Simultaneous multislice spiral imaging using z-gradient modulation and parallel receive coils

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INTRODUCTION: Simultaneous multislice imaging decreases the required scan time for a given volume, which is particularly beneficial in fMRI, as well as in a variety of other applications. We propose a new method for spiral simultaneous multislice imaging that uses a z-gradient modulation during readout, along with a multiple-coil receive array. Previous methods have used a z-gradient encoding only for Cartesian readouts [1], which do not share the benefits of spirals in fMRI such as improved signal recovery (spiral-in trajectories) and shorter readout times [2]. We use a model-based, iterative reconstruction method similar to [3] and [4] to separate the slices.

THEORY: In simultaneous multislice imaging, the complex signal acquired in coil u is $\mathbf{s}_u = \sum_{v=1}^{l} \mathbf{A}\mathbf{C}_{u,v}\boldsymbol{\rho}_v$, where $\mathbf{A}$ is the Fourier transform operator, each $\mathbf{C}_{u,v}$ is a diagonal matrix consisting of the sensitivity of coil u to slice v, each $\boldsymbol{\rho}_v$ is a vectorized slice v, and l is the number of slices simultaneously acquired. Combining the equations for all coils into one, we have

$$\begin{bmatrix}\mathbf{s}_1\\ \mathbf{s}_2\\ \vdots\\ \mathbf{s}_d\end{bmatrix} = \begin{bmatrix}\mathbf{AC}_{1,1} & \mathbf{AC}_{1,2} & \cdots & \mathbf{AC}_{1,l}\\ \mathbf{AC}_{2,1} & \mathbf{AC}_{2,2} & \cdots & \mathbf{AC}_{2,l}\\ \vdots & \vdots & \ddots & \vdots\\ \mathbf{AC}_{d,1} & \mathbf{AC}_{d,2} & \cdots & \mathbf{AC}_{d,l}\end{bmatrix}\begin{bmatrix}\boldsymbol{\rho}_1\\ \boldsymbol{\rho}_2\\ \vdots\\ \boldsymbol{\rho}_l\end{bmatrix}, \tag{1}$$

where d is the number of coils. Reconstruction of each slice becomes a matter of solving this linear equation for the $\boldsymbol{\rho}_v$ vector. However, if the physical arrangement of the receive coils and slice prescription is such that the sensitivity for each coil is not sufficiently different between slices, then the problem becomes very ill-conditioned. In other words, $\mathbf{C}_{u,v} \approx \mathbf{C}_{u,w}$ for $v \neq w$, and the system matrix in equation (1) contains column-blocks that are very similar to each other. This situation arises in a typical 8-channel head coil setup, where the coils are arranged around the head so that their sensitivities are very similar to different axial slices.

In order to overcome this issue, a z-gradient modulation can be used during spiral readout to help differentiate the columns of the system matrix in (1). The equation becomes

$$\begin{bmatrix}\mathbf{s}_1\\ \mathbf{s}_2\\ \vdots\\ \mathbf{s}_d\end{bmatrix} = \begin{bmatrix}\mathbf{M}_1\mathbf{AC}_{1,1} & \mathbf{M}_2\mathbf{AC}_{1,2} & \cdots & \mathbf{M}_l\mathbf{AC}_{1,l}\\ \mathbf{M}_1\mathbf{AC}_{2,1} & \mathbf{M}_2\mathbf{AC}_{2,2} & \cdots & \mathbf{M}_l\mathbf{AC}_{2,l}\\ \vdots & \vdots & \ddots & \vdots\\ \mathbf{M}_1\mathbf{AC}_{d,1} & \mathbf{M}_2\mathbf{AC}_{d,2} & \cdots & \mathbf{M}_l\mathbf{AC}_{d,l}\end{bmatrix}\begin{bmatrix}\boldsymbol{\rho}_1\\ \boldsymbol{\rho}_2\\ \vdots\\ \boldsymbol{\rho}_l\end{bmatrix}, \tag{2}$$

