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Technical notes

Intrinsic correction of system delays for radial magnetic resonance imaging☆

M. Krämer *, J. Biermann, J.R. Reichenbach

Medical Physics Group, Institute of Diagnostic and Interventional Radiology, Jena University Hospital-Friedrich Schiller University Jena

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ABSTRACT

Introduction: When using radial MR image acquisition techniques gradient or sampling delays due to hardware imperfections can cause mismatch between the expected and the actual *k*-space trajectory along the readout direction. To provide a robust and simple correction of such system delays we developed a new calibration method which is independent of using any reference data or applying sequence modifications.
Material and methods: Radial data obtained with 180°, 360° and golden-angle radial ordering schemes were deliberately shifted along the readout direction for a discrete range of gradient delays. Following 2D regridding, images were reconstructed and analyzed in image space for all applied shifts to estimate the optimal system delay. Phantom and *in vivo* measurements were performed to test the robustness of the algorithm.
Results: Using the 360° and golden-angle radial ordering schemes system delays in the range of 3.3 μs to 6.3 μs were estimated and corrected for several imaging applications and different conditions, including cardiac and real-time MRI as well as multiple acquisitions using different imaging parameters and slice orientations. When using the standard 180° radial acquisition scheme no automated correction was possible. With a mean computation time of 23.2 ± 14.0 s for the delay estimation computational demands were moderate allowing implementation of the algorithm on the image reconstruction system of any modern MR system.
Conclusion: We have demonstrated that radial data acquired with a 360° or golden-angle ordering scheme can be used for reliable intrinsic correction of system delays. The proposed technique enables a per-scan correction of system delays without the need for additional calibration data or modifications of the radial imaging sequence.

1. Introduction

With recent advances in computation and image processing capabilities non-Cartesian MR image acquisition techniques using spiral [1], radial [2] and PROPELLER [3] trajectories are increasingly used for a multitude of MR imaging applications [4–8]. However, these techniques are very sensitive against imperfections of the gradient and sampling hardware of the MR system that result in discrepancies between the expected and the actual *k*-space scanning trajectory. These deviations from the ideal expected trajectory can have multiple causes, including eddy currents and gradient delays or other system imperfections like heating effects and internal synchronization errors. With no correction of these delays non-Cartesian MR images are usually highly degraded.

Several corrective methods have been developed that are based on modifying the imaging sequence to acquire additional calibration data for estimating the actual *k*-space trajectory [9–13]. It has also been shown that gradient delays can differ between the physical gradient axis making individual correction necessary [14,15]. In general, these correction techniques include measurements of homogenous phantoms for image phase analysis, modifications of the pulse sequences or additional hardware for direct gradient waveform estimation. Especially with approaches based on phantom measurements, complex pulse sequences or sophisticated data analysis, calibration scans often need to be performed regularly on a daily basis or during scanner service which may be tedious and time consuming. Brodsky et al. [16] have shown that small scale changes of system delays can even occur during the day or during extended measurement sessions, requiring fast and reliable per-scan calibration.

Since these troublesome system delays depend on a wide range of hardware and sequence parameters as well as imaging conditions, the ideal delay calibration scan and analysis should take only a few seconds without making complex changes of the measurement sequence necessary. Furthermore, estimation of the delays from the calibration data should be subject and scanner independent and should provide reliable results for different anatomical regions.

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* Corresponding author at: Dipl.-Phys. Martin Krämer, Medical Physics Group, IDIR, Jena University Hospital, Philosophenweg 3, D-07743 Jena, Germany. Tel.: +49 3641 9 35827.
E-mail address: MartinKraemer84@gmail.com (M. Krämer).

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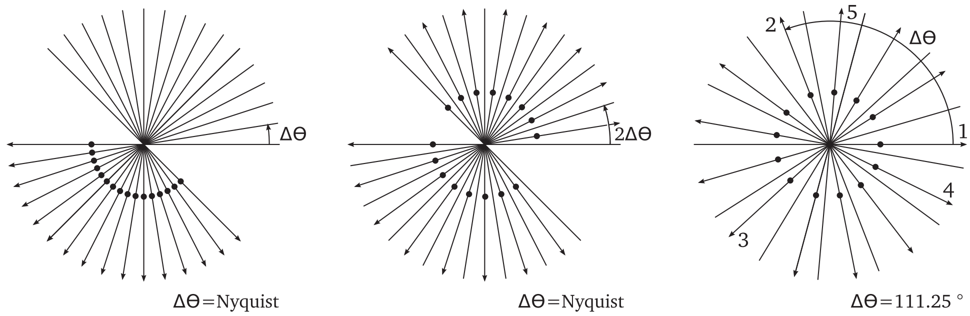

Fig. 1. *From left to right*: 180°, 360° and golden-angle radial ordering schemes. In the presence of a system delay the position of the *k*-space center (dots) is shifted along the readout direction (arrows). For the 360° and golden-angle schemes this results in a symmetric shift pattern encompassing all directions in *k*-space.

