Simultaneous signal loss correction from B1 and B0 field inhomogeneity in BOLD fMRI with parallel excitation

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Introduction

We propose a novel, fast parallel excitation pulse design method combining a B1 field inhomogeneity correction [1] and a signal loss correction with time-shifted slice selective pulse design [2] to provide uniform signal recovery for T2* weighted imaging in BOLD fMRI. T2* weighted imaging has been the dominant fMRI imaging technique, but it has suffered from a serious signal loss problem around air cavity regions such as the frontal sinus and ears. The signal loss arises from the local through-plane gradient developed in those regions, which brings about unwanted fast signal decay and typically creates dark holes in the reconstructed image. In our proposed method, we transmit a linear combination of time-shifted (kz) slice-selective pulses together with phase-encoding in-plane (kx,ky) gradient waveforms to achieve uniform signal recovery over a specified region of interest. Our method provides a systematic scheme to compute the RF pulse and gradient waveforms based on B0 fieldmap information. Also, it achieves a more uniform signal recovery than [2] by correcting for inhomogeneous RF transmission coil sensitivities.

Theory

In our method, we create an excitation pattern with a pre-compensatory through-plane phase [3] so that the magnetizations along the slice-selection direction are in phase at echo time. For that, we adopt an echo-volumar trajectory [3], and in each kz line of the k-space trajectory we transmit a linear combination of time-shifted sinc pulses weighted by complex gains. Therefore, in our design, the main issue is to determine the complex pulse gain parameters and phase encoding locations in the k-space trajectory. As in [1], the pulse weights can be computed by solving $\tilde{b} = \arg\min_b \|d - Ab\|_2$ where d is the desired 3D pattern with a pre-compensatory through-plane phase based on B0 fieldmap information and echo time as in [3]. For example, $d = e^{i2\pi\omega(r)TE}$ where $\omega(r)$ is B0 fieldmap, and TE is the echo time. The matrix $A = [S_1F, S_2F, \ldots S_RF]$ is a stack of matrices where each S_i is a diagonal matrix of the i-th coil's sensitivity pattern and F is a 3D-Fourier encoding matrix restricted to the support of d. The vector $b = [b_1; b_2; \ldots; b_R]$ is a vertically concatenated vector where each b_i is a vector composed of complex gains of slice-selective pulses transmitted by the i-th coil and its element indices denote phase encoding locations and the amount of time-shift along a kz line. To choose efficient phase-encoding locations, we enforce joint-sparsity on b as in [1] to iteratively choose phase encoding locations. At each iteration, we compute cumulative correlations [1] between unselected phase encoding locations and the residual of the desired pattern to estimate the correlation between a candidate phase encoding location and the residual, pick the one with the highest absolute value, and add it to the set of chosen phase encoding locations. Then we compute the orthogonal projection of the desired pattern onto the set of chosen phase encoding locations, and update the residual. After selecting a predetermined number of phase encoding locations, we connect the chosen phase encoding locations in a spiral-in manner and add the B0 fieldmap information to F to more precisely compute the pulse weights (b) in the optimization problem presented above.

Simulation and Results

To estimate the performance of our design, we set the initial magnetization with the magnitude image we reconstructed from human subject scans and ran a Bloch simulation with our parallel excitation pulse with 8 coils. The echo time is 30ms and the number of phase encoding locations is 12, yielding a 10.7ms long pulse. Slice selection used a hanning windowed sinc pulse with two side lobes for a 5mm thick slice. The desired pattern is a 5mm-thick slice of a 24cm x 24cm x 5mm volume over a 64x64x5 sampling grid. On a computer with Intel Core2 Quad CPU 2.4GHz, 4GB RAM and Matlab 7, we could compute our pulse in about 110 seconds. Fig 1. shows the magnitude pattern of 8 coil sensitivities and Fig 2. shows the B0 fieldmap over the desired pattern. Fig 3. shows the through-plane gradient map of the selected volume. We compared our pulse performance with the regular sinc pulse excitation and, as shown in the Bloch simulation of Fig 4., our pulse shows a good recovery over multiple signal loss regions.

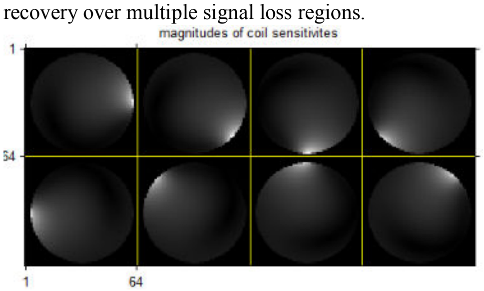

Fig 1. magnitude image of 8 coil sensitivity patterns.

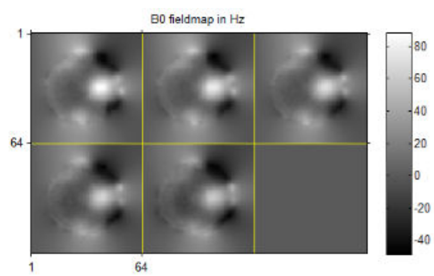

Fig 2. B0 fieldmap in Hz over the excited volume.

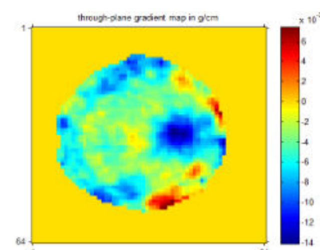

Fig 3. Through-plane gradient map in g/cm

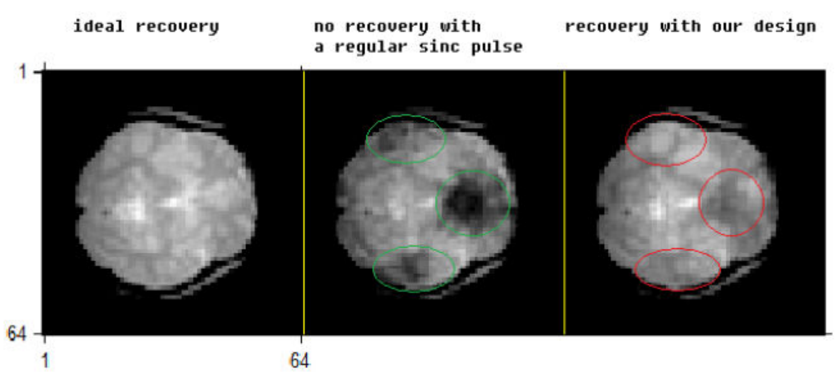

Fig 4. (at left). Signal recovery result and comparison amongst an ideal recovery (left), no recovery (a regular sinc pulse excitation, middle), and recovery with our pulse (right). As shown in this figure, our design shows signal recovery over multiple regions. To aid in the comparison, dominant signal loss regions are circled in green in the no recovery case and recovered regions are circled in red in our design.

Conclusion

We demonstrated that our parallel excitation pulse design can achieve a uniform signal recovery over multiple signal loss regions. Topics for further investigation include acceleration of pulse computation, more efficient algorithms for choosing phase encoding locations, and relaxation of in-plane phase of the desired excitation pattern.

References: [1] Yoon, ISMRM,2009,2595 [2] Stenger, *Mag.Res.Med.*,61(2) : 255-9, Feb.2009. [3] Yip, *Mag.Res.Med.*,56(5):1050-9,Nov.2006