Spatially Resolved Extended Phase Graphs: modeling and design of multi-pulse sequences with Parallel Transmission

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Introduction RF field (B_1) variation at high field strengths (3T+) leads to spatially variable signal and tissue contrast, reducing both clinical and scientific utility of resulting images. Parallel transmission (PTx) enhances control of the spatial properties of B_1 by allowing independent modulation of the relative weighting of multiple channels each of which drives a coil with its own spatial transmit sensitivity $\sigma_i(\mathbf{r})$. The resulting field $B_1(\mathbf{r})$ is the weighted sum over channels $B_1(\mathbf{r})=\alpha\Sigma w_i\sigma_i(\mathbf{r})$, where w_i are complex weights (otherwise known as RF shims) and α is an overall drive scale. RF shimming seeks to make the overall field as uniform as possible for each subject [1]. This helps stabilise performance since more uniform fields produce more uniform tissue signals and contrast. However most imaging sequences employ rapid successions of RF pulses, making the received signals depend on the combined history of many pulses interspersed with periods of relaxation. Dynamic modulation of the RF field throughout the sequence would allow interaction with this process, yielding more degrees of freedom with which to achieve desired signal properties. The MR signal and contrast, rather than the fields, are ultimately of interest and hence direct optimization of these would make sense. In order to make this possible, a model linking the spatially varying fields to the resulting signals is required: this is provided by the Spatially Resolved (SR) Extended Phase Graph (EPG) framework [2,3]. In this work we explore the shift from optimising fields to signals using 3D Turbo Spin Echo (TSE) brain imaging 3T as an example [4].

Methods 3D T2 weighted TSE sequences [4] employ long trains of low refocusing flip angles in order to control apparent T2 decay and thus extend the usable echo train duration. This also reduces SAR, however signal and contrast homogeneity of such sequences are often sensitive to RF inhomogeneity [5]. The SREPG framework allows the calculation of the echo amplitudes as a function of space and echo number, which we write as $S(\mathbf{r},n)=f(\boldsymbol{\theta}(\mathbf{r}),E_1(\mathbf{r}),E_2(\mathbf{r}))$ where $\boldsymbol{\theta}(\mathbf{r})=\{\theta_0(\mathbf{r}),\theta_1(\mathbf{r}),\ldots,\theta_n(\mathbf{r})\}$ is the set of flip angles used ($\theta_0(\mathbf{r})$ is the excitation pulse); $E_{1,2}(\mathbf{r})=\exp\{-ES/T_{1,2}(\mathbf{r})\}$ where ES is the echo spacing; and "f" represents use of the EPG algorithm[2]. The flip angles are written as functions of space such that $\boldsymbol{\theta}(\mathbf{r}) = \mathbf{A}\,\boldsymbol{\sigma}(\mathbf{r})$ where $\boldsymbol{\sigma}(\mathbf{r})$ contains the sensitivity of each element of the array at location $\mathbf{r}$ and $\mathbf{A}_{ij}$ contains the complex drives applied to the j^{th} channel for the i^{th} pulse. RF shimming experiments were performed with two separate approaches:
(i) RF field shimming using magnitude least squares [6] to derive a single (global) RF shim for all pulses
(ii) Signal shimming using SREPG

In the latter the following cost function was minimised using a Non-Linear Conjugate Gradient (NLCG) method: $\min_{\mathbf{A}}\left\{\sum_{n,\mathbf{r}\in\Omega}\left(|S(\mathbf{r},n)|-\mathbf{I}_T\right)^2+\beta\|\mathbf{A}\|_F^2\right\}$ where I_T is a target echo amplitude and β is a regularization parameter designed to limit RF power. The TSE sequence used non-selective RF pulses, TR=4500ms, ES=5.5ms and 106 echoes per shot, with SENSE factor 2 in both phase encoded directions. The base sequence employed a pseudo-steady-state (PSS) with relative echo amplitude of 0.5; the first five refocusing angles achieve transition into the PSS [7] and all subsequent pulses were fixed at 38°. Subsequent optimizations used I_T=0.5 for all locations and echoes. In principle all 106 refocusing pulses plus the excitation could be independently optimised using SREPG. Rather than start with such a large problem, increasing levels of complexity were tested. In the first instance a single scaling factor was optimized, corresponding to a simple RF power scale but with the resulting signal distribution (not measured flip angle) as the driver. The next level of complexity allows for a global RF shim such that the relative weights on each channel may vary but their relationship is fixed through the sequence. More complex still is to allow three separate RF shim settings for (i) the excitation pulse, (ii) the first five refocusing pulses and (iii) the remaining pulses, reflectjng the structure of the sequence with its approach to and maintenance of a PSS. Finally the most complex solution allowed independent optimisation of shims for the first 6 pulses with the 6^{th} shim set also applying to all subsequent refocusing pulses (referred to as "6 Shims"). Note that the issue of maintaining the CPMG condition is sidestepped since the SREPG inherently seeks to maintain echo amplitude; the solution RF fields may have different phases but by design this does not lead to loss of signal. In all cases the optimization neglected relaxation effects (i.e. $T_1=T_2=\infty$), but solutions were subsequently evaluated including relaxation. Imaging was performed using a Philips 3T Achieva MRI system with an 8-channel PTx body coil [8]. B_1 mapping used the AFI sequence [9] with improved RF spoiling [10] and slice selective excitation [11]. The method has been tested on phantoms and healthy volunteers; an in-vivo example is presented.

