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Single shot whole brain imaging using spherical stack of spirals trajectories

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ABSTRACT

MR-encephalography allows the observation of functional signal in the brain at a frequency of 10 Hz, permitting filtering of physiological "noise" and the detection of single event activations. High temporal resolution is achieved by the use of undersampled non-Cartesian trajectories, parallel imaging and regularized image reconstruction. MR-encephalography is based on 3D-encoding, allowing undersampling in two dimensions and providing advantages in terms of signal to noise ratio. Long readout times, which are necessary for single shot whole brain imaging (up to 75 ms), cause off-resonance artifacts. To meet this issue, a spherical stack of spirals trajectory is proposed in this work. By examining the trajectories in local k-space, it is shown that in areas of strong susceptibility gradients spatial information is fundamentally lost, making a meaningful image reconstruction impossible in the affected areas. It is shown that the loss of spatial information is reduced when using a stack of spirals trajectory compared to concentric shells.

Introduction

Echo planar imaging (EPI) (Mansfield, 1977) is the established technique for functional magnetic resonance imaging (fMRI). For whole brain imaging EPI has a temporal resolution of typically 2–3 s. This is sufficiently fast compared to the blood oxygenation level dependent response (BOLD) (Ogawa et al., 1990), but there are several issues making it desirable to achieve high temporal resolution for fMRI. A single event hemodynamic response function (HRF) has significant signal fluctuations lasting approximately 20 s. Therefore about 10 data points are measured in the meantime when acquiring with a standard EPI protocol. With MR-encephalography (MREG) 200 data points are supplied for the same period of time, allowing a better analysis of the onset and shape of the HRF (Zahneisen et al., 2011). Furthermore the increased amount of data points improves the statistical power from which single event fMRI can benefit (LeVan et al., 2012), as well as the study of functional connectivity of networks (Lee et al., 2013; Lin et al., 2008). Fast acquisition also allows direct filtering of respiration and cardiac artifacts, since they are not aliased in the frequency domain anymore (Hennig et al., 2007; Posse et al., 2012).

While in the original implementation of MREG (Hennig et al., 2007) only coil sensitivities were used for spatial encoding, there have been several approaches for combining a small amount of gradient encoding with the spatial information multi coil arrays provide. An early attempt was to perform a fully sampled sagittal 2D-EPI measurement, while encoding the third dimension purely by coil sensitivities, which is referred to as inverse imaging (Lin et al., 2006). Further approaches were to use a small number of projections for 2D functional imaging (Grotz et al., 2009) or a rosette k-space trajectory for single shot 3D imaging (Zahneisen et al., 2011).

For all single shot acquisition techniques (and this includes standard EPI) the image quality is strongly affected by susceptibility induced local field inhomogeneities. For EPI this is reasonably well understood and leads to the well-known susceptibility artifacts — primarily signal attenuation and geometric distortions. Distortions predominantly occur along the phase encoding direction and can be corrected with suitable methods (Jezzard and Balaban, 1995).

For non-Cartesian trajectories the off-resonance behavior is more complex. As outlined by (Zahneisen et al., 2012), self-intersecting trajectories like rosettes (Zahneisen et al., 2011) and single shot radial trajectories (Hugger et al., 2011) suffer from a high sensitivity to off-resonance, T_2^*-decay and gradient imperfections. Concentric shells (Zahneisen et al., 2012) are designed to have no intersections. They have a more benign off-resonance behavior, allowing longer readout times and therefore higher spatial resolution, while keeping the temporal resolution below 100 ms.

Like rosettes, concentric shells sample k-space symmetrically. As a consequence of the symmetry their point spread functions (PSF) do not show any off-resonance dependent shifts and therefore no image distortions. On the downside off-resonance leads to blurring and signal dropout. In practice geometric distortions are easier to

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handle compared to blurring since they can be corrected later on with suitable coregistration procedures.

Therefore and in order to maintain the comparatively beneficial off-resonance behavior of EPI, a single shot stack of spirals (SoS) technique is proposed. Sampling the k-space monotonously from one end to the other in variable sized steps, while acquiring variable density spirals at each step, the slow encoding direction follows one of the Cartesian k-space coordinate axes. Each individual spiral is acquired sufficiently fast, avoiding appreciable artifacts in the other directions. Therefore the main effects occur along a single direction, similar to EPI.

In the following methodology, implementation and first results of SoS will be presented together with a more detailed discussion of B_0 inhomogeneity effects in non-Cartesian imaging.

Theory

Trajectory design

With the polar coordinates of k-space r and ϑ, a variable density spiral (Spielman et al., 1995) can be expressed with the complex function

$$k_{xy}(t) = k_x(t) + ik_y(t) = r(t)e^{i\vartheta(t)}. \tag{1}$$

k_x and k_y denote the Cartesian coordinates of the trajectory in 2D k-space. Time is indicated by t.

The Nyquist constraint in the radial direction is given by

$$\frac{dr}{d\vartheta} = \frac{R_r(r, |k_z|)}{FOV_{xy}}, \tag{2}$$

where R_r is the radial undersampling factor of the variable density spiral. It is chosen to increase with $|k_z|$ as well in order to shorten the trajectory. Nyquist sampling is described by $R_r = 1$ and undersampling by $R_r > 1$. The in-plane field of view is indicated by FOV_{xy}, which is isotropic for spiral imaging. Dividing both sides by the differential of the time, Eq. (2) can be rewritten as

$$\dot{\vartheta} = \frac{FOV_{xy}}{R_r}\dot{r}. \tag{2'}$$

The first and second derivative of r with respect to time are indicated by $\dot{r}$ and $\ddot{r}$, respectively.

