K-space Based Estimation for R2* mapping

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INTRODUCTION

Quantitative T2* mapping, or R2* mapping, is used in many applications, including iron concentration estimation [1] and enhancement of BOLD contrast in fMRI [2]. In most studies, R2* maps are obtained by acquiring T2* weighted images at different TEs and fitting the parameters of the exponential signal decay. However, susceptibility-induced magnetic field gradients (SIFG) influence the signal evolution of R2* decay [3,4], leading to inaccurate R2* estimates. Previous work in this area has provided models and corrections for the effects of through-plane SIFG [5,6] but few studies have looked at the impact of in-plane SIFG [7,8]. Yablonskiy *et al.* [9] showed the importance of SIFG in the phase and frequency directions in gradient echo sequences. They proposed a signal decay equation that modeled the influence of intra-voxel magnetic gradients. R2* estimates were obtained by acquiring images at eleven TEs and fitting the signal decay with their model. The present work proposes a different approach, based on a k-space estimation algorithm and a model of in-plane SIFG. With this type of approach, flexibility in the k-space trajectory is achieved, allowing single-shot imaging, data undersampling, and shorter acquisition time. The proposed method proves to be robust to in-plane SIFG and high resolution R2* maps are also obtained.

THEORY:

In order to take into account intra-voxel magnetic gradients, the field map, ω, is assumed to be piecewise linear: $\omega(\boldsymbol{r}) \approx \sum_{n=0}^{N-1} rect(\boldsymbol{r}-\boldsymbol{r}_n)(\omega_n + \boldsymbol{G}_n.(\boldsymbol{r}-\boldsymbol{r}_n))$, with $\boldsymbol{G}_n = [X_n, Y_n]$, the vector of intra-voxel magnetic gradients in x- and y- directions, ω_n the field map value at the location $\boldsymbol{r}_n = [x_n, y_n]$. The MR signal can be written: $s(t) \approx \sum_{n=0}^{N-1} \varphi(\boldsymbol{r}_n, \boldsymbol{G}_n, t) f_n\, e^{-(i\omega_n + R_{2,n}^*)t} e^{-i2\pi(\boldsymbol{k}(t).\boldsymbol{r}_n)}$ with $\varphi(\boldsymbol{r}_n, \boldsymbol{G}_n, t) = sinc(k_x(t)\Delta x + \gamma X_n t)\, sinc(k_y(t)\Delta y + \gamma Y_n t)$, and $R_{2,n}^*$ the R_2^* value at the location r_n. R_2^* maps and the field map, ω are jointly estimated by minimizing the cost function $\psi = \frac{1}{2}\|y - A(\omega, R_2^*) f\|^2 + \beta R(f, \omega, R_2^*)$. The function R is introduced to enforce *a priori* knowledge about the smoothness of the image f, the field map, and R_2^* maps. A is defined by $a_{m,n} = \varphi(\boldsymbol{r}_n, \boldsymbol{G}_n, t_m) f_n\, e^{-(i\omega_n + R_{2,n}^*)t_m} e^{-i2\pi(\boldsymbol{k}(t_m).\boldsymbol{r}_n)}$. Nonlinear conjugate gradient combined with NUFFT are used to minimize the cost function ψ [4]. Jointly estimating the field map when estimating the R2* map is important to correct for any drift in the magnetic field during the acquisition.

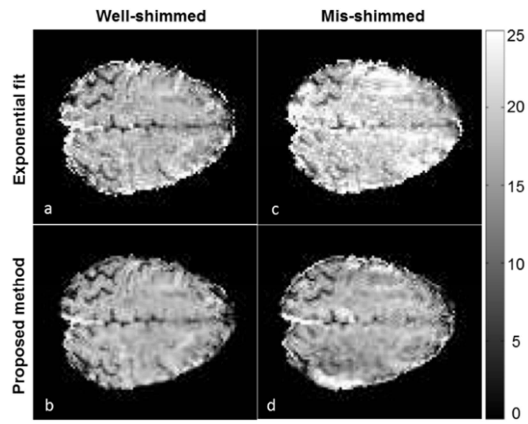

Figure 1: R2* maps in well-shimmed case (a. b.) and mis-shimmed case (c. d.) for the exponential fit (a. c.) and the proposed k-space estimation (b. d.).

