 shows the achieved SNR per unit time of the two approaches and shows that the measured improvement factors of SNR per unit time of SV over IM IF_m are close to the predicted values, IF_t . Figure 3g shows an image of the localization and the T_2 measurement result from the OVS alone, as described above. The OVS suppressed anything outside the rectangular ROI and the adiabatic pulse

excited everything within. The result of the localization of the rectangular ROI by our technique (Fig. 3f) is comparable with that by the OVS approach (Fig. 3g).

Discussion

This work has demonstrated a novel single-voxel approach for T_2 measurements. The results show good agreement between imaging and our method on multiple occasions, on a water phantom and excised brain tissue. The ideal gain of SNR per unit time of the single-voxel approach over the conventional imaging approach is proportional to the square root of the number of pixels contained in the ROI. For large ROIs with complex shapes, this method can significantly increase SNR and reduce the partial-volume effect compared to conventional single-voxel methods, which are limited to rectangular voxel shapes. Compared to imaging, the SNR gain is even more significant, albeit with the loss of the flexibility to study other regions of interest that may be present in a full image.

The accuracy of the localization depends on the choice of excitation pulse length. A short subpulse length ($T_{EX} \approx 3.5$ ms) was used in this work to allow measurement of relatively short T_2 signals (≈ 10 ms). Given the analogy between this spiral excitation trajectory and the imaging counterpart [21], the blurring due to T_2 decay during excitation is approximately 0.49×3.5 ms / 10 ms = 0.17 pixels (see the section on “Spiral trajectory in k -space”) [22]. Likewise, with an estimated 20 Hz resonance variation across the ROI, the blurring caused by B_0 inhomogeneity was approximately 2×3.5 ms $\times$ 20 Hz = 0.14 pixels.

In addition to pulse length sensitivity, our method is also uniquely sensitive to spoiling of unwanted signal compared to conventional multi-echo imaging. Although the calculated T_2 spectra from our new single-voxel sequence were close to the corresponding results from the imaging approach, some small sinusoidal waves during the T_2 decay curves were observed. This indicates signal from undesired coherence pathways due to the presence of inhomogeneous B_0 and B_1 fields. The employed crusher gradient pattern [24] may not be ideal for the complex spin distribution created by the interleaved spatially selective excitation pulses and further investigation will be necessary to determine the optimal spoiling approach. It is worth noting that T_2 spectral estimation is inherently ill-posed, and subtle changes in the decay curve through the effects of noise, artifact, or underlying T_2 distribution differences can have a substantial effect on the resultant T_2 distribution. This is particularly true when the T_2 spectrum contains several and/or closely spaced components, as is the case for the data shown in Fig. 3.

The achieved SNR per unit time with this technique is close but slightly lower than the theoretical values. The differences may be explained by excitation profile degradation by the B_0 and B_1 inhomogeneity. Especially for the B_1 effect, each subpulse of the excitation employed in this work has three (phantom) or four (in vitro) cycles per spiral interleave. These can only generate intact excitation profiles within the small-tip-angle regime [16], but increased B_1 inhomogeneity may lead to excess RF power in certain areas within the ROI, which would generate degraded excitation profiles. Signal loss of short- T_2 components during the excitation pulses may also contribute to the reduced SNR. To improve the SNR, the first echo can be collected immediately after the slice-selective refocusing, which would reduce the first echo time to TE_1 from $TE_1 + TE$. Some magnetization transfer effects from OVS pulses may also cause reduced SNR compared to the imaging approach.

Although the method was demonstrated on a small-animal scanner, it should also be feasible for application in a clinical setting. For a typical human scanner (4 G/cm maximum gradients with a 150 T/m/s slew rate), to excite an arbitrarily shaped ROI in an excitation matrix (64×64 , FOE = 256 mm), excitation pulses of $N_i = 8$ interleaved spirals with four cycles per

interleave are required ($\Delta t = 5 \mu s$) and each subpulse's excitation pulse length is about 4 ms. In fact, wherever the spiral trajectory has been utilized in imaging k -space successfully, it should in principle be suitable to excite k -space for the same application.

Our sequence localizes a selected 2D ROI during the excitation with a following slice selection for the third dimension. This method can be extended to excite a 3D arbitrarily shaped ROI directly with interleaved spiral pulses. The method can also be extended to study other MR properties of the selected ROI, such as T_1 , diffusion coefficients, or magnetization transfer.

