Reduction of Geometric and Intensity Distortions in Echo-Planar Imaging Using a Multireference Scan

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Echo-planar imaging (EPI) is very sensitive to patient-induced field inhomogeneity caused by susceptibility changes between different anatomical regions. This results in geometric and intensity distortions in the image, especially near tissue/air and tissue/bone interfaces. A new approach is presented to reduce geometric and intensity distortions in EPI. A phase-encoded multireference scan is used to estimate the amplitude and phase errors in the measured signals due to the field inhomogeneity. The EPI data is corrected using both the amplitude and phase of the measured errors. This technique has been evaluated using EPI pulse sequences implemented with conventional gradients and implemented with imaging systems that have special resonating gradients and fast analog to digital converters. The results in both phantom and human studies show that in the absence of object motion the new correction technique can effectively reduce the geometric and intensity distortions.

Key words: echo-planar imaging; rapid imaging; geometric distortion; phase correction.

INTRODUCTION

Echo-planar imaging (EPI) was first proposed by Mansfield in 1977 (1) and since then variations of the original echo-planar pulse sequence have been developed (2, 3). A single-shot echo-planar pulse sequence can acquire, in less than 100 ms, all of the data points required for reconstruction of a two-dimensional (2D) image after a single radiofrequency (RF) excitation pulse. This eliminates many of the physiological motion artifacts caused by cardiac activity, blood flow, bowel peristalsis, breathing, or the voluntary movements of severely ill or uncooperative patients. Additionally, EPI is useful for obtaining a true $T_2^*$-weighted contrast (spin-echo EPI) or true $T_2^*$-weighted contrast (gradient echo EPI) independent of any $T_1$ weighting, because the repetition time can be made arbitrarily long. However, the long echo readout time required by EPI combined with large internal magnetic field inhomogeneity caused by susceptibility differences near tissue/air and tissue/bone interfaces result in large geometric and intensity distortions in single-shot EPI images. Thus, it is important to develop methods that can reduce these severe image artifacts if EPI is to be an effective diagnostic tool.

EPI is very sensitive to patient-induced field inhomogeneity. Even though the EPI imaging time is short, the EPI data sampling time of 100 ms is long compared with that of conventional imaging sequences where typically data are acquired during an 8-ms interval following each repeated RF excitation. The $T_2^*$ is quite short near tissue/air and tissue/bone interfaces because of the large susceptibility differences between these regions. The signals from regions near these interfaces dephase quickly relative to those from homogeneous tissue regions. The longer the data sampling time, the larger the differences in phase of the spins between regions of large susceptibility differences. The resulting amplitude and phase dispersions produce both geometric and intensity distortions in the EPI magnitude images. In addition, even in regions of fairly uniform magnetic field, the $T_2^*$ decay in the signal intensity during the long data sampling time in single-shot EPI results in deterioration of spatial resolution.

Various methods have been proposed to correct for geometric and intensity distortions caused by patient-induced field inhomogeneity. Weisskoff and Davis (4) and Jezzard and Balaban (5) suggested that geometric distortions in EPI could be corrected using a field map calculated from offset spin echo images. In principle, this method should correct geometric distortions provided an accurate field map can be obtained; however, it is hard to determine phase changes near tissue/air and tissue/bone interfaces because of phase wrapping that results from using the arctangent function in calculating the field map. Also, the offset spin-echo pulse sequence used to measure the field inhomogeneity is very different from an EPI pulse sequence. In fact, because of the fast readout gradient transitions in the EPI sequence, eddy currents will result, which will in turn cause additional field inhomogeneity. Thus, the magnetic field inhomogeneity, which is mapped during the field measurement sequence, will be different from the field that actually occurs during the EPI sequence. Bowtell et al. (6) proposed a correction method based on two images acquired in opposite phase encoding directions. This technique should also be able to correct geometric distortions caused by susceptibility differences. However, this technique may have blurring artifacts near the edge of the image because of $T_2^*$ decay.

