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Review

Spiral imaging in fMRI

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ABSTRACT

T2*-weighted Blood Oxygen Level Dependent (BOLD) functional magnetic resonance imaging (fMRI) requires efficient acquisition methods in order to fully sample the brain in a several second time period. The most widely used approach is Echo Planar Imaging (EPI), which utilizes a Cartesian trajectory to cover k-space. This trajectory is subject to ghosts from off-resonance and gradient imperfections and is intrinsically sensitive to cardiac-induced pulsatile motion from substantial first- and higher order moments of the gradient waveform near the k-space origin. In addition, only the readout direction gradient contributes significant energy to the trajectory. By contrast, the spiral method samples k-space with an Archimedean or similar trajectory that begins at the k-space center and spirals to the edge (spiral-out), or its reverse, ending at the origin (spiral-in). Spiral methods have reduced sensitivity to motion, shorter readout times, improved signal recovery in most frontal and parietal brain regions, and exhibit blurring artifacts instead of ghosts or geometric distortion. Methods combining spiral-in and spiral-out trajectories have further advantages in terms of diminished susceptibility-induced signal dropout and increased BOLD signal. In measurements of temporal signal to noise ratio measured in 8 subjects, spiral-in/out exhibited significant increases over EPI in voxel volumes recovered in frontal and whole brain regions (18% and 10%, respectively).

Contents

Introduction

Blood Oxygen Level Dependent (BOLD) functional magnetic resonance imaging (fMRI) (Bandettini et al., 1992; Kwong et al., 1992; Mansfield, 1977; Ogawa et al., 1990) typically uses T2*-weighted imaging to detect changes in deoxyhemoglobin concentration consequent to altered neural metabolism. Because absolute MR image intensities have no intrinsic meaning, relative differences in signal levels (contrast) are utilized to infer neural differences in brain state. In effect, a time series signal of interest is always compared to another time series, which is either a model-based regressor in the case of task-evoked experiments or signals from another brain region in the case of functional connectivity. Motion of the brain can create apparent time-varying contrast differences that, because the BOLD contrast is small, can be interpreted as true activation. It is therefore important that each time frame be acquired with the shortest possible duration, consistent with tradeoffs between desired signal to noise ratio (SNR), brain coverage, and spatial and temporal resolution. This argues for the use of single- or few-shot acquisition methods, such as Echo Planar Imaging (EPI) (Mansfield, 1977) or spiral (Noll et al., 1995).

Unfortunately, such rapid acquisitions are affected by localized off-resonance conditions that result from susceptibility induced field gradients (SFGs), which cause geometric image distortion, ghosting

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and signal dropout, especially in brain regions adjoining air interfaces (Cho and Ro, 1992). These effects are exacerbated by long readout trajectories, and thus another requirement for fMRI acquisitions is to gather the data for each image as quickly as possible. A typical 3.4 mm × 3.4 mm pixel resolution EPI acquisition will have a readout duration of ~30 ms, which is a significant fraction of T2* at 3 T.

In the following, we begin with a discussion of EPI, followed by spiral methods with a comparison to EPI, concluding with a historical perspective on spiral applications to fMRI.

EPI

The typical fMRI acquisition uses EPI, with k-space trajectory and pulse sequence shown in Fig. 1(A). This trajectory covers a Cartesian k-space with lines whose direction alternates as k_y (phase encoding direction) advances from top to bottom as shown. Because of the need to turn the corner at the end of every k_x (frequency encoding or readout direction) line and the fact that much of the line is acquired during ramping of the G_x gradient, the speed in k-space, $d|k|/dt-y|G(t)|$, is not constant (γ = gyromagnetic ratio for protons, $|G|$ is the magnitude of the complex gradient waveform $G_x + i\, G_y$). Furthermore, the energy contributed by the G_y gradient is negligible compared to that of the G_x gradient, so that only one gradient axis (G_x) is active in the traversal of k-space. Both effects increase the duration of the readout acquisition. Note also that rate of k-space traversal is highly asymmetric, i.e., $d|k_x|/dt >> d|k_y|/dt$. Typically, each k_x line is acquired in ~0.5 ms, so the k_x bandwidth is ~2000 Hz/pixel, while the equivalent bandwidth in the k_y direction is ~30 Hz/pixel. In practice, therefore, geometric distortions and ghosting due to off-resonance occur primarily in the phase encoding direction. Nyquist ghosting (Jezzard and Clare, 1999) occurs because imperfections in the gradients cause asymmetry in the center of the k_x readout, and since the direction alternates every other line, the result is a perturbation manifested at a distance of half the field of view (FOV).

