R_2^* MEASUREMENT ERRORS AT ULTRAHIGH FIELD IN THE PRESENCE OF NONLINEAR B_0 INHOMOGENEITIES

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Introduction

Clinical studies have suggested that the transverse relaxation rate R_2^* is a good measure of tissue iron content [1, 2]. However, R_2^* measurement is subject to B_0-inhomogeneity-induced error in clinical scanners. B_0 inhomogeneity along the slice selection direction causes distortion and shift along the k_z direction. For a 2D image, only the k_z = 0 line is recorded. As a result, the free induction decay (FID) signal is no longer exponential but modulated by a complex function. At low field (1.5T and below), B_0 inhomogeneity can either be ignored or approximated by a linear gradient, which corresponds to a sinc modulation [3]. However, its nonlinearity becomes proportionally more important with the main magnetic field. In this study, we assess R_2^* measurement errors caused by quadratic B_0 inhomogeneity at ultrahigh field (7T) by simulation and phantom study.

Materials and Methods

When the B_0 inhomogeneity has the form: $\delta B_0 = B_0 \times (\alpha z^2 + \beta z)$, $-z_0 \le z \le z_0$, across a slice with perfect slice profile and slice thickness $2z_0$, the FID signal can be expressed analytically:

$$FID = A\sqrt{\frac{(C_1+C_2)^2+(S_1+S_2)^2}{\gamma|\alpha|B_0T_E}}\exp\left(-\frac{T_E}{T_2^*}\right) \qquad (1)$$

Inhomogeneity	α (mm^{-2})	β (mm^{-1})	$\delta B_0 / B_0$
Weak	10^{-8}	0.2×10^{-8}	0.012 ppm / mm
Medium	10^{-7}	2.2×10^{-7}	0.32 ppm / mm
Strong	10^{-6}	2.2×10^{-6}	3.2 ppm / mm

Table 1. B_0 inhomogeneity levels used in the simulation

Where A is an amplitude factor, T_2^* is the true transverse relaxation time, γ is gyromagnetic ratio. $C_{1,2}$ are cosine Fresnel integrals of TE, α, and β. $S_{1,2}$ are corresponding sine Fresnel integrals. We used equation (1) to simulate the observed FID signals for T_2^* ranging from 2ms to 200ms, and several (α, β) combinations listed in Table 1, which correspond to different levels of B_0 inhomogeneity. Rician noise [4] was added to achieve a more realistic simulation. The simulated observed FID signals were then fitted to three commonly used decay models: simple exponential, exponential plus a constant 'noise' term, and sinc-modulated exponential. R_2^* measurement error was quantified by the relative error $\Delta R_2^* = (R_2^{*(obs)} - R_2^{*(true)}) / R_2^{*(true)}$. For each ($\alpha$, β, T_2^*) combination, 100 repeats were performed to get an estimate of ΔR_2^* scattering. Echo time T_E ranged from 1.5ms to 49.5ms with a fixed increment of 0.75ms, which is typical under EPI test mode on Philips Achieva platforms (Philips, Cleveland, OH). Other parameters used in the simulation are: B_0 = 7T, z_0 = 1mm, and A = 1. All simulations were done with Matlab (Mathworks, Natick, MA).

B_0 map and EPI test mode data were collected on a phantom built with 0.125~1mM $MnCl_2$ solutions at a Philips Achieva 7T scanner. R_2^* of $MnCl_2$ solutions follow a linear relationship with Mn^{2+} concentrations. The slope coefficients were calculated in the presence and absence of B_0 inhomogeneity, and compared by the relative differences (RD): RD = (absolute difference) / mean.

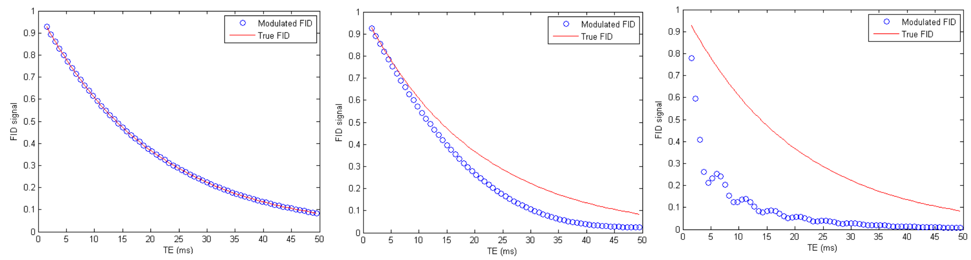

Fig. 1. Simulated FID curve (blue circle) and theoretical exponential decay curve (red line, T_2^* = 20 ms) for weak (left), medium (middle), and strong (right) B_0 inhomogeneities.

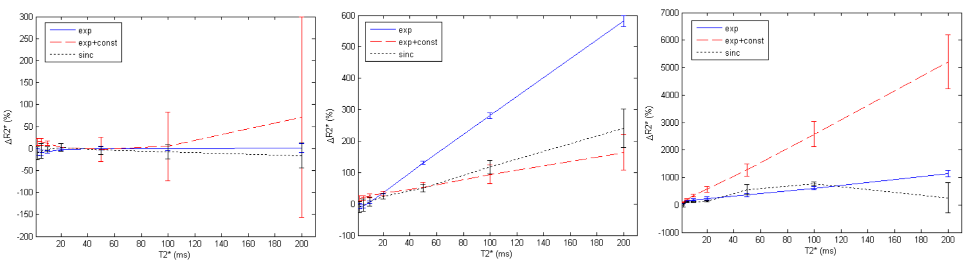

Fig. 2. R_2^* measurement errors by the exponential (blue solid line), exponential-plus-constant (red dashed line) and sinc-modulated exponential (black dotted line) models for weak (left), medium (middle), and strong (right) B_0 inhomogeneities. A Rician noise whose single-channel standard deviation is 5% of the first-echo signal intensity is added. Error bars are estimated from 100 repeats.

Results

Modulated FID signals are plotted versus T_E with the true exponential FID curve in Fig. 1. B_0 inhomogeneity effect can only be ignored when the inhomogeneity field is weak. R_2^* measurement errors are plotted versus the true T_2^* values in Fig. 2. R_2^* measurement error increases with B_0 inhomogeneity for all three methods, from ~10% at weak inhomogeneity to ~1000% at strong inhomogeneity. Phantom study results are listed in Table 2. RD of the quadratic correction method based on equation (1) and B_0 map is only 1/4 to 1/5 of those of the three commonly used models.

Discussion and Conclusion

Fig. 2 shows that R_2^* estimates from all three commonly used models are severely compromised when $\delta B_0 / B_0$ > 0.1ppm / mm. In human head imaging, air-tissue susceptibility differences at sinuses can generate a B_0 inhomogeneity of this scale several centimeters away from the air-tissue interface. Thus, R_2^* values measured with either of these three models are not reliable indeces of the actual tissue relaxometric property. This can be overcome, as demonstrated by the phantom data, by measuring B_0 map and conducting a quadratic correction.

Model	RD
Exp	43.2%
Exp + const	39.2%
Sinc	47.2%
Quadratic correction	9.6%

Table 2. RD of R_2^*-[Mn^{2+}] slopes in the presence and absence of B_0 inhomogeneity for the three commonly used models and a quadratic correction (based on B_0 map).

References

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[3] Fernandez-Seara et al, *MRM* 2000, **44**:358-366; [4] Gudbjartsson el al, *MRM* 1995, **34**: 910-914