In this work, we present a new method to reduce geometric and intensity distortions in single-shot EPI. Instead of measuring the field inhomogeneity, this method uses a set of reference scans, which we refer to as a multireference scan (7), to calculate phase and amplitude errors for every sample in the EPI data relative to measurements for the same phase encoding values obtained at the center of the spin echo. These measured errors are applied to the acquired EPI data to correct...
either phase only or both amplitude and phase errors. We show that this correction technique works for changes in image intensity between the multireference scan and the EPI scan. Although the multireference scan requires many RF excitations, it can then be applied to the large image data sets obtained in functional MR and contrast bolus studies. However, the technique will probably not work for complicated object motion like cardiac motion.

The multireference correction technique is evaluated using data obtained from an EPI pulse sequence with conventional gradients (EPI-CG) on the GE 1.5 T SIGNA system (GE Medical Systems, Milwaukee, WI). The EPI-CG sequence acquires data during both the flat and ramping parts of the readout gradient. Also, the new correction method was evaluated using an EPI sequence (EPI-ANMR) with the ANMR system (Advanced NMR Systems, Inc., Wilmington, MA). The EPI-ANMR pulse sequence samples the data during a fast sinusoidal switching gradient with a variable sampling rate so that the k-space is uniformly sampled. It is shown that this correction technique effectively reduces both geometric and intensity distortions.

THEORY
Geometric Distortion in EPI

In contrast with conventional MR techniques, where the acquisition of each line in k-space requires an RF excitation pulse, the EPI sequence acquires many lines of k-space in a continuous trajectory after a single RF excitation. The components of \( k_x \) and \( k_y \) are given by (8–10)

\[
\begin{align*}
  k_x(t) &= \gamma \int_0^t G_x(t') \, dt' \\
  k_y(t) &= \gamma \int_0^t G_y(t') \, dt'
\end{align*}
\]

where \( \gamma \) is the gyromagnetic ratio (2.68 \times 10^8 \text{ rad s}^{-1} \text{T}^{-1} \) for protons, and \( G_x(t) \) and \( G_y(t) \) are the time varying gradient fields in the x and y directions which are also called readout and phase encoding directions, respectively.

Using these definitions, Eq. [3] can be rewritten in terms of the signal at discrete times as

\[
S(m\Delta k_x, n\Delta k_y) = \int \int m(x, y) e^{-i\frac{\gamma B(t + m\Delta t + n\Delta B) T_2(x, y)}{T_2(x, y)}} e^{i\frac{\gamma B(t + m\Delta t + n\Delta B) T_2(x, y)}{T_2(x, y)}} dx dy
\]

where \( N_x \) is the number of data points in the readout direction, \( \Delta t \) is the sampling rate and \( \Delta k_x \) and \( \Delta k_y \) are gradient area increments in the readout and phase encod-
ing directions, respectively. Equation [6] can be rewritten as

\[ S(m\Delta k_x, n\Delta k_y) = \int \int m(x, y)\Phi(\pm m\Delta t) \cdot \Psi(nT)e^{-i\pi T/2}e^{i\pi/2(m\Delta k_x\cdot \Delta k_y) + n\Delta k_y}dxdy \]  \[ \text{[7]} \]

where

\[ \Phi(\pm m\Delta t) = e^{\pm i\phi m\Delta t} \]  \[ \text{[8]} \]

and

\[ \Psi(nT) = e^{i\phi n T} \]  \[ \text{[9]} \]

are phase terms varying with time along the readout (frequency) and phase encoding directions, respectively.

Depending on the distribution of the field inhomogeneity, both phase terms \( \Phi(\pm m\Delta t) \) and \( \Psi(nT) \) can cause phase distortions and position shifts of the signal measurements in \( k \)-space along the readout and phase encoding directions. The time shift term \( e^{\pm i\phi m\Delta t} \) causes \( k \)-space shifts in the opposite directions for even and odd echoes. The relative position shift in \( k \)-space between even and odd echoes, because of alternating readout gradients, leads to the so-called \( N/2 \) ghost. In general, the distorted phases result in geometric and intensity distortions in the image. In addition to the ghosting artifact, regions with field inhomogeneity will be misregistered in position, modulated in intensity and scaled in size [11] according to Eq. [7]. Since \( T_s \) is at least \( N_y \) times greater than \( \Delta t \), the geometric and intensity distortions are much worse in the phase encoding direction because the phase accumulation is proportional to time. Also, the \( T_s \) decay term causes a blurring artifact in echo-planar images and hence spatial resolution loss [11]. However, the geometric and intensity distortions result mainly from phase errors.