To meet these needs we have developed a per-scan calibration method for radial acquisitions, which works by analyzing the MR raw data acquired with the imaging sequence and requires no additional calibration data or sequence modification. To demonstrate the reliability of the technique we tested our approach with several subjects and different anatomical regions.

2. Methods

2.1. Sequence design

A standard 2D imaging sequence with symmetric radial readouts rotated around the *k*-space center was implemented using the open source object-oriented development interface for NMR (ODIN) [17]. In contrast to the widely used 180° temporal ordering scheme the sequence inverts the readout direction of every other spoke and thus creates an ordering scheme with spoke angles ranging from zero to 360° [18]. The sequence also allows to acquire radial data with a rotation scheme using angles based on the golden-ratio [19]. All three ordering schemes together with the resulting impact of system delays on the regridded *k*-space center are illustrated in Fig. 1. For each radial acquisition a delay will cause a shift of the *k*-space center along the readout direction. However, after 2D-regridding on a Cartesian *k*-space the radial 360° and golden-angle ordering schemes will result in a circular, symmetric blurring of the center of *k*-space whereas the standard 180° ordering scheme will produce asymmetric center blurring biased towards one half of *k*-space. The way the *k*-space center is blurred by these delays, however, strongly influences the resulting artifacts in image space.

2.2. System delay estimation

For any MR scanner, the system's specific delays can be separated into an orientation dependent part which varies with the actually active physical gradient axis [14] and a major contribution which is constant on all gradient axes. To estimate the constant system delay a subset of the radial data $S(k)$, e.g., center slice or data from the first measured repetition, is shifted along the readout direction by modulating linear phase gradients into the corresponding inverse 1D *Fast-Fourier-Transform* (FFT) $\mathcal{F}^{-1}$ of the readouts:

$$S_{\Delta k}(k) = \mathcal{F}\left\{\mathcal{F}^{-1}[S(k)] \cdot e^{-2\pi i \cdot x \Delta k}\right\}, \tag{1}$$

where Δk is the deliberately applied *k*-space shift of the raw data. This modulation is sequentially performed for N linearly spaced values of Δk spanning from zero, i.e. no shift, to a maximum shift which should be characteristic for the actual MRI system. If typical delay values for the used MRI system are not known a one-time calibration scan of a homogeneous water phantom can be performed in order to characterize the gradient system [14]. The modulation of the raw data could alternatively be performed during the gridding process by shifting the coordinates of the readout points. This, however, requires recalculation of the grid weights for each modulation step, which increases computation time substantially. Following shift modulation all radial readouts belonging to the same Δk and receive channel are combined to a 2D Cartesian *k*-space. Applying 2D inverse FFT a total of N images $I_{\Delta k}(x,y)$ are obtained for each receive channel, exhibiting artifacts to varying levels that depend on Δk. Prior to further analysis the receive channels are combined using *sum-of-squares* channel combination. To detect the image with the lowest artifact level the sum of the magnitude of all image voxels is calculated, yielding the optimal shift correction Δk_{opt} from the local maximum of the calibration function

$$f(\Delta k) = \sum\nolimits_{x,y} |I_{\Delta k}(x,y)|. \tag{2}$$

closest to $\Delta k = 0$. With the estimated optimal shift, Δk_{opt}, correction of the entire radial dataset is then performed by shifting the *k*-space positions of all radial readout points along the readout direction during the 2D gridding process or by applying Eq. (1) using Δk_{opt}. The complete algorithm is schematically depicted in Fig. 2. With the optimal readout shift in k-space Δk_{opt}, the dwell time d_t and the field-of-view *FOV* of the measurement the actual system delay Δt_{opt} can be obtained from

$$\Delta t_{opt} = \Delta k_{opt} \cdot FOV \cdot d_t. \tag{3}$$

Based on this relation all previously discussed equations can also be expressed as a function of Δt (e.g. $S(t)$ and $f(\Delta t)$), describing the problem as a time delay Δt between sampling and gradient hardware.

For the used MR system and radial measurement sequence the modulation was performed for $N = 40$ linearly spaced values of Δt ranging from zero to a maximum shift of 10 µs. For radial data combination 2D gridding with iterative sampling density estimation with an optimized kernel [20] was applied. To this end, the available source code for 3D gridding from [20] was modified to work in two dimensions. Grid weights for sampling density correction were calculated using 12 iterations. In order to use an optimized gridding kernel and to minimize side lobes caused by the gridding process, grid oversampling of factor two (R = 2) was used. Additionally, the effective grid support matrix was increased by a factor of 1.5 during gridding. The field of view increase resulting from grid oversampling as well as the increased grid support matrix was removed in image space after 2D inverse FFT. Receive channels were combined in image space using the *sum-of-squares* method. Processing of the radial MRI data and image reconstruction were performed offline using Matlab (2011b, The MathWorks, Natick, MA). Computation time for delay estimation