A Fast-k_z 3D Tailored RF Pulse for Reduced B1 Inhomogeneity

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Introduction: The presence of B1 inhomogeneity is a limitation at high field. One means of mitigating B1 inhomogeneity is with multi-dimensional tailored RF (TRF) pulses designed using the small tip angle approximation (1). Current multi-dimensional TRF methods, however, have poor slice-selection properties or require long pulse lengths (2,3). We present a new 3D TRF design using continuous slice-select gradients along k_z and phase encoding along k_x-k_y (termed "fast-k_z") to correct for a quadratically varying inhomogeneity in the *x*-*y* direction in a 5 mm thick slice. Examples of T2-weighted images of the human brain at 3T are presented.

Theory: The small tip angle equations can be modified to include an inhomogeneous transmit sensitivity $\alpha(\mathbf{r})$:

$$M(\mathbf{r}) = \alpha(\mathbf{r})\int_0^T B_1(t)e^{-i\mathbf{k}(t)\cdot\mathbf{r}}dt \quad \text{where} \quad B_1(t) = W(\mathbf{k}(t))\,|\gamma\mathbf{G}(t)|\,. \qquad [1]$$

The weighting $W(\mathbf{k}(t))$ is the Fourier transform of $M(\mathbf{r})/\alpha(\mathbf{r})$ sampled along the *k*-space produced by $\mathbf{G}(t)$. The inhomogeneity $\alpha(\mathbf{r})$ can be obtained by direct measurement or by approximation. Because the inhomogeneity in a slice at 3T is typically characterized by brightness in the image center, $\alpha(\mathbf{r})$ can be approximated as a quadratic function. This can be accomplished by modeling the k_x-k_y spatial frequency weighting by five delta functions:

$$W(k_x,k_y) = A_0\delta(k_x,k_y) + A_1e^{i\phi_0}\delta(k_x-k_0,k_y) + A_1e^{-i\phi_0}\delta(k_x+k_0,k_y) + A_1e^{i\phi_0}\delta(k_x,k_y-k_0) + A_1e^{-i\phi_0}\delta(k_x,k_y-k_0)\,. \qquad [2]$$

This will produce a spatial weighting function

$$w(x,y) = A_0 + 2A_1\cos(2\pi k_0(x+x_0)) + 2A_1\cos(2\pi k_0(y+y_0))\,. \qquad [3]$$

Here $x_0=y_0=\phi_0/2\pi k_0$. If k_0 is small (< 1/(2FOV)), the cosine functions can be approximated to give

$$w(x,y) = (A_0+4A_1) - A_1(2\pi k_0)^2[(x+x_0)^2+(y+y_0)^2]\,. \qquad [4]$$

Here $A_1(2\pi k_0)^2$ is the curvature of the profile and (x_0,y_0) is the center of the RF bulge. At the edge of the FOV, the RF intensity will be different by a fractional amount ε:

$$\varepsilon = -A_1(\pi k_0 FOV)^2/(A_0+4A_1)\,. \qquad [5]$$

One makes A_1 negative to reduce excitation in the center. A pulse producing this weighting can be obtained using a design consisting of five slice-select pulses that switch along k_z for phase-encodes in k_x-k_y. Figure 1 (a) shows the *k*-space and (b) and an example pulse. The k_z direction (slice-profile) was weighted by a Gaussian. Figure 1 (c) shows a Bloch equation simulation of the magnetization.

Methods: The pulses were inserted into a spiral sequence and normal volunteers were scanned on a GE 3T system running under the VH4 software version. The slice thickness was set to 5mm and the flip angle to 45^0. The central sub-pulse in the 3D TRF pulse was widened and variable rate excitation (VERSE) (4) was used to reduce the peak B1. The spiral acquisition parameters were 24 interleaves, 24 slices, 256x256 matrix size, 24cm FOV, TE=5ms, and TR=1.5sec. Three 3D TRF pulses were created in Matlab for ε equal to 1.0, 1.5, and 2.0. Slices having the best image uniformity were chosen by inspection. A standard birdcage head coil was used.

Results: Figure 1 (d) and (e) show examples images excited with a standard 5 mm sinc pulse and the corresponding fast-k_z pulse, respectively with ε = 1.5. The image excited with the fast-k_z pulse shows more uniformity.

(a) **(b)** **(c)** **(d)** **(e)**

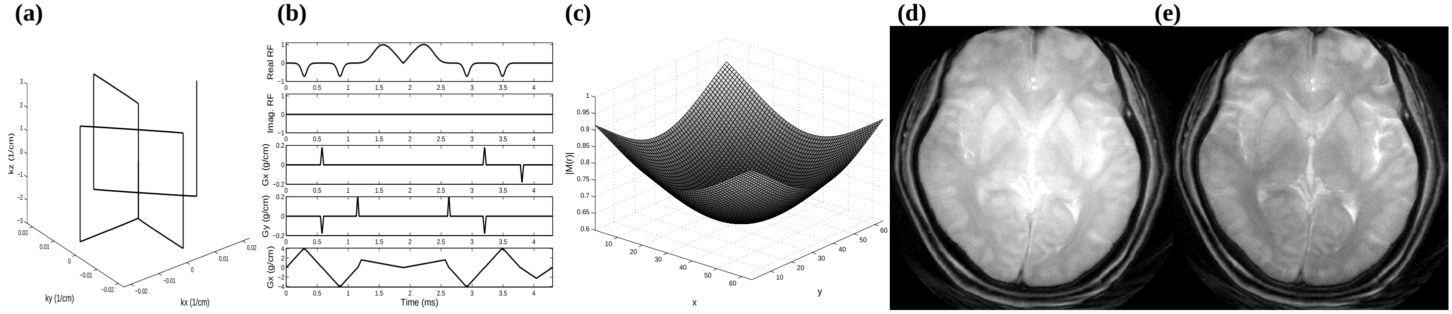

Figure 1. (a) Fast-k_z trajectory and **(b)** sample pulse. The rows (top to bottom) are the real and imaginary RF, and the *x*-, *y*-, and *z*-gradients. **(c)** Simulation of the spatial weighting. **(d)** 5 mm thick image acquired with a 45^0 sinc pulse at 3T. **(e)** Same slice acquired with 3D TRF pulse.

Conclusions: A fast-k_z 3D TRF pulse used for the reduction of B1 inhomogeneity at 3T is presented. The pulse is capable of thin slices (2-5mm) and has a length of 3-5ms. We found that the image uniformity was improved in all slices. Future work will address calibrating the fractional correction for different slice locations as well as flip angle.

References: (1) Pauly JM, Nishimura D, Macovski A. J. Magn. Reson. 1989;81:43-56. (2) Deichmann R, Good C, Turner R. Magn Reson Med 2002;47:398-402. (3) Saekho S, Boada FE, Noll DC, Stenger VA. Magn Reson Med 2004; *to be published*. (4) Conolly S, Nishimura D, Macovski A. J Magn Reson 1988;78:440. Work supported by the NIMH 1 R01 MH66066-01 and NIDA 1 R21 DA015900-01.