where each $\mathbf{M}_v$ is a diagonal matrix representing the z-gradient modulation to slice v. For a given z-gradient, the modulation will add $\gamma r_v \int_0^t G_z(\tau)d\tau$ amount of phase to slice v at readout time t, where γ is the gyromagnetic ratio, r_v is the distance of slice v to isocenter, and $G_z(\tau)$ is the magnitude of the z-gradient as a function of readout time. The reliance of the phase modulation on r_v causes the modulation to differ from slice to slice, so that $\mathbf{M}_v \neq \mathbf{M}_w$, for $v \neq w$. The z-gradient modulation can be arbitrary, as long as slew rate and gradient amplitude limits are not breached. However, it is advantageous to choose a $G_z(\tau)$ function that makes the condition number of the system matrix in (2) as low as possible. It may also be important to have the running integral of $G_z(\tau)$ periodically go to 0 to minimize through-plane intravoxel dephasing.

METHODS: Two scans were acquired using a visual stimulus block paradigm: one using the proposed multislice method, and another using conventional single-slice imaging. Each multislice acquisition consisted of two simultaneous 3 mm thick slices. The multislice RF pulse was created by the summation of two Hamming-windowed sincs, one of which was modulated in frequency to create a 20-slice gap in the slice profile between the 2 simultaneous slices. Twenty of these simultaneous slices were acquired per TR, resulting in 40 slices total. A z-gradient was applied during readout, and consisted of 32 cycles of a sine wave with amplitude 0.389 G/cm, which imparted a maximum phase addition of 2π to the location 20 slices away from isocenter. A spiral-in sequence with a TR of 1 s was used. For the conventional scans, forty 3 mm slices were acquired using a Hamming-windowed sinc pulse with bandwidth 1.25 kHz and a spiral-in sequence with a TR of 2 s. For both scans, an 8-channel head coil was used.

The Conjugate Gradient algorithm with 8 iterations and regularization was used to solve (2), where $\mathbf{A}$ consisted of a non-uniform FFT operation for spiral data. Jeff Fessler's image reconstruction toolbox was used for this process [5]. Sensitivity maps were acquired by using the sum-of-squares method on one time frame of the conventional scan.

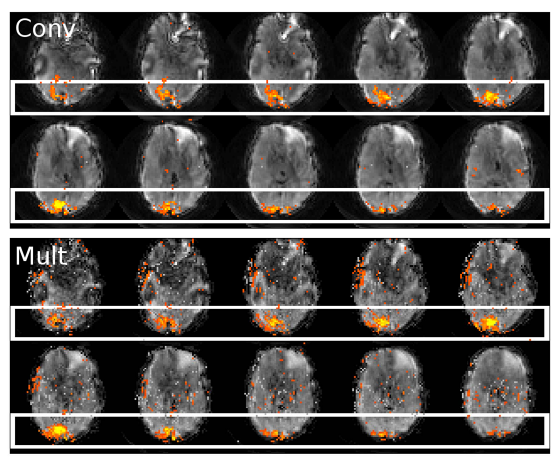

Fig. 1: Activation *t*-score maps of the conventional (top) and the proposed multislice (bottom) scans. A threshold of t = 4 was used to determine activation. Visual cortex areas are encircled in white boxes.

RESULTS: Figure 1 shows activation maps in the visual cortex region for the conventional and multislice scans. The activated voxel count in the areas encircled by the white boxes was 655 and 804 for the conventional and proposed multislice methods, respectively.

CONCLUSION: The feasibility of the proposed method was demonstrated for fMRI, with activation results comparable to those of a conventional scan. Although each acquisition consisted of only 2 simultaneous slices, the method is readily extensible to 3 or more simultaneous slices, with the maximum number depending on the number of receive coils available and the physical configuration of the coils in relation to the slices. The method enables a decrease in volume TR, potentially allowing the imaging of faster activation events, in addition to more effective physiological noise removal. Further investigation is needed to examine the effects of noise in the readout z-gradient, the timing of the gradient, and magnetic field inhomogeneity.

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