METHODS

All experiment were performed on a Siemens Trio 3 T with a 32 channel head coil. Axial slices of the brain were imaged with an asymmetric spin-echo multi-echo spiral out trajectory with the following parameters: FOV = 24 cm; matrix size 128x128; 10 slices; voxel size= 1.875x1.875x1.7 mm^3; TR= 200 ms; one shot; TE = 0/40/60/80 ms. A reduction factor of 2 for the spiral trajectory was used to shorten the acquisition time. A field map was also acquired and used as an initial guess for ω in the estimation algorithm. The image obtained at TE=0 ms was used as f. In order to demonstrate the impact of in-plane SIFG, the magnetic shim was manipulated, creating a magnetic gradient of 15 Hz/cm through the brain. The proposed method was applied to estimate R2* values in the case of the additional magnetic gradient of 15 Hz/cm, referred as the mis-shimmed case, but also in the case of no additional gradient, referred as the well-shimmed case. For comparison, T2*-weighted images were also reconstructed at each TE with field inhomogeneity correction and SENSE, and an exponential fit was performed pixel by pixel on the signal decay. The R2* maps obtained with the proposed method and the exponential fit were compared in the well-shimmed and mis-shimmed cases.

To further demonstrate the estimation technique, a higher resolution R2* map was also obtained with the following parameters: Matrix size: 256x256; voxel size= 0.9x0.9x1.2 mm^3; TR= 200 ms; twelve shot; TE = 0/23/48/73 ms. A standard MPRAGE was acquired and used to define a gray/white matter mask. Average R2* values in the gray and white matter were calculated based on these masks.

RESULTS and DISCUSSION

Figure 1 shows the R2* maps in the well-shimmed and mis-shimmed case for the proposed method and the standard exponential fit. In the well-shimmed case, the R2* values for the two methods agree (Fig. 1a, 1b) since this region of the brain does not have strong SIFG. However, in the mis-shimmed case, the R2* values obtained with the exponential fit are overall higher (Fig. 1b). Even though any distortion due to in-plane SIFG in the T2* weighted images have been corrected, SIFG still leads to incorrect R2* values, due to a non-exponential R2* signal decay caused by intra-voxel gradients. With the proposed method, the R2* maps are very similar in both well-shimmed and mis-shimmed cases (Fig. 1b, 1d). Figure 2 shows a 0.9x0.9x1.2 R2* map from our joint estimation. The average values are 15.6 s^{-1} for the gray matter and 17.5 s^{-1} for the white matter, which agree with previous literature [1,10].

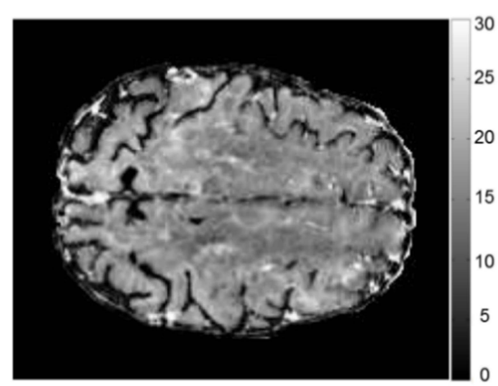

Figure 2: 0.9x0.9x1.2mm R2* map. Mean white matter R2* value is 17.5 s^{-1} and mean gray matter R2* value is 15 s^{-1}.

CONCLUSION

In-plane SIFG due to susceptibility differences in and around the brain has an important effects on R2* estimation. In this work, a fast R2* mapping method robust to SIFG is proposed and applied with different k-space trajectories. Using a k-space method can be more time efficient, since the images do not need to be reconstructed first, especially when using non-Cartesian sampling. Moreover, faster acquisition is achieved by using a single shot spiral trajectory. Future work will include through-plane magnetic susceptibility in the estimation in order to complete the SIFG model.

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