Spiral

The spiral k-space trajectory and pulse sequence is shown in Fig. 1(B). First described by Ahn (Ahn et al., 1986), the trajectory starts at the origin and spirals to the edge of k-space, typically using an Archimedean design,

$$\begin{aligned} k(t) &= k_x(t) + ik_y(t) = \frac{N_{\mathrm{int}}\theta(t)}{D} e^{i\theta(t)}, \\ G(t) &= G_x(t) + iG_y(t) = \frac{1}{\gamma}\frac{dk}{dt} \end{aligned} \tag{1}$$

where N_{int} is the number of interleaves, D is the FOV, and θ is a function of time t that describes the azimuthal trajectory sweep. The design of θ is governed by constraints of the gradient system determined by slew rate limit S_0 imposed by either hardware or biophysical considerations of peripheral nerve stimulation, or the gradient amplitude limit, $G_{\max}$:

$$\begin{aligned} &\left|\frac{dG}{dt}\right| \le S_0 \\ &|G| \le G_{\max}. \end{aligned} \tag{2}$$

Eqs. (1–2) result in a differential equation that can be solved for θ by iterative means (King et al., 1995; Meyer et al., 1992) or by various approximations (Duyn and Yang, 1997; Glover, 1999). Generally the trajectory is slew rate limited near $k = 0$, where it begins with zero gradient amplitude (top condition in Eq. (2)). As the amplitude of $|G|$ increases during the readout, it may reach $G_{\max}$, at which the solution switches to one in which the slew rate S_0 can no longer be maintained (second condition in Eq. (2)). This rarely happens in fMRI acquisitions, however. In the early days of fMRI before so-called "EPI gradients" were commonly available, the limited amplitude and slew-rate capabilities required multi-shot spirals to maintain a modest readout trajectory, as shown in Fig. 2.

Comparison of the two pulse sequences in Fig. 1 shows qualitatively that the spiral sequence has a shorter readout duration than that of EPI, which is directly attributable to 1) spiral's use of both G_x and G_y gradient axes to drive the trajectory, as opposed to just G_x for EPI, and 2) not acquiring the k-space corners. The shorter readout reduces off-resonance-induced geometric distortion (in EPI) and blurring (in spiral) as well as SFG-initiated signal dropout (Glover and Lai, 1998) (Yang et al., 1998). Fig. 3 shows T2*-weighted images for EPI and spiral acquisitions. The asymmetry of the EPI trajectory's

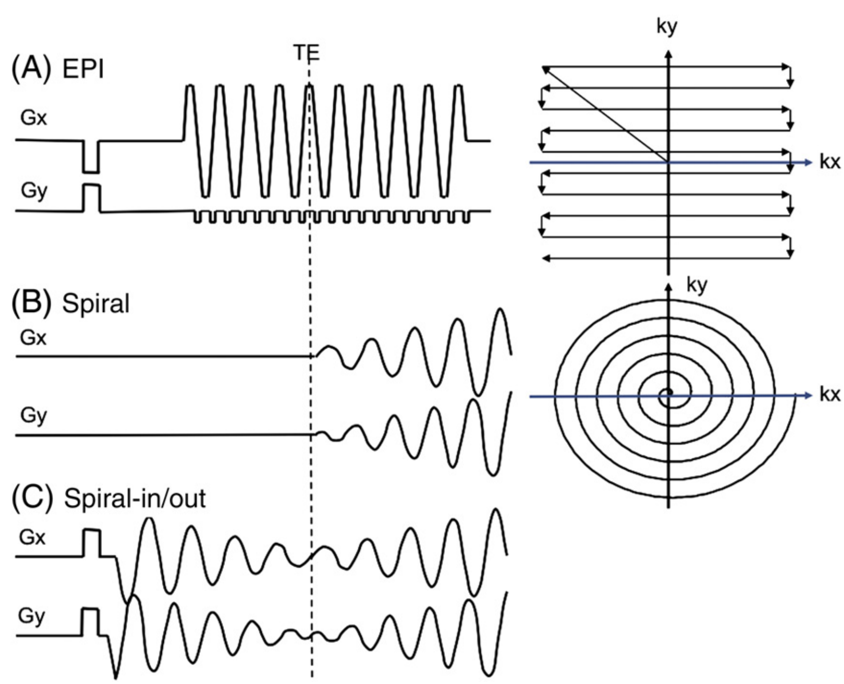

Fig. 1. K-space trajectory and pulse sequence diagrams for (A) EPI, (B) spiral, (C) spiral-in/out.