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

A. C. ZELINSKI[1], K. SETSOMPOP[1], V. ALAGAPPAN[2], V. K. GOYAL[1], L. L. WALD[3,4], AND E. ADALSTEINSSON[1,4]

[1]RESEARCH LABORATORY OF ELECTRONICS, MASSACHUSETTS INSTITUTE OF TECHNOLOGY, CAMBRIDGE, MA, UNITED STATES, [2]A. A. MARTINOS CENTER FOR BIOMEDICAL IMAGING, CHARLESTOWN, MA, UNITED STATES, [3]DEPARTMENT OF RADIOLOGY, HARVARD MEDICAL SCHOOL, LONGWOOD, MA, UNITED STATES, [4]HARVARD-MIT DIVISION OF HEALTH SCIENCES & TECHNOLOGY, CAMBRIDGE, MA, UNITED STATES

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_z placed at different locations in (k_x, k_y) that play along an echo-volumnar trajectory [1,2]. In the small-tip-angle regime [3], the sinc segments excite a slice in z, while the (k_x, k_y) 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 ($TR \ll T_1$) 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 $\mathbf{r}$ due to an SSSP with peak voltage V is:

$$I_V(\mathbf{r}) = c\cdot\rho(\mathbf{r})\cdot B_1^-(\mathbf{r})\cdot\sin(\alpha_o(\mathbf{r}))[1-E_1(\mathbf{r},TR)][1-E_1(\mathbf{r},TR)\cos(\alpha_o(\mathbf{r}))]^{-1} \quad \text{(Eq.1)},$$

where c is a constant, ρ proton density, B_1^- the receive profile, $E_1(\mathbf{r},TR) = \exp(-TR/T_1(\mathbf{r}))$, and $\alpha_o(\mathbf{r}) = \gamma V\tau\cdot B_1^+(\mathbf{r})$, where τ is the SSSP's duration & B_1^+ is in Tesla/volt. Let $R(\mathbf{r}) \equiv \rho(\mathbf{r})\cdot B_1^-(\mathbf{r})$. With a reset pulse [5], $I_V(\mathbf{r}) = c\cdot R(\mathbf{r})\cdot[1-E_1(\mathbf{r},TR)]\cdot\sin(\alpha_o(\mathbf{r}))$ (Eq.2), i.e., the T_1-denominator is removed (even if $TR \ll T_1$). Finally, if α_o is small and a reset pulse is **not** used, $\cos(\alpha_o) \simeq 1$, $\sin(\alpha_o) \simeq \alpha_o$, and thus $I_v(\mathbf{r}) = c\cdot R(\mathbf{r})\cdot\alpha_o(\mathbf{r})$ (Eq.3).

Profile estimation. To estimate $B_1^+(\mathbf{r})$, we collect N images with increasing V using an SSSP + reset pulse [5]. Then $\forall \mathbf{r} \in$ FOX, we fit the N values to Eq.2. To estimate $R(\mathbf{r})$, we collect a low-flip-angle image, $L_o(\mathbf{r})$, *without* a reset pulse. Eq.3 now holds, and $L_o(\mathbf{r}) / B_1^+(\mathbf{r})$ yields $R(\mathbf{r})$ 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 $[B_1^+(\mathbf{r})]^{-1}$, so that the overall magnetization $m(\mathbf{r})$ is uniform. One may use SESP [4,6] to determine good spoke coordinates: First, discretize space at locations $\mathbf{r}_i$, $i = 1\ldots N_s$. Next, define a set of candidate spoke locations in 2-D k-space, $\mathbf{k}_i$, $i = 1\ldots N_f$, with weights g_i. Let $\mathbf{m} \in \mathcal{C}^{N_s}$ be a vector of $m(\mathbf{r}_i)$ samples, $\mathbf{g} \in \mathcal{C}^{N_f}$ a vector of g_is, $\mathbf{D}$ a diag. matrix of $B_1^+(\mathbf{r}_i)$ samples, and $\mathbf{A} \in \mathcal{C}^{N_s\times N_f}$, where $\mathbf{A}_{m,n} \propto \exp(j2\pi\mathbf{r}_m\cdot\mathbf{k}_n)$; then, $\mathbf{m} = \mathbf{DAg}$. Next, define a target magnetization, $d(\mathbf{r})$, sample it, and form $\mathbf{d} \in \mathcal{C}^{N_s}$. Finally, solve $\min_{\mathbf{g}} \|\mathbf{d}-\mathbf{DAg}\|_2^2 + \lambda\|\mathbf{g}\|_1$ (for fixed λ): this yields a *sparse* $\mathbf{g}$, one with few large weights, revealing a small set of T locations to be traversed by the gradients. The pulse is designed by fixing spoke shape in k_z, truncating all but T of $\mathbf{A}$'s columns, & retuning the weights by least-squares fitting $\mathbf{d}=\mathbf{DA}_{\text{trunc}}\mathbf{g}_{\text{trunc}}$.