Reduction of Ghost Using a Single Reference Scan

The \( N/2 \) ghost due to the position shift in \( k \)-space \((\pm \Delta k_{sn})\) can be reduced by using a nonlinear phase correction technique based on a single reference scan [12–14]. The reference scan is performed by turning off the blipped phase encoding gradients of the EPI pulse sequence. A uniform position shift in \( k \)-space corresponds to a phase shift in the image space (Fourier-shift theorem) in the one-dimensional (1D) case. Because of the complexity of the field inhomogeneity distribution, the position shifts of \( k \)-space data are not uniform and the corresponding phase shifts in the image space are nonlinear.

The nonlinear phase errors can be evaluated by comparing each of the 1D Fourier transforms (projections) of the single reference scan echoes with respect to the 1D Fourier transform of the echo acquired near the spin-echo center based on the following equation:

\[ \Delta \phi_n(x) = \arg(\tilde{S}_n(x)/\tilde{S}_n(x)) \]  \[ \text{[10]} \]

where \( \tilde{S}_n(x) \) is the 1D Fourier transform of the \( n \)th line of the measured reference scan, \( S_n(k_x) \), and \( \tilde{S}_n(x) \) is the 1D Fourier transform of the chosen data line of the reference scan. \( S_o(k_x) \). \( S_o(k_x) \) can be acquired as close as possible to the echo time of the spin echo and will thereby suffer least from phase distortions. It is assumed in Eq. [10] that the projections are only phase distorted. This assumption is true only if the field inhomogeneities \( \Delta B(x, y) \) are small enough so that the contribution from \( \Psi(nT) \) in Eq. [7] can be ignored.

The nonlinear phase correction can be applied to the measured EPI data as follows

\[ S'(x, n\Delta k_y) = S(x, n\Delta k_y)e^{i\phi(x)} \]  \[ \text{[11]} \]

where \( S'(x, n\Delta k_y) \) is the corrected 1D Fourier transform of the \( n \)th line of the measured EPI data \( S(k_x, n\Delta k_y) \). This correction will be denoted as CORRECTION TECHNIQUE 1 hereafter. The second Fourier transform is then performed in the phase encoding direction following the nonlinear phase correction to produce the final image. This correction technique corrects the position shift of \( k \)-space data \((\pm \Delta k_{sn}) \) caused by the time shift and, thus, reduces the \( N/2 \) ghost artifact. The phase term in Eq. [11] also corrects some of the extra distortion caused by the inhomogeneity term, \( \Phi(\pm m\Delta t) \).

Reduction of Geometric and Intensity Distortions Using a Multireference Scan

The phase error \( \Psi(nT) \) term in Eq. [7] cannot be corrected using the nonlinear phase correction based on the single reference scan mentioned in the previous section. Because this term results in a linear phase shift for the \( k \)-space signal in the phase encoding direction, the slope of which is proportional to the field inhomogeneity, the resulting artifact is a direct geometric (and potentially intensity) distortion of the image due to the field inhomogeneity. Because of the position dependence of the phase error terms, Eq. [6] cannot be directly simplified. However, a relationship between the measured signal, \( S \), and the undistorted signal, \( S_o \), can be written as:

\[ S(m\Delta k_x, n\Delta k_y) = F(m, n)S_o(m\Delta k_x, n\Delta k_y) \]  \[ \text{[12]} \]

with

\[ S_o(m\Delta k_x, n\Delta k_y) = \int \int m(x, y)e^{-(m\Delta t + \pi T/2)}e^{i\pi/2(m\Delta k_x \cdot \Delta k_y + n\Delta k_y)}dxdy \]  \[ \text{[13]} \]

where \( F(m, n) \) is a complex number and \( S_o(m\Delta k_x, n\Delta k_y) \) is the signal that has the same frequency and phase encoding as \( S(m\Delta k_x, n\Delta k_y) \), but with minimal decay and phase errors. We note that if \( \Delta B \) and \( T_s \) were constant as a function of position, Eq. [12] would give an analytical solution for \( F(m, n) \). To the extent that \( \Delta B \) and \( T_s \) are slowly varying functions of position, the factor \( F(m, n) \) is a valid approximation.