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References

1. Carr HY, Purcell EM. Effects of diffusion on free precession in nuclear magnetic resonance experiments. *Phys Rev* 1954;94:630–638.
2. Meiboom S, Gill D. Modified spin echo method for measuring nuclear relaxation times. *Rev Sci Instrum* 1958;29:688–691.
3. Does MD, Gore JC. Rapid acquisition transverse relaxometric imaging. *J Magn Reson* 2000;147(1):116–120. [PubMed: 11042054]
4. Oh J, Han ET, Pelletier D, Nelson SJ. Measurement of in vivo multi-component T2 relaxation times for brain tissue using multi-slice T2 prep at 1.5 and 3 T. *Magn Reson Imaging* 2006;24(1):33–43. [PubMed: 16410176]
5. Madler, B.; Mackay, AL. Imaging Myelin: formation, destruction and repair. Vancouver, Canada: 2006. In-vivo 3D T2-relaxation measurements for quantitative myelin imaging; p. 72
6. Granot J. Selected Volume Spectroscopy (SVS) and chemical-shift imaging. A comparison. *J Magn Reson* 1986;66:197–200.
7. Graham SJ, Stanchev PL, Bronskill MJ. Criteria for analysis of multicomponent tissue T-2 relaxation data. *Magn Reson Med* 1996;35(3):370–378. [PubMed: 8699949]
8. Singh S, Rutt BK, Henkelman RM. Projection presaturation: a fast and accurate technique of multidimensional spatial localization. *J Magn Reson* 1990;87:567–583.
9. Ordidge RJ, Connelly A, Lohman JAB. Image-selected in vivo spectroscopy (ISIS)—a new technique for spatially selective NMR-spectroscopy. *J Magn Reson* 1986;66(2):283–294.
10. Frahm J, Merboldt KD, Hancike W. Localized proton spectroscopy using stimulated echoes. *J Magn Reson* 1987;72(3):502–508.
11. Bottomley PA. Spatial localization in NMR-spectroscopy in vivo. *Ann N Y Acad Sci* 1987;508:333–348. [PubMed: 3326459]
12. Saab, G.; Thompson, RT.; Marsh, GD. A technique for localized multicomponent T_2 measurement in skeletal muscle. Proceedings of the international society for magnetic resonance in medicine. Fifth Scientific Meeting and Exhibition; Vancouver, Canada. 1997; p. 1372
13. Saab G, Thompson RT, Marsh GD. Multicomponent T_2 relaxation of in vivo skeletal muscle. *Magn Reson Med* 1999;42:150–157. [PubMed: 10398961]
14. Saab G, Thompson RT, Marsh GD. Effects of exercise on muscle transverse relaxation determined by MR imaging and in vivo relaxometry. *J Appl Physiol* 2000;88:226–233. [PubMed: 10642385]
15. Graham SJ, Bronskill MJ. MR measurement of relative water content and multicomponent T-2 relaxation in human breast. *Magn Reson Med* 1996;35(5):706–715. [PubMed: 8722822]
16. Pauly J, Nishimura D, Macovski A. A k -space analysis of small-tip-angle excitation. *J Magn Reson* 1989;81(1):43–56.
17. Pauly J, Nishimura D, Macovski A. A linear class of large-tip-angle selective excitation pulses. *J Magn Reson* 1989;82(3):571–587.

18. Hardy CJ, Cline HE. Spatial localization in 2 dimensions using NMR designer pulses. *J Magn Reson* 1989;82(3):647–654.
19. Hardy CJ, Bottomley PA. P-31 spectroscopic localization using pinwheel NMR excitation pulses. *Magn Reson Med* 1991;17(2):315–327. [PubMed: 2062206]
20. Qin Q, Gore JC, Does MD, Avison MJ, DeGraaf RA. 2D Arbitrary shape selective excitation summed spectroscopy (ASSESS). *Magn Reson Med* 2007;58(1):19–26. [PubMed: 17659614]
21. Bornert P, Aldefeld B. On spatially selective RF excitation and its analogy with spiral MR image acquisition. *Magn Reson Mater Phy* 1998;7:166–178.
22. Rahmer J, Bornert P, Groen J, Bos C. Three-dimensional radial ultrashort echo-time imaging with T-2 adapted sampling. *Magn Reson Med* 2006;55(5):1075–1082. [PubMed: 16538604]
23. Mao J, Mareci TH, Scott KN, Andrew ER. Selective inversion radiofrequency pulses by optimal control. *J Magn Reson* 1986;70:310–318.
24. Poon CS, Henkelman RM. Practical T_2 quantitation for clinical-applications. *J Magn Reson Imag* 1992;2(5):541–553.
25. Whittall KP, Mackay AL. Quantitative interpretation of NMR relaxation data. *J Magn Reson* 1989;84(1):134–152.
26. D’Arceuil HE, Westmoreland S, de Crespigny AJ. An approach to high resolution diffusion tensor imaging in fixed primate brain. *Neuroimage* 2007;35(2):553–565. [PubMed: 17292630]

Fig. 1.

