Selection of T2 components from segmented k-space multiecho data: improving efficiency of T2 relaxometry for myelin quantification.

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Introduction: Analysis of multiexponential T_2 relaxation decay of the MR signal from a voxel may provide important information about water compartmentation within the voxel. For example, the ratio of a short T_2 component (T_2=10-20 ms) associated with intramyelinic water to the total water signal, or myelin water fraction (MWF), has been used to quantify myelin content in the white matter (WM) [1]. The traditional approach is to sample the T_2 decay curve using a CPMG echo train (usually 32 echo points) and fit the data to a multiexponential model using a nonnegative least squares algorithm. This results in long acquisition times (25 minutes per slice [1]). We propose a new method to improve efficiency of data collection for T_2 spectrum analysis in which only a part of the CPMG echo train is used for T_2 decay sampling, while the rest of the echoes are utilized for *k*-space segmentation. The method relies on linear combination techniques [2,3], which apply linear filters to multiecho data for T_2 component selection.

Theory and Methods: The example CPMG pulse sequence combining T_2 decay sampling and *k*-space segmentation is shown in Fig 1a. The echo train is split into two subsets consisting of odd/even echoes for *k*-space segmenting, each segment having different T2 weighting. Our method for selection of T_2 components from the data with nonuniform T_2-weighting is illustrated in Fig. 1b and given by:

$$I^c = \mathrm{Re}\left(\sum_{i=1}^{N/2} c_i^{(1)} f_{2i-1} e^{-i\hat{\phi}_{2i-1}} + \sum_{i=1}^{N/2} c_i^{(2)} f_{2i} e^{-i\hat{\phi}_{2i-1}}\right).$$

Complex images given by $f_{2i-1} = \mathrm{IFFT}\{b(\Delta)s_{2i-1}\}$, $f_{2i} = \mathrm{IFFT}\{[1-b(\Delta)]s_{2i}\}$ are combined using coefficients $c_i^{(1)}, c_i^{(2)}$. Here, the box function $b(\Delta)$ of width Δ models *k*-space segmenting, and s_i is full *k*-space at echo time TE_i. In order to avoid signal cancellation due to image phase differences between different echoes, the low resolution image phase $\hat{\phi}$ is estimated from the odd echo datasets and used for phase correction of both odd and even echo datasets. The coefficient sets $c^{(1)}, c^{(2)}$ are tuned to extract specific T_2 components from odd and even echoes. The corresponding filter responses given by $A^{(1)}(T_2) = \sum_i c_i^{(1)} e^{(-TE_{2i-1}/T_2)}$, $A^{(2)}(T_2) = \sum_i c_i^{(2)} e^{(-TE_{2i}/T_2)}$ approximate each other to avoid Gibbs ringing in the final component image I^c, which arise due to *k*-space segmentation. The coefficients were found as the ones maximizing

$$SNR = K\sum_{i=1}^{N/2}\left(c_i^{(1)} e^{(-TE_{2i-1}/T_{2,0})} + c_i^{(2)} e^{(-TE_{2i}/T_{2,0})}\right) \Bigg/ \sqrt{\sum_{i=1}^{N/2}\left[\left(c_i^{(1)}\right)^2 + \left(c_i^{(2)}\right)^2\right]},$$

subject to the constraints given in Table 1, which were derived from the ones given in [3]. The multiecho data (48 echoes, TE_1=7 ms, echo spacing 7 ms, T_R=5 s, quadratic image phases) was simulated using fuzzy model of brain with MS lesions from BrainWeb database (www.bic.mni.mcgill.ca/brainweb/). WM was modeled by two water compartments, the first one with T_2 =20 ms, the other one with T_2=80 ms. Other MR tissue parameters were typical for 1.5 T. Gaussian noise was added in quadrature to the multiecho data. Image SNR was estimated in ROI drawn in WM regions. The results are shown in Table 2 and Fig. 2.