The factor, \( F(m, n) \), is the relationship between the signal that would be measured at the time of the echo center and the signal obtained at its actual time of measurement. The relationship must be determined separately for each line of \( k \)-space. One way to accomplish this is to repeat the EPI pulse sequence \( N_y \) times without
Reduction of Geometric and Intensity Distortion in EPI

Phase encoding blips during the readout period. As shown in Fig. 2, during the ith reference scan, (i-1) phase encoding blips are played out before readout gradients so that all data from the ith excitation are phase encoded equally, where \( i = n + N_i/2 + 1 \). The data of the ith reference scan can be expressed as

\[
R_i(m \Delta k_x, n \Delta k_y) = \int \int m(x, y) e^{-i(T_z + m \Delta r + n \Delta k_y)/T_z} e^{i h \Delta (T \pm m \Delta t + T_E)/T_z} dx dy
\]

where \( l = -N_i/2, -N_i/2 + 1, \ldots, N_i/2 - 1 \) correspond to the first, second, \ldots, N_i\text{th} echo, and so on.

Based on Eq. [12], a filter can be formulated as

\[
F^{-1}(m, n) = R_0(m \Delta k_x, n \Delta k_y)/R_0(m \Delta k_x, n \Delta k_y)
\]

where \( F^{-1}(m, n) \) is the reciprocal of \( F(m, n) \). Once the filter \( F^{-1}(m, n) \) is evaluated for all \( m \) and \( n \), a corrected EPI data set can be obtained by

\[
S'(m \Delta k_x, n \Delta k_y) = F^{-1}(m, n)S(m \Delta k_x, n \Delta k_y)
\]

Since the corrected EPI data, \( S'(m \Delta k_x, n \Delta k_y) \), have minimal decay and phase error, the geometric and intensity distortions in the image will be greatly reduced.

The position shift between odd and even echoes in k-space results in a linear phase shift in the readout direction in the projection space (p-space) data that alternates in sign as a function of the phase encoding index. Here, the p-space is defined as the 1D Fourier transform along the readout direction of k-space (12). This position shift can be corrected by calculating the phase correction from the multireference scan in the projection space (p-space). The correction is performed using a p-space filter defined as

\[
P^{-1}(m', n) = \tilde{R}_0(m' \Delta x, n \Delta k_y)/\tilde{R}_0(m' \Delta x, n \Delta k_y)
\]

where \( \tilde{R}_0(m' \Delta x, n \Delta k_y) \) and \( \tilde{R}_0(m' \Delta x, n \Delta k_y) \) are the p-space projections of \( R_0(m \Delta k_x, n \Delta k_y) \) and \( R_0(m \Delta k_x, n \Delta k_y) \), respectively. A nonlinear phase correction can be applied to the projections of the EPI data using only the argument of \( P^{-1}(m', n) \). This step corrects the echo position shift and some of the distorted phases. The phase corrected projections are transformed back to the k-space where the filter \( F^{-1}(m, n) \) can be applied to correct additional phase distortions.

The filter described by Eq. [15] will be very sensitive to noise in regions where the magnitude of the denominator is close to zero. This spike-like behavior of the filter results in magnification of noise. The spikes can be removed, for example, by applying a median filter to \( F^{-1}(m, n) \) before the data correction.

It should be pointed out that the proposed geometric distortion correction based on a multireference scan is exact if the noise is ignored and the image intensity distribution, \( m(x, y) \), is unchanged between the reference scan and EPI scan. For a dynamically changing image intensity distribution with no object motion, this correction still should be a good approximation based on the fact that the geometric information of the image is mainly contained in the phases of the k-space signal (15).

