A METHOD FOR CONTINUOUS ACCELERATED ECHO-PLANAR IMAGING WITH
SELF-REFERENCED PARALLEL MR RECONSTRUCTION AND ARTIFACT CORRECTION

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ABSTRACT

Echo planar imaging (EPI) is a widely used rapid MRI technique to image dynamic neurological functionally, but can suffer from Nyquist ghosts and distortions due to magnetic field inhomogeneity. Distortion effects can be reduced by parallel MR imaging (pMRI), via a reduced echo train length, and/or methods that measure and estimate magnetic field distortions using multiple TEs. Recent Nyquist ghost correction methods can also improve self-referenced pMRI image reconstruction quality. Here, we present a combination of these improvements in a framework for continuous echo planar imaging. Through a combination of shifted variable density EPI trajectories and alternating echo spacings, our approach can remove Nyquist ghosts, apply pMRI reconstruction, and correct off-resonant distortions with no need for additional reference data. Results from accelerated in-vivo data are presented.

Index Terms—Magnetic resonance imaging, EPI Distortion Correction, EPI Nyquist Ghost Removal, Parallel MR Imaging

1. INTRODUCTION

Imaging of neurological functions requires high temporal resolution and moderate spatial resolution. Echo planar imaging (EPI) is widely used in MR imaging of the brain because of its very short image acquisition time. However, two imaging artifacts persistently degrade EPI image quality: Nyquist (or N/2) Ghosts, which are caused by the inherent mismatch between data sampled along positive and negative readout gradients during EPI acquisitions; and image distortion caused by magnetic field inhomogeneities.

Many methods to correct both Nyquist ghosts and magnetic susceptibility artifacts have been presented, notably [1][2][3][4]. More recently, Xiang and Ye [5] suggested shifted EPI trajectories to remove both Nyquist ghosts and to measure and correct distortions caused by field inhomogeneities. The EPI trajectory shifts enable one to interleave data measured on either positive or negative readout gradients, which removes Nyquist ghosts by eliminating any sampling mismatch between odd and even k-space lines. From images measured with three different trajectories, two Nyquist ghost free images can be generated with different echo spacings. A phase-difference map generated from these two images can provide a map of the magnetic field inhomogeneity, which can then be used to correct geometric distortions.

EPI image distortions can also be reduced by shortening the readout echo train. This can be achieved via segmented EPI acquisitions and/or parallel MRI imaging (pMRI) [6]. In this manuscript, we integrate parallel MRI imaging with our innovations on Xiang and Ye’s method to reconstruct a continuous stream of accelerated EPI data. Our approach removes Nyquist ghosts, performs a pMRI reconstruction, and corrects for magnetic susceptibility distortions—all with self-referenced calibration data and minimal latency. Furthermore, our method is able to maintain more than 90% of the original temporal sampling bandwidth—an improvement over previous interleaving methods which drop the temporal bandwidth by one-half. The ultimate goal of this work is to achieve high-quality continuous real-time accelerated EPI imaging, with minimal artifacts and off-resonance distortion.

2. THEORY

2.1. Nyquist Ghost Elimination

Nyquist ghosts are an inherent artifact in EPI, and there are many approaches to correct them. The goal of all Nyquist ghost removal strategies is to correct the data sampling mismatch that occurs between positive and negative EPI read-out gradients. Methods to achieve this include the acquisition of a reference scan to estimate the shift between the sampling grids [1], or by shifting the EPI sampling trajectory and then interleaving the read-out data to form two separate images that satisfy the Nyquist sampling criteria (PLACE) [5].

In [7], we proposed the use of UNFOLD [8] with the EPI trajectory shifting of the PLACE method to provide Nyquist ghost removal while maintaining nearly all of the original temporal bandwidth. As described in [5], when the images associated with positive and negative readout gradients are phase aligned and added together, any residual Nyquist ghosts
effectively cancel. Our approach is to process the positive and negative readout data separately with temporal filters, via UNFOLD, to generate images free of Nyquist ghosts at each sampled time point. This approach removes any need for reference data, yet maintains more than 90% of the original temporal resolution.

