

Iterative Trajectory Correction for Radial Projection Imaging

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Introduction: Radial trajectories are widely used in MRI, as they allow for effectively undersampling the data acquisition process (e.g. [1]). However, eddy currents and gradient timing errors often cause deviations between the desired and the actually measured points in k-space, leading to artifacts in the reconstructed images. Additional scans to calibrate the system in order to correct for these errors are time-consuming and do not take into account any changes in-between both acquisitions. In this work, we propose a technique to correct for trajectory errors of radial acquisitions that does without any additional calibration data. The method uses only self-consistency in oversampled k-space regions in order to determine the true coordinates of the measured trajectory.

Theory: The procedure is based on small variations in the locations of acquired k-space-points (k_x, k_y) using GRAPPA operator shifts [2] similar to [3]. The algorithm is initialized by individually gridding each of the acquired projections into a separate Cartesian k-space using self-calibrating GROG [4]. In addition, GROG is used to grid all available projections into one Cartesian grid, as usually performed for a standard reconstruction of non-Cartesian data. In the center of this so-called *averaged k-space*, entries represent the average over the entries of several radial projections. Hence, it is considered as “closer to the true k-space” in this region compared to the respective points of the individually gridded projections.

The correction method compares the entries of those individual projections under small translations in k-space to the entries at the initial positions of the averaged k-space in order to find a maximized accordance and thus an improved set of coordinates (k_x, k_y). Within each iteration, every individually gridded projection is simultaneously shifted in 9 different directions (including the zero-shift) in the two-dimensional k-space using the GRAPPA operator. The shift-size is chosen smaller than half the size of a discrete element of the Cartesian grid ($0.5\Delta k$). The translation vector leading to the smallest difference between the entries of the shifted projection and those at the initial positions of the averaged k-space is used to update (k_x, k_y). Performing this optimization for every acquired projection represents one iteration step.

The subsequent iteration is initialized using the updated coordinates (k_x, k_y) to calibrate new GROG weights and to grid new single-projection k-spaces and a new averaged k-space. This procedure is performed until the accordance of none of the projections can further be improved by any shift which means that the true coordinates have been found (see Fig. 1).

Materials & Methods: A validation study was conducted using a numerical phantom (200 projections, 128 readout points) with artificial Gaussian noise and eight different coil sensitivities to simulate a phased-array acquisition. The coordinates were corrupted with a random translation of each projection up to the distance between two subsequent readout points (Δk) in order to simulate trajectory deviations. Figure 2a shows the reconstruction using GROG with the falsified trajectories. Subsequently, the proposed correction algorithm was applied using a shift-size of $0.1\Delta k$ followed by an additional run with a step-size of $0.01\Delta k$ after convergence (steady state). Additionally, the method was applied to an *in-vivo* 2D radial acquisition of the cardiac function (short axis view, Magnetom Trio, Siemens, Erlangen, bSSFP, 390 projections, 256 readout points) using a step-size of $0.1\Delta k$. The non-corrected GROG reconstruction is depicted in Fig. 2c.

Results & Discussion: The results of the reconstructions after applying the proposed trajectory correction are depicted in Fig. 2b (phantom) and Fig. 2d (*in-vivo*). For both examples, the algorithm proved to be capable of improving the image quality. The majority of the streaking artifacts present in the phantom image which were introduced by the corrupted projections were removed. In the *in-vivo* image, the apparent artifacts (e.g. red arrow in Fig. 2c) could be clearly reduced by applying the trajectory correction.

In conclusion, the proposed method allows for a robust iterative correction of trajectory deviations typically affecting the image quality of radial acquisitions. The technique is not limited to radial projection imaging but can likewise be used for other non-Cartesian acquisitions where averaging is used in any manner.

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References: 1. Block KT et al. MRM 2007;57:1086-1098 2. Griswold MA et al. MRM 2005;54:1553-1556 3. Deshmane A et al. ISMRM 2012; 2542 4. Seiberlich N et al. MRM 2007;58:1257-1265

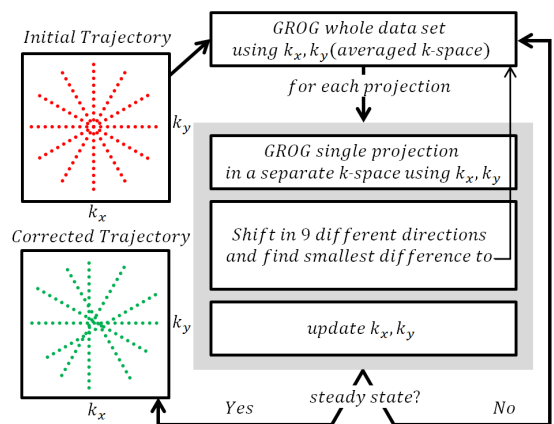


Fig. 1: Iterative correction for radial trajectories

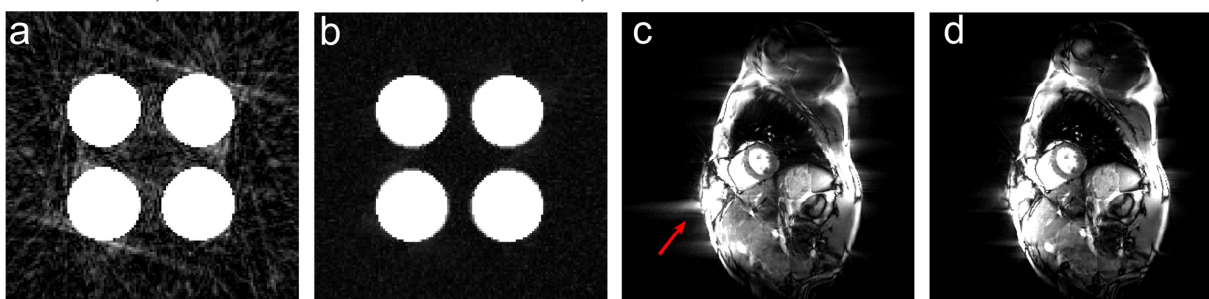


Fig. 2: Phantom images reconstructed with the corrupted (a) and corrected trajectory (b). In-vivo images reconstructed without (c) and after (d) applying the trajectory correction.