Robust radial imaging with predetermined isotropic gradient delay correction

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Introduction

There is continued interest in Non-Cartesian (NC) acquisition schemes due to their promise of higher SNR and PAT acceleration factors and better motion insensitivity than standard Cartesian schemes. However, in clinical routine Cartesian imaging is used virtually exclusively. The main reason for this lack of routine use, besides the increased complexity of reconstruction algorithms, is probably that NC imaging requires more precise realizations of k-space trajectories than Cartesian imaging, which in turn requires higher precision in scanner control. For example, timing delays between reference clock events (NCO) and receive events (ADC) have been shown [1] to be detrimental to the quality of radially acquired images but not to Cartesian images. The system imperfections that degrade NC image quality most are delays between played out gradient moments (M_G) and ADCs. D. Peters et. al. [2] demonstrated that these gradient delays can be described by a model that assumes that each physical gradient axis has its own delay value.

Measuring trajectories frequently before taking the actual image is adequate only in an academic setting, but quite impossible in a clinical routine setting. Therefore the goal of this paper is to parameterize the delays between M_G and ADC with sufficient accuracy for clinical radial imaging applications.

Methods

Gradient delays were measured on the following Magnetom systems (Siemens AG, Erlangen): Symphony, Sonata, Trio, Avanto and Espree.

The principal method for measuring the trajectory was to determine the position of the k-space center both in phase encode and read out direction with sub pixel accuracy from Cartesian gradient echo images by approximating the echo maximum with a two-dimensional quadratic model. The echo maximum represents the k-space center for all slice orientations if the following conditions are fulfilled: 1. B1 field lines inside the object are straight and parallel, 2. the imaged object is isotropic, and 3., the imaged object contains only one chemical species. Therefore we used a spherical water phantom, placed in the magnet's isocenter and measured with the body coil. The measurements covered most of the value ranges of RO-relevant parameters (FOV, receiver bandwidth and base resolution) that were available in the product Cartesian gradient echo sequences. For some systems measurements were repeated after several months to prove parameter stability. In order to process a large number of measurements, a system of analysis scripts was written in Matlab (The MathWorks, Inc.)

Results

In a first step the position of the k-space center was measured for many different read out and slice normal orientations. The results were consistent with separate delay values for each physical gradient axis [2]. Therefore the following measurements were limited to the read out direction being parallel to one of the three physical gradient axes. For these cases the k-space center is shifted in RO direction only.

Fig. 1 shows as an example gradient delays versus twofold oversampled receiver dwell time t_{os} for a Magnetom Avanto, (a) for the three physical axes, and (b) for three FOV before and after applying the correction.

Observation 1: The delay difference between the different gradient axes are always smaller than t_{os}. Therefore this difference will be neglected and only the isotropic average delay will be compensated.

Observation 2: The delays variations with FOV are much smaller than t_{os}. Therefore the dependency of the gradient delay value on RO gradient amplitude can be neglected.

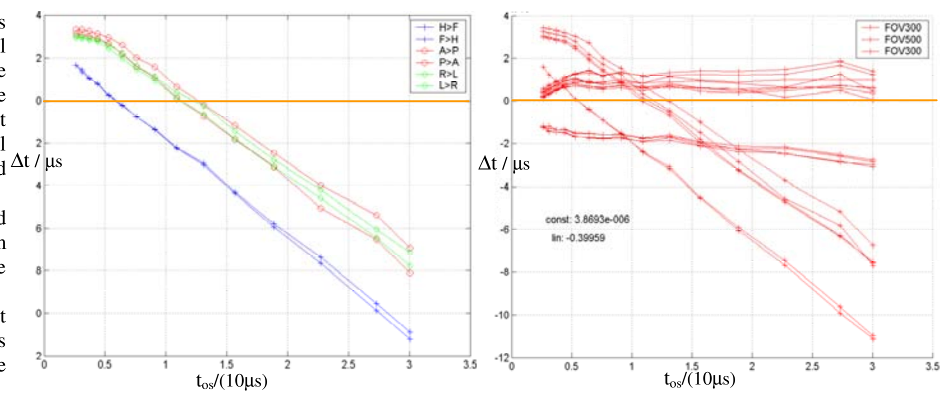

Figure 1: measured RO echo shift Δt versus t_{os} (a) for different RO directions and (b) for different FOV, uncorrected and corrected with the linear approximation to the isotropic average.

Observation 3: The isotropic average can be approximated well with a linear equation $\Delta t = A * t_{os} + B$. Results for the different scanner types are shown in table 1.

The slope A is remarkably close to the NCO-ADC delay slope of 0.5[1], suggesting that a delay in the receiver is the main source of both delays (NCO-ADC and M_G-ADC). Repeat measurements after several months yielded identical results within the measurement accuracy.

Application

The isotropic gradient delays were corrected for in radial cine gradient echo sequences (balanced and unbalanced) by adjusting the RO prephaser (and dephaser). Phantom and volunteer experiments showed that the correction produces near perfect image quality for all receiver bandwidths, proving that the isotropic predetermined correction is sufficient to provide robust clinical radial imaging on the tested scanners. As an example Figure 2 presents radially acquired double oblique cardiac images measured with and without the correction.

System	A	B / µs
Avanto	0.39	3.59
Espree	0.48	3.27
Trio, a TIM system	0.42	2.45
Symphony, a TIM system	0.39	3.27
Trio	0.40	1.60

Table 1: coefficients for the linear approximation $A * t_{os} + B$ for the isotropic gradient delay Δt

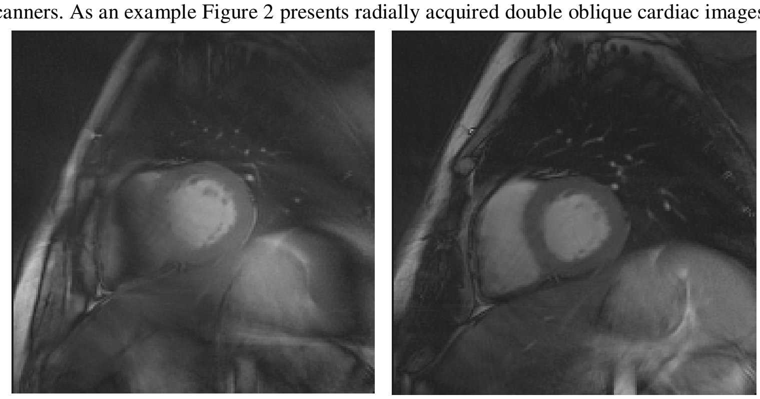

Figure 2: radial cardiac short axis TrueFisp (bSSFP, FIESTA) images, (a) without, (b) with gradient delay correction

References

[1] P. Speier, and F. Trautwein: ISMRM2005, poster 2295

[2] D.C. Peters, J.A. Derbyshire, and E.R. McVeigh: Magnetic Resonance in Medicine 50:1-6, 2003