In Vivo $B_1^+$ Inhomogeneity Mitigation at 7 Tesla using Sparsity-Enforced Spatially-Tailored Slice-Selective Excitation Pulses

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INTRODUCTION. We design & demonstrate a 7-ms slice-selective pulse that mitigates $B_1^+$ inhomogeneity in the human brain at 7T without the use of a parallel transmission system. At high field, severe RF inhomogeneity due to wavelength interference & attenuation causes standard slice-selective pulses (SSSPs) to produce non-uniform flip angles across the field of excitation (FOX), leading to contrast & SNR non-uniformity. One way to mitigate $B_1^+$ inhomogeneity is to use spoke-based RF pulses; these are comprised of weighted sinc-like segments in $k$ placed at different locations in $(k_x, k_z)$ that play along an echo-volumetric trajectory [1,2]. In the small-tip-angle regime [3], the sinc segments excite a slice in $z$, while the $(k_x, k_z)$ weights tailor the in-plane excitation into the pointwise-inverse of the inhomogeneity. The work here extends our earlier effort [4] to in vivo trials & makes use of recent techniques: a magnetization reset pulse to permit fast (TR<51s) acquisition of multiple images [5], the fitting of these images to an intensity equation to estimate $B_1^+$, & a novel sparsity-enforced spoke placement to find a small set of spoke locations & weights [6].

THEORY & METHODS. Signal intensity equations. Image intensity $I_V$ at location $r$ due to an SSSP with peak voltage $V$ is:

$$I_V(r) = \frac{\gamma v B_{1+}(r) \sin(\alpha_o(r)) [1-E_r(r,TR)][1-E_r(r,TR) \cos(\alpha_o(r))]^2}{(1-E_r(r,TR) \cos(\alpha_o(r)))^2} \quad (Eq.1),$$

where $c$ is a constant, $\rho$ proton density, $B_1$ the receive profile, $E_r(r,TR) = \exp(-TR/T_1(r))$, and $\alpha_o(r) = \gamma v r B_1(r)$, where $T_1$ is the SSSP’s duration & $B_1$ is in Tesla/volt. Let $R(r) = \rho(r) B_1(r)$. With a reset pulse [5],

$$I(r) = c R(r) [1-E_r(r,TR) \cos(\alpha_o(r))] \quad (Eq.2),$$

i.e., the $T_1$-denominator is removed (even if TR<T_1). Finally, if $\alpha_o$ is small and a reset pulse is not used, $\cos(\alpha_o) \approx 1$, $\sin(\alpha_o) \approx \alpha_o$, and thus $I(r) = c R(r) \cos(\alpha_o) \approx \alpha_o$.

Profile estimation. To estimate $B_1^+(r)$, we collect $N$ images with increasing $V$ using an SSSP & reset pulse [5]. Then $\forall r \in$ FOV, we fit the $N$ values to Eq.2. To estimate $R(r)$, we collect a low-flip-angle image, $L_m(r)$, without a reset pulse. Eq.3 now holds, and $L_m(r) / B_1^+(r)$ yields $R(r)$ within a constant.

Sparsity-Enforced Spoke Placement (SESP) & pulse design. To minimize pulse duration, only a few spokes may be used; each must be placed & weighted such that the excitation resembles $B_1^+(r)$', so that the overall magnetization $m(r)$ is uniform. One may use SESP [4,6] to determine good spoke coordinates: First, discretize space at locations $r_i$, $i = 1...N$. Next, define a set of candidate spokes in 2-D k-space, $k_j$, $j = 1...N_k$, with weights $g_j$. Let $m \in \mathbb{C}^N$ be a vector of $m(r)$ samples, $g \in \mathbb{C}^{N_x}$ a vector of $g_j$, $D$ a diag. matrix of $B_1^+(r_j)$ samples, and $A \in \mathbb{C}^{N_x,N}$ where $A_{m,g} = \exp(j \pi r_m k_j)$; then, $m = D A g$. Next, define a target magnetization, $d(r)$, sample it, and form $d \in \mathbb{C}^N$. Finally, solve $\min_{g} \|d - D A g\|_2^2 + \lambda \|A g\|_1$ (for fixed $\lambda$); this yields a sparse $g$, one with few large weights, revealing a small set of $T_1$ locations to be traversed by the gradients.

The pulse is designed by fixing spoke shape in $k_x$, truncating all but $T$ of A’s columns, & retuning the weights by least-squares fitting $d = D A_{m,g} \|_{m,g}$. Post-mitigation flip angle estimation & quality metrics. $B_1^+$ mitigation is quantified by playing the pulse and analyzing the resulting flip angle map, $\alpha_m(r)$. This is achieved by obtaining a low-flip mitigation image, $L_m(r) \propto R(r) \alpha_m(r)$ (per Eq.3). Since $R(r)$ is known, $L_m(r) / R(r)$ gives $\alpha_m(r)$ within a multiplicative constant. The uniformity of $\alpha_m(r)$ is quantified by computing its in-FOX normalized standard deviation, $\sigma$, and worst-case maximum variation, MV (maximum in-FOX value divided by minimum in-FOX value); these values are then compared to those of the initial $\alpha_o(r)$.

RESULTS. Human studies used a 7T scanner, body gradients, and a quadrature birdcage coil in accordance with the institution’s HRC. Ten images were collected using SSSPs ($V = 20V, 60V, ... 380V; TR = 1s$) followed by resets. Data was fitted to obtain $\alpha_o(r)$ and $B_1^+(r)$ (Fig. 1: C); each is highly non-uniform with $(\sigma, MV) = (0.15, 2.24)$. An $R(r)$ estimate was obtained from a low-flip SSSP image without reset pulse (Fig. 1: A: B). $B_1^+$ was fed to SESP, and with $\lambda = 0.35$, 19 spoke locations were determined (Fig. 2). After fixing spokes to be Hanning-windowed slices (TBW=4), these locations & weights yielded the 7-ms pulse shown (Fig. 3). This pulse was simulated (Fig. 1: D) to verify that it yielded approximately $B_1^+(r)$'s.

The pulse was applied in vivo, and a low-flip image obtained (Fig. 1: E); slice selection worked properly (Fig. 1: F). This image was divided by $R(r)$ to yield $\alpha_m(r)$ (Fig. 1: G). Qualitatively, $\alpha_m(r)$ is significantly more uniform than $\alpha_o(r)$ (compare the 1-D profiles). Quantitatively, $\sigma$ and worst-case MV have been reduced by factors of 3 and 1.7, respectively, a major flip angle uniformity improvement relative to $\alpha_o(r)$.

CONCLUSION. In vivo $B_1^+$ inhomogeneity present in the human brain at 7T was mitigated using a 7-ms slice-selective SESP-designed pulse. Commercially-available head-only gradients with amplitude & slew rates of 35 mT/m and 600 T/m/s would allow the use of a 19-spoke, 10-mm excitation pulse that performs $B_1^+$ mitigation in only 5.25 ms.