Furthermore, this approach is so effective at Nyquist ghost removal, that immediate benefits to self-referenced pMRI reconstructions of accelerated EPI data can be observed [7]. Image reconstruction quality in pMRI is dependent on high-quality calibration data[9]. The elimination of Nyquist ghosts using the UNFOLD approach improves the calibration data used in methods such as GRAPPA [10] and GEYSER [9]. Consequently, we have observed that higher acceleration factors are possible in accelerated EPI after Nyquist ghost removal using our approach.

2.2. Distortion Correction

If one considers both encoding and off-resonance effects (and ignores relaxation), the MRI signal acquisition equation for EPI can be expressed as

\[ S[k_x, k_y] = \int \int \rho(x, y) e^{i (k_x x + \frac{\Delta B(x, y)}{G_x} \tau) + i (k_y y + \frac{\Delta B(x, y) \tau}{G_y})} dx dy. \]

(1)

where \( \rho \) is the material to be imaged, \( \Delta B(x, y) \) is the magnetic field position as a function of position, and \( G_x \) and \( G_y \) are the maximum amplitude of linear magnetic field gradients used for spatial encoding and \( \tau \) is the effective pulse width of the phase-encode blip gradient. This equation reveals that the majority of the distortion caused by magnetic field inhomogeneities will occur along the phase encode direction \( y \) as the sampling time \( T \) increases.

One can remove this distortion by measuring the magnetic field inhomogeneity. This is commonly done by acquiring two images at different echo time (TE) values. With proper scaling, the phase difference between these two images is a direct measure of the magnetic field inhomogeneity. Displacement maps can then be determined from

\[ \Delta y|_{(x,y)} = \text{angle}(I_1(x,y) \circ I_2^*(x,y)) \cdot \frac{\text{FOV}}{2\pi (\delta_{TE}/T)} \]

(2)

where \( \delta_{TE} \) is the difference in echo spacing between the acquisition of \( I_1 \) and \( I_2 \), FOV is the number of pixels along \( y \) in the field-of-view, and \( T \) is the echo spacing.

3. METHODS

Our approach to implement self-referenced Nyquist ghost removal, pMRI reconstruction, and off-resonant distortion correction in a framework suitable for continuous real-time EPI imaging starts with a shifted EPI trajectory as in PLACE. On alternate frames, we delay the start of phase encoding by one echo in the echo train. This ensures that in the case of accelerated EPI imaging, the same \( k_y \) readout lines are acquired for each frame, but with an alternating readout gradient polarity. In addition, we vary the echo spacing (TE) on each pair of frames by inserting a delay between the excitation RF pulse and the readout echo train. This sampling strategy is illustrated at the top of Fig. 1 where open and closed circles indicate data sampling of opposite polarity and the TE shifts on every pair of images. This is followed by Nyquist ghost removal, pMRI reconstruction to form an image from the accelerated data, and distortion correction based on TE-shift-generated phase maps.

Next the data acquired on positive and negative readout gradients is recombined to eliminate Nyquist ghost artifacts. First, the data associated with each readout gradient is separated. Each data set is effectively sub-sampled by two, with each time frame alternating between odd and even lines. Filtering along the temporal dimension using a high-frequency narrow band-stop filter effectively removes the Nyquist-rate signal variation present along the time dimension of the data.
Our distortion correction algorithm is based on concepts from digital image warping, where pixels in the undistorted image are "pulled" from a linear combination of pixels in the distorted image. Based on the distortion map, $\Delta y$, calculated from the phase difference between images acquired at two TE values, Eq. $\Delta$ the location of the source pixel is calculated. If this source location lies between two sampled pixels, then the undistorted pixel is generated from a simple linear interpolation between the two neighboring pixels. The distortion maps will be unreliable in signal void regions, so a mask is generated from pixels greater than 1% of the maximum pixel value in the image. This mask is then expanded to the maximum coverage needed, as dictated by the distortion map. A rank-one estimate of this map is generated, calculated via the SVD, for use in regions complimentary to the mask.