The Nyquist constraint in the angular direction is not discussed here, since in the readout direction it can be fulfilled by choosing a sufficiently short dwell time for the entire trajectory.

Applying the Nyquist constraint (2′) to the first temporal derivative of Eq. (1), the gradient strength can be derived:

$$G_{xy} = \frac{1}{\gamma}\frac{dk_{xy}}{dt} = \frac{e^{i\vartheta}}{\gamma}\dot{r}\left(1 + ir\frac{FOV_{xy}}{R_r}\right). \tag{3}$$

γ denotes the gyromagnetic ratio. Differentiating the gradient strength with respect to time and applying Eq. (2′) again leads to the slew rate

$$S_{xy} = \frac{dG_{xy}}{dt} = \frac{e^{i\vartheta}}{\gamma}\left(i\dot{r}^2\frac{FOV_{xy}}{R_r}\left(2 + ir\frac{FOV_{xy}}{R_r} - \frac{r}{R_r}\frac{\partial R_r}{\partial r}\right) + \ddot{r}\left(1 + ir\frac{FOV_{xy}}{R_r}\right)\right). \tag{4}$$

Note that R_r is a function of $r(t)$ and k_z, where k_z is independent of time within a single spiral.

The absolute values of gradient strength and slew rate do not depend on ϑ. They are set to exploit the system limits. The differential equation that results from the gradient strength constraint is solved iteratively for $\dot{r}$, while the equation resulting from the slew rate constraint is solved for $\ddot{r}$. Integration of the latter solution results in $\dot{r}$ as well. For each iteration the solution with the smaller $|\dot{r}|$ is taken in order to fulfill both maximum gradient strength and maximum slew rate constraints. The Nyquist constraint (Eq. (2′)) is used to calculate $\dot{\vartheta}$. After integration over time the two dimensional spiral trajectory in entirely defined. A more detailed description of the design of spirals can be found in (Glover, 1999; Spielman et al., 1995).

As shown by (Hebrank and Gebhardt, 2000), peripheral nerve stimulation increases with the area of the gradient pulse, therefore with the radius of the spiral r. In order to reduce stimulation, the slew rate is reduced by $S_{xy}' = S_{xy} \cdot e^{-\chi r}$. The empirical factor $\chi = 0.2$ m is obtained by simulating the peripheral nerve stimulation with software provided by the manufacturer (Hebrank and Gebhardt, 2000). The reduction factor allows sampling the center of k-space with the maximal slew rate of the system. At the outer k-space the slew rate is reduced in order to achieve a more constant stimulation level over time.

Spirals are calculated for each k_z in steps of

$$\Delta k_z = \frac{2\pi \cdot R_z(|k_z|)}{FOV_z}, \tag{5}$$

where $R_z(|k_z|)$ indicates the undersampling factor in z-direction.

The maximum radius of each spiral is set to result in a spherical surface with the radius $\left|\vec{k}_{\max}\right| = 2\pi/voxel\ size$, where $\vec{k} = (r, \vartheta, k_z)^T$ represents the trajectory in 3D k-space.

The acquisition direction of the spirals alternates between inside out and outside in, with the central spiral ($k_z = 0$) always inside out in order to maintain a reasonable echo time. The spirals are rotated in-plane so that the end of one spiral is aligned with the beginning of the next one. Thereafter the endings of the spirals are bent to obtain a smooth connection. The time optimal connection is calculated using a linear programming algorithm (Hargreaves et al., 2004).

Fig. 1(a) shows a resulting spherical stack of spirals trajectory. The slew rate with and without the reduction factor are displayed in Figs. 1(b) and (c), respectively.

Image reconstruction

Image reconstruction is based on solving the inverse problem. The signal equation is

$$b_c(t) = \int d\vec{x}\rho\left(\vec{x}\right)\cdot S_c\left(\vec{x}\right)\exp\left(i\vec{k}(t)\ \vec{x} + i\omega\left(\vec{x}\right)t - T_2^*\left(\vec{x}\right)t\right) \tag{6}$$

with the signal b of the coil element c. The macroscopic transversal magnetization directly after excitation is denoted by ρ, S_c is the coil sensitivity. $\vec{k}$ represents the k-space trajectory, $\vec{x}$ the image space and ω the off-resonance frequency. In matrix form, this equation can be approximated by

$$\vec{b} = \mathbf{E}\,\vec{\rho}\ . \tag{7}$$

Here $\vec{b}$ includes the signal of all coils at all acquired, discrete time points and $\vec{\rho}$ is the vectorized image. **E** denotes the forward operator. Using the complete matrix as a forward operator would be computationally impractical. Therefore, **E** is implemented as an operator that first multiplies the image with the coil sensitivities and thereafter performs a non-uniform fast Fourier transformation (NUFFT) with min-max interpolation (Fessler and Sutton, 2003) for each coil. This implementation disregards off-resonance and T_2^* -decay.

In order to account for off-resonance while keeping the computational benefits of the NUFFT, a segmented approach was applied, as described by (Sutton et al., 2003; Zahneisen et al., 2011). This approach applies of the NUFFT to each segment and uses Hann window functions