Correction of gradient-induced phase errors in radial MRI

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Target Audience: Physicists who are interested in non-Cartesian imaging scenarios.

Purpose: Gradient-induced eddy currents affect the MRI data acquisition by gradient delays and phase errors that may lead to severe image artifacts for non-Cartesian imaging scenarios such as radial trajectories[1,2]. While several methods have been proposed to correct the effect of linear eddy-current in radial MRI[3,4], B0 eddy currents have usually been neglected. In small-bore high-field MRI systems, the artifacts caused B0 eddy currents become more severe. Therefore, the main goal of this work is to correct phase errors induced by eddy currents in radial MRI.

Methods: The accumulated phase due to eddy currents can be described as:

$$\phi(\vec{r},t) = \gamma \int_0^t B_{ec}(\vec{r},t)dt.$$

where r is the location of the sample and B_{ec} the field inhomogeneity induced by eddy currents. Consequently each acquired spoke in k-space is affected by a global phase error:

$$\phi(\theta) = \psi_x G_x \cos(\theta) + \psi_y G_y \sin(\theta) + \phi_0$$

with ψ_x and ψ_y the phase errors per gradient strength in units of rad m mT^{-1} as well as G_x and G_y the gradient strengths in units of mT m^{-1}. The term Φ_0 describes a constant phase offset summarizing contributions that are not caused by the in-plane gradients. If Φ is experimentally known for a variety of spokes, then ψ_x, ψ_y and $\o_0$ can be estimated by fitting the equation to the measured raw data. Accordingly, the phase error can be estimated and corrected by multiplying the raw data with the complex conjugate of the estimated phase error.

All measurements were performed on a 9.4 T 30 cm-bore MRI system (Bruker Biospin, Ettlingen, Germany) with a Bruker BGA12S gradient system yielding a maximum gradient strength of 660 mT m^{-1} and slew rate of 4570 T m^{-1} s^{-1}. T1-weighted data sets were obtained using a RF-spoiled radial FLASH sequence (201 spokes, TR/TE = 10/3 ms, flip angle = 15°, FOV = 24 × 24 mm^2, spatial resolution = 150 × 150 × 500 µm^3, acquisition bandwidth = 50 kHz, read gradient strength of 60.7 mT m^{-1}). These parameters are referred to as reference condition throughout this work. The linear eddy-current effect of the radial acquisition (i.e., the k-space shift) was corrected by multiplying the dephase gradient with a correction factor. Measurements with variations of the echo time (5 ms), acquisition bandwidth (200 kHz) and in-plane resolution (200 µm^2) were acquired to evaluate the effect of the acquisition protocol on the estimated parameter ψ_x, ψ_y and $\o_0$. These modified acquisition parameters corresponded to gradient strengths of 60.7, 242.6, and 45.5 mT m^{-1}, respectively.

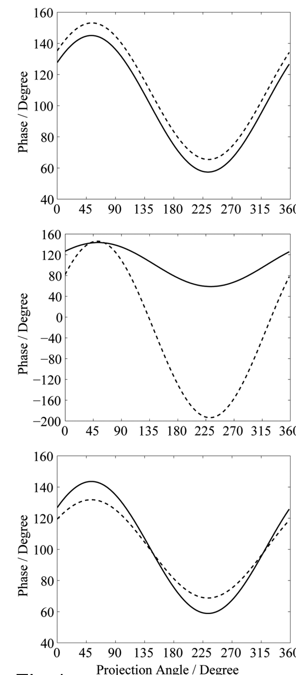

Fig.1

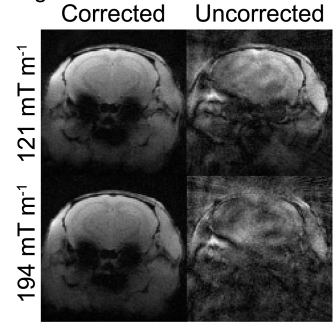

Fig.2

Results: Figure 1 demonstrates that the phase error derived from the central data point of each spoke in a high-field radial MRI acquisition follows a sinusoidal form when plotted as a function of radial projection angle. Also the sensitivity of the phase error to several acquisition parameters is demonstrated in Fig. 1 (solid lines refer to the reference condition). The estimated values are summarized in Table 1. The prolongation of the echo time from 3 ms to 5 ms only affected Φ_0 with a significant change from 98.08° to 109.27° (Fig. 1, top). Increasing the bandwidth or decreasing the resolution leads to a stronger or weaker gradient, respectively (Fig. 1, middle and bottom). Although these conditions cause large differences in phase error relative to the reference condition, the estimated values for ψ_x and ψ_y are only mildly affected. Fig. 2 illustrates the image artifacts that result from eddy-current induced phase errors for in vivo mouse brain studies with two different gradient strengths.

Discussion: For Cartesian imaging the phase error induced by B_0 eddy currents poses no major problem. On the other hand, radial trajectories require a sinusoidal modulation of the frequency-encoding gradients as a function of projection angle. The resulting phase error exhibits a well-defined relationship which may be characterized by the parameters ψ_x and ψ_y as well as a gradient independent phase error Φ_0. Respective image artifacts may then be removed by multiplying the raw data with the complex conjugate of the phase error as given by the pre-determined parameters ψ_x, ψ_y and $\o_0$. Because of the dependence of the eddy-current phase error on the gradient strength, the phase problem becomes more important for small-bore high-field MRI systems equipped with strong gradients for high-resolution animal studies.

Parameter	Gradient Strength mT m^{-1}	ψ_x rad m mT^{-1}	ψ_y rad m mT^{-1}	$\o_0$ Degree
Reference Condition	60.7	2.82	3.67	98.08
Longer Echo Time	60.7	2.75	3.70	109.27
Higher Bandwidth	242.6	2.90	3.60	-23.67
Lower Resolution	45.5	2.77	3.64	100.28

Conclusion: A phase error correction method is introduced that is able to distinguish between and separately correct for phase errors due to eddy currents caused by the in-plane gradients and other phase alterations which are constant for an image and that may carry relevant physiologic information. A phase error correction method is introduced that is able to distinguish between and separately correct for eddy current effects and other phase alterations that may carry relevant physiologic information. Moreover, the proposed method is easy to implement and operates without reference scans. Its application appears mandatory for imaging scenarios using strong magnetic field gradients.

References: **1.** Shankaranarayanan A, Wendt M, Lewin J, Duerk J. Two-Step Navigatorless Correction Algorithm for Radial k-Space MRI Acquisitions. Magn Reson Med 2001;45:277–288. **2.** Brodsky EK, Klaers JL, Samsonov AA, Kijowski R, Block WF. Rapid measurement and correction of phase errors from B_0 eddy currents: Impact on image quality for non-Cartesian imaging. Magn Reson Med 2012; doi:10.1002/mrm.24264. **3.** Duyn JH, Yang Y, Frank JA, van der Veen JW. Simple correction method for k-space trajectory deviations in MRI. J Magn Reson 1998;132:150-153. **4.** Peters DC, Derbyshire JA, McVeigh ER. Centering the projection reconstruction trajectory: Reducing gradient delay errors. Magn Reson Med 2003;50:1-6.