4. RESULTS

Images were acquired from a healthy volunteer after informed written consent. Single-shot gradient echo EPI images were acquired at 3 acceleration rates (1x, 2x, and 3.2x) from 5 axial slices on a 1.5T MRI scanner (GE Healthcare, Milwaukee WI) with an 8 channel phased array head coil (Invivo Devices, Gainesville FL). The TE was determined automatically for each data set, resulting in decreasing TE as the acceleration factor increased (TE=49.9ms, 28.0ms, 19.9ms for R=1x, 2x, 3.2x, respectively). Imaging parameters were: matrix size = 128 (frequency) x 128 (phase); TR= 1s; Flip angle = 90°, time frames = 84; BW = 250kHz with ramp sampling; FOV = 24cm; Slice thickness = 8mm spaced 8mm apart, echo spacing = $\delta_T E = 692$ microseconds.

![Image 3](image3.png)

**Fig. 3.** Reconstruction of Slice 3 from 1x (a,b) and 3.2x (d,e) accelerated data before (left column) and after (middle column) distortion correction. (c, right column) shows a reference SPGR image for comparison.

Fig. 3 shows image reconstructions from data for the middle-most slice and two accelerations, 1x and 3.2x. (The contrast differs between the two accelerations because of the different TE values.) Images (a) and (d) show before distortion correction, and (b) and (e) show after, with a distortion free SPGR image shown twice in (c) for comparison. Notably, the accelerated data (d) shows significant reduction in
off-resonance distortion compared to the unaccelerated data in (a). The distortion correction provides further refinement, generating an image (e) more true to the SPGR reference image in (c) than the undistorted 1x image in (b).

Fig. 4 shows images at three acceleration factors from a slice near the sinuses, which are known to be prone to severe distortion and signal loss due to large magnetic field inhomogeneity. As the acceleration factor increases, there is noticeably less signal loss and less distortion. However, the high magnetic susceptibility in this slice disrupts the self-referenced coil sensitivity estimation, resulting in more visible pMRI artifacts than in Fig. 3.

**Fig. 4.** Reconstruction of slice 4 from (a) 1x, (b) 2x, (c) and 3.2x accelerated data.

**5. DISCUSSION**

We presented a method to acquire and reconstruct high-quality self-referenced images from accelerated EPI data. This method compensates for Nyquist ghosts and distortion caused by magnetic field inhomogeneity without additional reference data while maintaining 90% of the temporal bandwidth. Our self-referred pMRI reconstruction approach provides images at acceleration rates up to at least 3.2x, with high quality images available in regions with low magnetic susceptibility.

A number of technical issues remain. First, we have observed large variability in the signal phase of each acquisition frame. Our approach is able to successfully compensate for this (via UNFOLD) in the correction of Nyquist ghosts. The distortion correction and coil sensitivity estimation steps, however, shows corresponding high variability. Thus, the approach may require the inclusion of cross-image information to stabilize the output image stream. Additionally, clinical applications may need a more sophisticated interpolation method, dependent on the application requirements and real-time reconstruction constraints.

Our design was guided with the intent to implement a continuous real-time self-calibrated reconstruction system for accelerated EPI data. We are currently able to complete the most computationally demanding stage, pMRI, in ~100 msec [13] and the other steps are trivial to parallelize. Thus, we fully expect to be able complete each image reconstruction within a single TR, and will present this implementation at the meeting.

Finally, we note that all of the reconstruction algorithms described in this paper are available to other research groups through the generous support of the National Center for Image Guided Therapy. See (http://www.ncigt.org/pages/Downloads) for details.

**6. REFERENCES**