a Pinwheel excitation k -space trajectory consisting of four interleaved spirals each with two cycles; **b** the RF pulse (solid line, B_{1x} ; dashed line, B_{1y}) and accompanying gradients (solid line, G_x ; dashed line, G_y , double dashed line, G_z) for one interleave; **c** pulse sequence diagram for T_2 measurement using pinwheel excitation and CPMG acquisition. Each interleave consists of five blocks, OVS, pinwheel excitation, slice-selective refocusing, CPMG acquisition, and saturation

Fig. 2. Validation of the single-voxel approach on ROIs of a phantom. The *dashed square line* indicates the FOE (48 × 48, FOE = 9.6 mm), zoomed from the larger matrices (128 × 128, FOV = 25.6 mm). T₂ measurements are compared between multi-echo imaging approach (a, c, e) and single-voxel approach (b, d, f) for each of the three ROIs (solid lines): one within the tubing, one in the bath, and the third close to a combination of the first two ROIs. Images (a, c, e) are from the same multi-echo image set. Images (b, d, f) are the localization results for each ROI, respectively. Both obtained T₂ decay curves (TE / NE = 8.5 ms / 24) and correspondingly estimated T₂ spectra are shown. The normalized chi-square statistic (χ^2) of

each T_2 spectral fit is shown. The estimated mean T_2 values and their fractional amplitude are displayed next to each peak for both the image (IM) and single-voxel approach (SV)

Fig. 3.

Validation of the single-voxel approach on ROIs from a squirrel monkey brain in vitro. The dashed square line indicates the FOE (64 × 64, FOE = 20 mm), zoomed from the larger matrices (128 × 128, FOV = 40 mm). T_2 measurements are compared between the multi-echo imaging (a, c, e) and single-voxel approach (b, d, f) for each of the three ROIs (solid lines): one within the white matter, one within the gray matter, and the third rectangular ROI including the first two ROIs. The dashed dot rectangular lines indicate the OVS box. Images (a, c, e) are from the same multi-echo image set. Images (b, d, f) are the localization results for each ROI separately. Both obtained T_2 decay curves (TE / NE = 10 ms / 32) and the correspondingly estimated T_2 spectra are shown. A rectangular ROI was also localized using the OVS approach

(g) to compare with the 2D selective excitation approach (f). The normalized chi-square statistic (χ^2) of each T_2 spectral fit is shown. The estimated mean T_2 values and their fractional distribution are displayed next to each peak for both the image (IM) and single-voxel approach (SV)

Table 1
SNR per unit time analysis of phantom results compared between the imaging (IM) and single-voxel approaches (SV)

ROI	N_{ROI}	$s(1)^{IM}$	$(S/N)^{IM}$	$(S / N)^{IM}_{unit}$	ROI	$s(1)^{SV}$	$(S/N)^{SV}$	$(S / N)^{SV}_{unit}$	IF_m	IF_t
a	139	57,200	561	157	b	642	193	176	1.12	2.42
c	85	84,000	644	180	d	1,058	318	290	1.61	2.48
e	219	67,000	824	230	f	1,662	500	456	1.98	3.50

ROI definitions (b, c, ...) indicate correspondence to frames of Fig. 2

$\phi^{IM} = 90^\circ, NEX^{IM} = 2, N_{read}^{IM} = N_{phase}^{IM} = 128, \sigma^{IM} = 1.203, T_{total}^{IM} = 1.28min$
 $\phi^{SV} = 30^\circ, NEX^{SV} = 2, N_i^{SV} = 8, \sigma^{SV} = 26.6, N_{read}^{SV} = 64, DS^{SV} = 1, T_{total}^{SV} = 1.2min$

Table 2 SNR per unit time analysis of in vitro results compared between the imaging (IM) and single-voxel approaches (SV)

ROI	N_{ROI}	$s(I)^{IM}$	$(S/N)^{IM}$	$(S / N)^{IM}_{unit}$	ROI	$s(I)^{SV}$	$(S/N)^{SV}$	$(S / N)^{SV}_{unit}$	IF_m	IF_i
a	184	187,000	1,490	295	b	5,300	1,128	797	2.70	3.87
c	461	279,000	3,520	696	d	19,600	4,170	2,949	4.24	6.13
e	922	257,000	4,585	906	f	37,500	7,979	5,642	6.23	8.67
					g	91,200	19,404	13,721	15.14	17.34

ROI definitions (b, c, ...) indicate correspondence to frames of Fig. 3

$\phi^{IM} = 90^\circ$, $NEX^{IM} = 4$, $N_{read}^{IM} = N_{phase}^{IM} = 128$, $\sigma^{IM} = 1$, 702 , $T_{total}^{IM} = 25.6min$
 $\phi^{SV} = 30^\circ$ (b, d, f), $\phi^{SV} = 90^\circ$ (g), $NEX^{SV} = 4$, $N_i^{SV} = 8$, $\sigma^{SV} = 37.6$, $N_{read}^{SV} = 64$, $Ds^{SV} = 1$, $T_{total}^{SV} = 2min$