Advanced 3D tailored RF pulse for signal loss recovery in T2*-weighted functional MRI

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Introduction

T_2^* contrast, used in blood oxygenation level dependent (BOLD) fMRI, is coupled to the signal loss artifact caused by through-plane dephasing between excitation and data acquisition. Various methods have been shown to be effective at recovering signal loss, but they typically involve high cost in different aspects of the fMRI study. In particular, using three-dimensional tailored RF pulse to "precompensate" for the dephasing (hereafter, 3DTRF method) [1] was demonstrated to be very effective, but the pulses are exceedingly long, and their slice selectivity is limited and very susceptible to off-resonance distortion. We propose major advances in the 3DTRF method that lead to multiple-fold pulse length reduction and improved slice selection, while effective signal recovery is achieved. We demonstrate signal recovery on an inferior brain slice, using a 3DTRF pulse that is only 15.4 ms long.

Theory

The desired excitation pattern for phase precompensation comprises a slice profile and negative of the (negatively-signed) differential through-plane phase variations accumulated up to T_E:

$$d_i = d(x_i, y_i, z_i; z_0) = p(z_i - z_0) \times \exp(i\omega_{diff}(z_i; x_i, y_i, z_0) T_E), \qquad (1)$$

$i = 1,\ldots, N_s$, where γ is the gyromagnetic ratio, $p(z_i-z_0)$ is the slice profile centered at z_0, and $\omega_{diff}(z_i; x_i,y_i,z_0)$ is the differential frequency offsets through the slice at location (x_i,y_i). ω_{diff} can be estimated based on field maps from a scan prescribed on subslices that constitute the profile. If such desired pattern was excited, then spins would be in phase through the slice at T_E, and thus signal loss would be recovered significantly. Using the excitation k-space perspective [1], RF pulse samples ($\hat{\mathbf{b}}$) for selective excitation of (1) can be designed iteratively [2]:

$$\hat{\mathbf{b}} = \arg\min_{\mathbf{b}} \left\{ \|\mathbf{Ab} - \mathbf{d}\|_{\mathbf{W}}^2 + \mathrm{R}(\mathbf{b}) \right\} \qquad (2)$$

where $\mathbf{d}=[d_1,\ldots,d_{Ns}]^T$, matrix $\mathbf{A}$ is a function of the excitation k-space trajectory and frequency offsets [2], diagonal $\mathbf{W}$ specifies the region of interest, and $\mathrm{R}(\mathbf{b})$ denotes regularization terms for controlling pulse power. $\mathbf{Ab}$ approximates the 3D pattern excited by pulse samples $\mathbf{b}$, and with the first term in (2) we attempt to minimize its difference from $\mathbf{d}$ in a weighted-least-squares sense. Incorporation of measured frequency offsets in $\mathbf{A}$ ensures that $\mathbf{Ab}$ well approximates the actual excitation pattern in the scanner. The minimization problem is solved iteratively via conjugate gradient [2], with acceleration using NUFFT and time segmentation technique [4].

The adaptive 3D trajectory in [5] can be used to "frequency encode" in the slice selection direction and "phase encode" in-plane. Compared to the stack-of-spirals trajectory [1], this trajectory efficiently provides rapid through-plane phase variations, and superior slice selection with high immunity to off resonance and no excitation sidelobe problems [1]. In-plane selectivity is good enough for the 3DTRF method. The improved sampling efficiency in k-space leads to significant pulse length reduction for a given recovery efficacy.

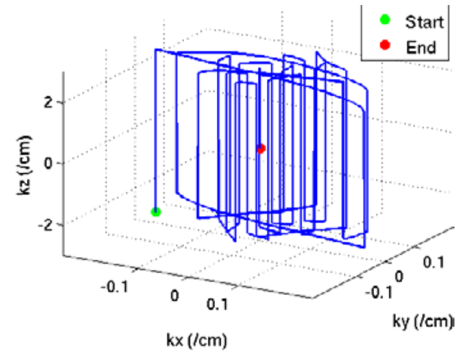

Fig. 1: Excitation k-space trajectory which frequency encodes in the slice-selection direction, and phase encodes in-plane.

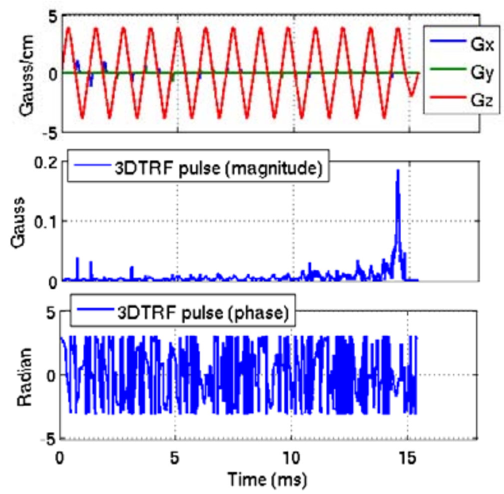

Fig. 2: 3DTRF pulse accompanied by the gradient waveforms that underlie the trajectory in Fig. 1.

Methods and results

With a GRE spiral-out sequence (T_E =30 ms, 2 interleaves, resolution = 1.875 mm x 1.875 mm) using a standard sinc pulse selective on a 5 mm slice, signal loss was observed in an inferior brain slice (Fig. 3, left). To test signal recovery with the 3DTRF method, we first scanned the subject for field maps on 1 mm subslices. Through-plane differential frequency offsets were derived from the field maps, from which we formed the desired pattern (1) with Gaussian profile (FWHM = 5 mm). A 3DTRF pulse (Fig. 2) was then computed to replace the sinc pulse in the original GRE sequence, with T_E measured from pulse end to acquisition onset. The acquired image showed significant recovery of signal originally lost using the sinc pulse (Fig. 3, right). Signals from outside the loss region remained minimally unaffected. The 3DTRF pulse length (15.4 ms) was several times shorter than needed in the original design [1].

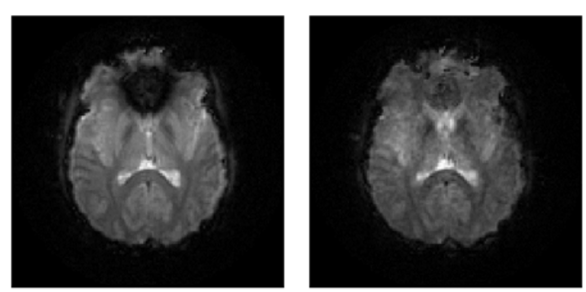

Fig. 3: Left: signal loss with standard sinc pulse. Right: signal recovery with the 3DTRF pulse in Fig. 2. (Flip angles were not matched in the two images.)

Discussion

The major advances we have introduced to the original 3DTRF method leads to 3DTRF pulses that are short, effective in signal loss recovery, and accurate in slice selection. With the 3DTRF method, it is possible to image loss-plagued brain regions with reduced cost in fMRI temporal resolution compared to methods such as z-shimming [6]. In addition, the method does not compromise image SNR as with slice thickness reduction, nor does it requires extra hardware as in the localized shimming methods (e.g. [7]). The 3DTRF method appears promising for routine fMRI studies, either alone or in conjunction with the other recovery methods.

References

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