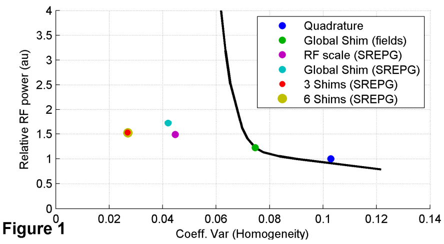


Figure 1

Results & Discussion Figure 1 summarises the RF shim solutions; homogeneity is quantified by the coefficient of variation (CV) of echo amplitudes through space and time. The black line corresponds to solutions possible using RF field based static shimming. Interestingly, using the SREPG to optimise the global scaling of the field optimised static shim improves the overall signal homogeneity beyond what can be achieved with any of the field based optimizations. The resulting scale factor was seen to vary between subjects and does not directly relate to the mean $\boldsymbol{\sigma}(\mathbf{r})$. The SREPG based global shim offers further improvement, while the solutions for 3 and 6 shims are the best and essentially overly one another. Although the optimisation excludes relaxation effects, it was found that the solutions applied when tissues have finite relaxation times provided T_2>5xES. Figure 2 displays predicted contrast between CSF and grey matter for each solution normalised to the expected value. The quadrature excitation and field based shimming solution both produce inhomogenous images and correspondingly large variations in T2 weighting. Simply scaling the RF can improve homogeneity however Fig 2 shows that a large spread in T2 weighting is still apparent. The 3 and 6 shim solutions produce a much more uniform image both in terms of signal and tissue contrast, as seen in Fig 3 where in particular signal in the periphery is increased (arrows).

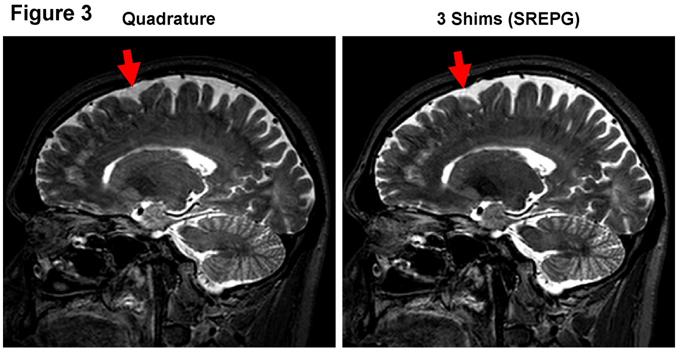
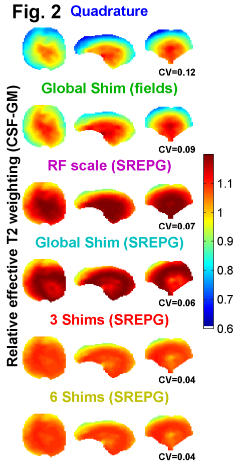
Figure 3


Fig. 2


Conclusions Shimming based on a signal model rather than field purity resulted in improved signal homogeneity and contrast. The direct use of the signal model takes advantage of complex relationships between RF pulses and offers a high degree of flexibility generating uniform signal from field distributions which are themselves not necessarily uniform. The complexity of optimisation can be adjusted to control the computation time. In the case of 3D TSE, joint design of 3 independent shim values for excitation, initial approach to steady state and all subsequent pulses provided a highly effective solution with a limited number of free parameters.

References [1] Ibrahim et al, MRM 2001 19:1339-1347 [2] Hennig J, JMR 1988 78:397-407 [3] Malik SJ et al, Proc ISMRM 2011:208 [4] Mugler JP et al, Proc ISMRM 2000:687 [5] Rozenkrantz AB et al, AJR 2010 194:634-641 [6] Setsompop K et al, MRM 2008 59:908-915 [7] Hennig J et al, MRM 2004 51:68-80 [8] Vernickel P et al, MRM 2007 58:381-389 [9] Yarnykh VL, MRM 2007 57:192-200 [10] Nehrke K, MRM 2009 61:84-92 [11] Malik SJ et al, MRM 2011 65:1393-1399