## Through-plane Signal loss recovery and B1 inhomogeneity reduction in vivo at 7T using parallel transmission

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**Introduction**: Two main obstacles still exist in $T_2$*-weighted functional MRI at ultra high field (UHF): through-plane signal loss and $B_1$ inhomogeneity. The through-plane signal loss artifact arises from the field inhomogeneity caused by magnetic susceptibility differences between air cavities and tissue at near functionally relevant brain regions such as the orbital-frontal and inferior temporal cortices. One approach to mitigate the signal loss is slice selection with through-plane phase precompensation using multidimensional tailored RF pulses excitation [1], which has been well demonstrated at lower fields. Similarly, parallel transmission (PTX) has been demonstrated as a promising means to reduce $B_1$ inhomogeneity at 3T [2]. In this study, we combined the two aforementioned approaches to simultaneously recover signal loss and reduce $B_1$ inhomogeneity in $T_2$*-weighted fMRI at 7T.

**Method:** Three dimensional tailored RF pulse (3DTRF) pulse desgin [3] can reduce signal loss by precompensating the through-plane phase variation in the excitation stage. We use the derivations in Ref. [1] to calculate the precompensated phase. Assuming an imaging at slice location $z_0$ and with $T_E$ the center of the k-space acquisition, the through-plane differential phase variation that is responsible for signal loss at in-plane location (x,y) can be described as the follows,

$$\phi(x,y,z;z_0) = -2\pi \cdot \mathrm{T_E} \cdot [\Delta f(x,y,z) - \overline{\Delta f}(\mathrm{x,y;z_0})] \qquad (1)$$

$$\mathrm{D_{comp}}(\mathbf{r},z_0) = \mathrm{D_{orig}}(\mathbf{r},z_0) \cdot \mathrm{p}(z-z_0) \cdot \exp(-i \cdot \phi(\mathbf{r},z_0)) \quad (2)$$

where $\overline{\Delta f}$ is the mean frequency offset of field maps for multiple slices; p(z) is the slice profile; $D_{orig}$ and $D_{comp}$ are the original and precompensated desired patterns at each slice, respectively. The dephasing variations are calculated via field maps and precompensated into desired patterns for RF pulse design.

Using the small tip angle excitation, we extend the method proposed by Grissom et al [4] to design RF pulses for parallel transmission. To control the excitation at a set of N different slices, we extend the set of equations in Ref [4] and concatenate the desired patterns $D_{comp}\{D_{Slice1}, D_{Slice2},\ldots, D_{SliceN}\}$ and the encoding matrix A $\{A_{Slice1}, A_{Slice2},\ldots, A_{SliceN}\}$ where N different slices of spatial sensitivity maps ($B_1$+) and field maps ($B_0$) are encoded, to form the following concatenated equation that can be solved via Conjugate Gradient (CG) optimization.

$$\begin{bmatrix} [\mathrm{D_{Slice1}}] \\ [\mathrm{D_{Slice2}}] \\ \vdots \\ [\mathrm{D_{SliceN}}] \end{bmatrix} = \begin{bmatrix} [\mathbf{A}_{\mathrm{Slice1}}] \\ [\mathbf{A}_{\mathrm{Slice2}}] \\ \vdots \\ [\mathbf{A}_{\mathrm{SliceN}}] \end{bmatrix} \times \mathbf{b} \qquad (3)$$

All human brain studies were performed on a 7T Siemens (Erlangen, Germany) whole body scanner equipped with a PTX extension. An 8-channel Tx/Rx coil was used to drive independent RF waveforms through each Tx channel. The $B_1$ map (64×64) for each transmit channel was obtained sequentially with a 2D GRE sequence by varying the excitation voltages from 12.5V to 150V in a total 12 steps. The RF design was implemented in Matlab R2010a (Mathworks, Natick, MA, USA) and targeted a slice thickness of 5mm. Field maps for multiple slices were incorporated into encoding the matrix *A* for off-resonance correction. A fly-back fashioned rungs trajectory (Fig. 1) was used instead as a means to overcome the RF artifacts typical of the conventional rungs [5]. The maximum gradient amplitude and slew rate were 36 mT/m and 150 mT/m/ms, respectively. To investigate the performance of the proposed method, standard and compensated RF pulses were designed with TE=8ms, 15ms and 24ms, respectively (TR=250ms, 7 Slices, FA=10°, FOV=220mm×220mm).