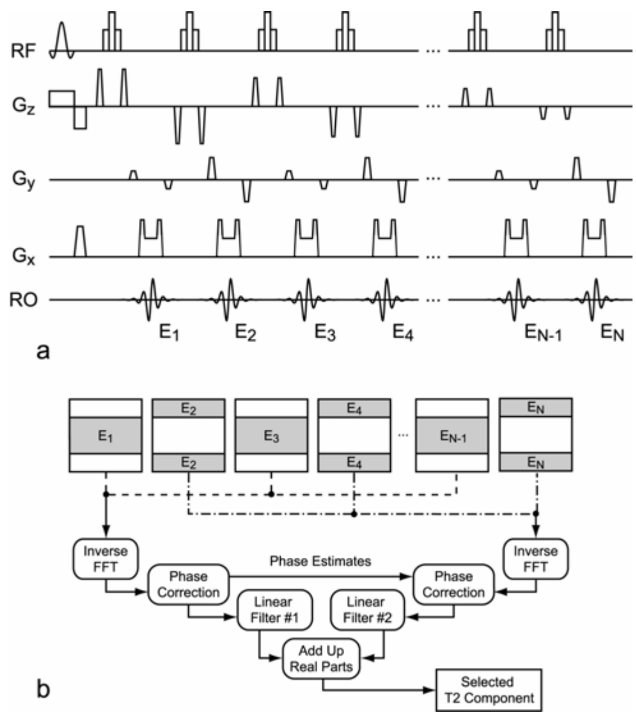

Figure 1. Selection of T_2 spectrum components from segmented *k*-space multiecho data. **a:** Pulse sequence. **b:** Flowchart of the method.

Myelin Component (I^m) Filter	Water Signal (I^w) Filter
$\delta = \max\left[A_m^{(1,2)}(T_2)\right]/100,\ T_{2,0} = 25\,ms$	$\delta = \max\left[A_w^{(1,2)}(T_2)\right]/100$ $T_{2,0} = 80\,ms$
1. $A_m^{(1,2)}(T_2) \geq 0, 10 \leq T_2 \leq 70\,ms$ 2. $A_m^{(1,2)}(T_2) \leq \delta,\ T_2 > 80\,ms$ 3. $\sum c_m^{(1,2)} = 0$ 4. $\int_0^\infty A_m^{(1,2)}(T_2)dT_2 = 1 \pm 0.01$ 5. $\left\lvert A_m^{(1)}(T_2) - A_m^{(2)}(T_2)\right\rvert \leq \delta/2$ $10 \leq T_2 \leq 25\,ms$	1. $A_w^{(1,2)}(T_2) \geq 0$ $T_2 \geq 10\,ms$ 2. $\left\lvert A_w^{(1)}(T_2) - A_w^{(2)}(T_2)\right\rvert \leq \frac{\delta}{2}$ 3. $\left\lvert A_w^{(1,2)}(T_2) - M\right\rvert \leq M/40$ $M = \max\left\lvert A_w^{(1,2)}(T_2)\right\rvert$

Table 1. Constraints used in calculation of coefficients.

Discussion: The proposed method may be useful for improving efficiency of data acquisition for T_2 decay analysis. The assumption made in using the method is that the image phase is described by slowly varying functions in both spatial and echo time directions. This property is well-known for well-designed CPMG sequences, and was previously explored in a number of reconstruction techniques [4,5]. The method may be extended to a larger number of *k*-space segments and may be used for other applications that require analysis of multiexponential T_2 decays [6]. The drawback of the method is SNR loss relative to images reconstructed from a full set of echoes (Table 2). Using optimized echo times [2] and reducing the first echo time may partially alleviate this problem. The method may be applied along with other data acquisition speed-up techniques such as parallel MRI and partial Fourier imaging to facilitate 3D whole brain myelin imaging.

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References: [1] MacKay AL, et. al. MRM 1994;31:673. [2] Vidarsson L, et. al. MRM 2005;53:395. [3] Jones C, et. al. MRM 2004;51:495. [4] Liang Z-P, et. al. Rev MRM 1992; 4:67. [5] Kholmovski EG, et. al. MRI 2005;23:711. [6] Gambarota G, et. al. MRM 2005;46:592.

	Full *k*-Space	Segmented *k*-space
Time Efficiency	1	2
Noise Level	1	0.54

Table 2. Comparison of standard and new techniques for MWF mapping on simulated data.

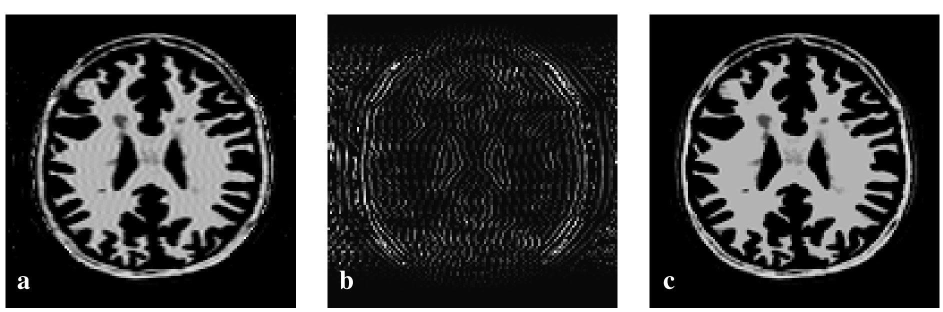

Figure 2. Short T2 component calculated from odd echoes (**a**, low frequency content), even echoes (**b**, high frequency content), and the combined full-resolution short T2 image (**c**).