METHODS AND MATERIALS

EPI-CG Pulse Sequence

The EPI-CG pulse sequence was implemented on a standard GE 1.5-T SIGNA system with conventional gradients under the 5X version of software. The pulse sequence is shown in Fig. 1. The data were acquired during both flat and ramping periods of the readout gradient. Figure 3 illustrates the difference between the data acquisition window of the EPI-CG pulse sequence and that of a conventional MR pulse sequence.

For all studies using the EPI-CG pulse sequence, the field of view was 32 cm and the slice thickness was 10 mm. The data sampling was acquired at a maximum frequency of \( \pm 62.5 \) kHz, or 8 \( \mu \)s per complex (real and imaginary) pair. The ramping period from zero to the maximum amplitude of 0.94 Gauss/cm or from this am-

![FIG. 2. The multireference scan sequence. (i-1) phase encoding blips are played out before readout gradients for the ith reference scan so that all data from the ith excitation are phase encoded equally.](image-url)

![FIG. 3. Difference between the data acquisition window of the EPI-CG pulse sequence and that of conventional MR pulse sequences. Conventional MR pulse sequences acquire image data only during the flat portion of each readout gradient. The proposed EPI-CG pulse sequence uses both the flat and ramping portions of the readout gradient.](image-url)
The width of the flat portion of the readout window was 520 μs (65 complex pairs). The phase encoding gradient was a half sine pulse of 160 μs with an amplitude of 0.07 Gauss/cm centered halfway along each ramping period of the readout gradient. Sixty-four echoes, each having 190 data points, were acquired for each image for a total data acquisition time of 97.28 ms. The TE was 102 ms and the TR was 1000 ms. The single-shot EPI image was obtained first followed by the acquisition of the phase encoded multireference scan.

One phantom study was performed using a cylindrical plastic bottle filled with water. The bottle was positioned with the long axis of the bottle parallel to the axis of the main magnetic field. The EPI-CG pulse sequence was used to obtain a transaxial slice orthogonal to the axis of the magnetic field. The readout direction was anterior/posterior (up/down in the image).

A second phantom study was performed using a coaxial cylindrical phantom filled with water. This phantom was positioned upright with its long axis perpendicular to the main field. The EPI-CG sequence was used to obtain a coronal slice of the coaxial cylinder, for which the S/I phase encoding direction was parallel to the main magnetic field. The readout direction was right/left.

One study was performed on a healthy volunteer. The EPI-CG sequence was used to give a transaxial slice through the cerebellum and temporal lobes. A fat saturation pulse was used to reduce chemical shift artifacts between fat and water. The image was obtained with an effective TE of 120 ms and a TR of 1000 ms.

As an example to test the effectiveness of the proposed correction technique for a changing image intensity distribution, a dynamic study using Gd-DTPA contrast agent was performed on a patient. A 10-mm transaxial slice of the brain was imaged with an effective TE of 115 ms and a TR of 1000 ms. One hundred twenty EPI-CG images were acquired. The patient received 0.1 mmol/kg of Gd-DTPA as a 10-s bolus injection in a dorsal vein of the hand, which was started 10 s after the start of the pulse sequence.

As an example to test the effectiveness of the proposed correction technique for a changing image intensity distribution, a dynamic study using Gd-DTPA contrast agent was performed on a patient. A 10-mm transaxial slice of the brain was imaged with an effective TE of 115 ms and a TR of 1000 ms. One hundred twenty EPI-CG images were acquired. The patient received 0.1 mmol/kg of Gd-DTPA as a 10-s bolus injection in a dorsal vein of the hand, which was started 10 s after the start of the pulse sequence.

EPI-ANMR Pulse Sequence

The EPI-ANMR pulse sequence used in this study is shown in Fig. 4. The high strength gradient resonator in the ANMR system was used for the frequency encoding gradient. The maximum gradient strength was 2.5 Gauss/cm and the gradient resonance frequency was fixed at 1 kHz. The data were acquired during the sinusoidal readout gradient waveform with a variable sampling rate so that k-space was sampled uniformly.