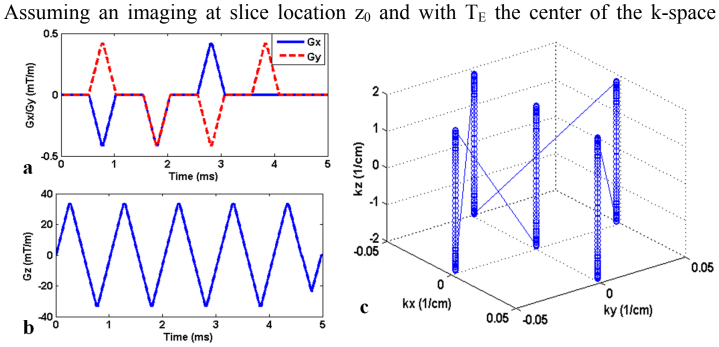


**Fig.1** (a) Gx, Gy, (b) Gz and (c) k-space trajectory.

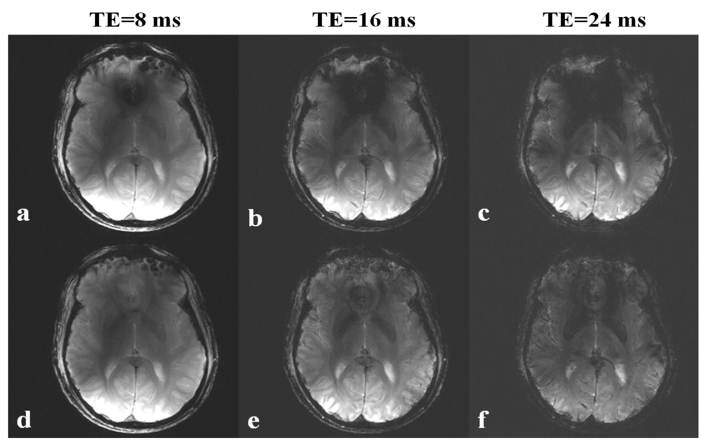


**Fig.2** Human brain images acquired with standard (top, a-c) and compensated 3D PTX (bottom, d-f) RF pulses.

**Results and Discussion:** Figure 2 shows examples of acquired human brain images with standard and compensated 3D PTX RF pulses for different TEs. Signal loss was significantly mitigated with the use of compensated 3D PTX RF pulses and, in addition, there is also a reduction of $B_1$ inhomogeneity. A more detailed depiction of the signal recovery obtained is presented in Fig.3 where line profiles through the orbital-frontal cortex are presented. Note that as TE is increased, the intravoxel dephasing is more severe and the effectiveness of the proposed correction is decreased. This is a direct reflection of the higher spatial frequency content in the through-plane phase, which cannot be effectively compensated for using the linear model assumed above. One potential solution for this problem is to employ more rungs in the RF pulse design, albeit at the expense of extending the pulse duration, which makes the excitation profile more susceptible to RF and $T_2$* relaxation effects. We also found that the 3D PTX RF pulse performance can be significantly improved by incorporating more slices into the RF design. This approach, however, requires faster computational tools but it could be made practical using Graphics Processing Units (GPU).

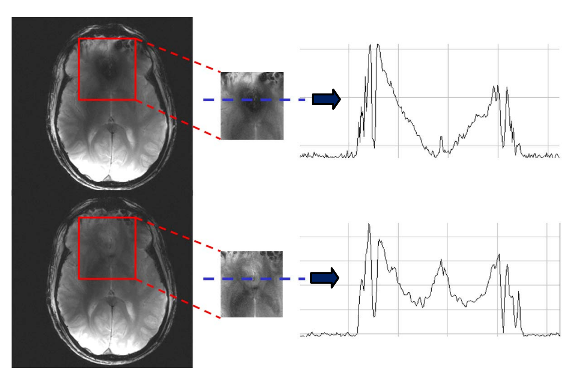

**Fig.3** Expanded view of the signal recovery for TE=8ms.

**Conclusions and Future works:** We have demonstrated an effective method for designing parallel transmission RF pulses to simultaneously recover signal loss and mitigate $B_1$ inhomogeneity using short RF pulse durations during $T_2$*-weighted functional MRI at 7T. The proposed approach can be improved further using hybrid rungs trajectories to successfully incorporate higher orders of spatial frequencies into the through-plane phase compensation profile.

**Reference:** [1] Yip et al., Magn Reson Med 2006;56:1050-1059. [2] Setsompop et al, Magn Reson Med 2006;56:1163-1171. [3] Stenger et al., Magn Reson Med 2000;44:525-531. [4] Grissom et al.,Magn Reson Med 56:620-629.2006. [5] Jankiewicz et al., JMR 2010; 203:294-304. **Supported in part by PHS Grant R01-MH088370-01**