Post-mitigation flip angle estimation & quality metrics. B_1^+ mitigation is quantified by playing the pulse and analyzing the resulting flip angle map, $\alpha_m(\mathbf{r})$. This is achieved by obtaining a low-flip mitigation image, $L_m(\mathbf{r}) \propto R(\mathbf{r})\cdot\alpha_m(\mathbf{r})$ (per Eq.3). Since $R(\mathbf{r})$ is known, $L_m(\mathbf{r})/R(\mathbf{r})$ gives $\alpha_m(\mathbf{r})$ within a multiplicative constant. The uniformity of $\alpha_m(\mathbf{r})$ is quantified by computing its in-FOX normalized standard deviation, σ, 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 $\alpha_o(\mathbf{r})$.

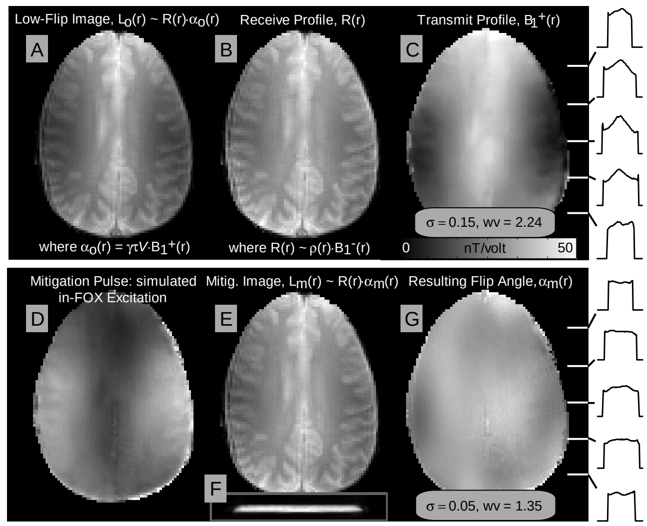
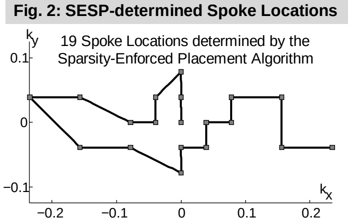


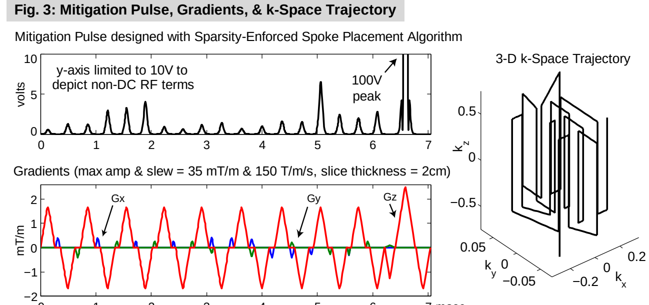
Fig. 1: B_1^+ Mitigation Results in the Human Brain at 7T
A. Low-flip image, $L_o(r)$, collected using standard slice-selective pulse
B. Receive profile, R(r), contains proton-density weighting
C. ***Highly nonuniform*** transmit profile, $B_1^+(r)$, **σ = 0.15, worst-case variation = 2.24**
D. B_1^+ Mitigation pulse: simulated in-FOX excitation, strongly resembles $[B_1^+(r)]^{-1}$
E. B_1^+ Mitigated image (in-plane), closely resembles R(r), implying successful mitigation
F. B_1^+ Mitigation image (through-plane), slice selection is evident
G. ***Highly uniform*** flip angle after mitigation, **σ = 0.05, worst-case variation = 1.35**

Fig. 2: SESP-determined Spoke Locations



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 $\alpha_o(\mathbf{r})$ and $B_1^+(\mathbf{r})$ (Fig. 1: C); each is highly non-uniform with (σ, MV) = (0.15, 2.24). An $R(\mathbf{r})$ estimate was obtained from a low-flip SSSP image *without* reset pulse (Fig. 1: A, B). $B_1^+(\mathbf{r})$ was fed to SESP, and with λ = 0.35, 19 spoke locations were determined (Fig. 2). After fixing spokes to be Hanning-windowed sincs (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 $[B_1^+(\mathbf{r})]^{-1}$. 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 $R(\mathbf{r})$ to yield $\alpha_m(\mathbf{r})$ (Fig. 1: G). Qualitatively, $\alpha_m(\mathbf{r})$ is significantly more uniform than $\alpha_o(\mathbf{r})$ (compare the 1-D profiles). Quantitatively, σ and worst-case MV have been reduced by factors of 3 and 1.7, respectively, a major flip angle uniformity improvement relative to $\alpha_o(\mathbf{r})$.

Fig. 3: Mitigation Pulse, Gradients, & k-Space Trajectory



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.

REFERENCES. [1] Saekho et al. MRM '06;55(4):719-724. [2] Ulloa et al. 2005; ISMRM, p 21. [3] Pauly et al. JMR '89;81:43-56. [4] Setsompop, Zelinski et al. ISMRM '07, p 356. [5] Cunningham et al. MRM '06;55:1326-1333. [6] Zelinski et al. ISMRM '07, p 1691.