For all studies using the EPI-ANMR sequence, the field of view was 40 cm and the slice thickness was 10 mm. The TE was 90 ms and the TR was 1000 ms. The phase encoding gradient was a triangle pulse of 144 μs with an amplitude of 0.08 Gauss/cm centered at the zero point of the sine wave of the readout gradient. There were 128
Reduction of Geometric and Intensity Distortion in EPI

FIG. 6. 128 x 128 EPI-CG transaxial images of a cylindrical bottle filled with water. (a) The reconstructed image without any correction and (b) the reconstructed image with nonlinear phase correction based on the single reference scan.

FIG. 7. 128 x 128 EPI-CG coronal images of a coaxial cylinder filled with water. (a) The original image, (b) the reconstructed image using CORRECTION TECHNIQUE 1, (c) the reconstructed image using CORRECTION TECHNIQUE 2, and (d) the reconstructed image using CORRECTION TECHNIQUE 3.

K-Space Sampling

For the EPI-CG sequence, if the data sampling rate is constant in time and data acquisition occurs only during the flat portion of the readout gradient, k-space is sampled equidistantly, as shown in Fig. 5a. The data taken during the ramping periods extend the sampling region in k-space in a nonequidistant pattern as shown in Fig. 5b (shaded area). A 160-μs half sine wave is applied symmetrically halfway through the ramping of the x gradient to phase encode the data sampling for the next k_y. For the EPI-ANMR sequence, the sampling rate is variable such that the area under the readout gradient between any two adjacent samples is a constant. In other words, k-space is sampled uniformly.

Data Interpolation

Since k-space was sampled more densely during the ramping periods of the readout gradient with the EPI-CG sequence, the corresponding nonequidistant data points were interpolated to an equidistant data grid using cubic spline interpolation. Due to the higher k-space sampling density during the blip phase encoding gradient, most of the data acquired during this period is discarded without compromising image quality. Only the data value for the last k_x is used, which is the sample obtained halfway between two phase encoding rows of k_y.
FIG. 8. 128 × 128 EPI-CG axial images of a selected slice through the cerebellum/temporal lobes in the head of a healthy volunteer. (a) The original image, (b) the reconstructed image using CORRECTION TECHNIQUE 1, (c) the reconstructed image using CORRECTION TECHNIQUE 2, and (d) the reconstructed image using CORRECTION TECHNIQUE 3.

is the sample at the center of the blip. This data value is assigned to the \( k \) for the phase encoding row below. No interpolation is performed between phase encoding rows of \( k \). Then 1D cubic spline interpolation is performed along each row to obtain samples at equidistant points along the readout direction \( k \) shown in the shaded region in Fig. 5b. After interpolation of the acquired data, 128 equidistant data points were obtained for each of the 64 lines sampled in \( k \)-space.

Image Reconstruction

An off-line image reconstruction program was developed to run on SUN workstations. For both the EPI-CG sequence and the EPI-ANMR sequence, the MR signal was sampled with alternating readout gradient polarities. Therefore, an order inversion of every other data line was first required before data processing. Then three correction techniques were evaluated using EPI-CG and EPI-ANMR data.

CORRECTION TECHNIQUE 1 corrected the phase of each point in the projections based upon a single reference scan without any phase encoding. Each line of \( k \)-space of the single reference scan was first Fourier transformed to obtain all projections without phase encoding. Then the phase error of each data point in each projection line was calculated by comparing its phase to the phase of the corresponding data point in the projection through the echo center of the spin echo. These phase errors were applied to the projections (the line by line Fourier transform) of the EPI data. The corrected projections were inverse Fourier transformed to obtain the corrected EPI data (\( k \)-space phase encoded data). The primary purpose of this technique was to line up echo centers in \( k \)-space.

CORRECTION TECHNIQUE 2 corrected the phase of the EPI data based on a multireference scan instead of a single reference scan. For each \( k \)-space EPI data line (one phase encoded line), phase corrections were performed. First, the EPI data line, the corresponding \( k \)-space line of the phase encoded reference scan, and the spin-echo data line of the phase encoded reference scan were Fourier transformed. Then phase errors were calculated by comparing the phases of the two projections of the reference scan. Next, both projections of the reference and EPI scan were phase corrected. Then the corrected EPI projection and the corrected reference projection were inverse Fourier transformed. At this point all echo centers should be lined up. Additional phase errors were calculated by comparing the phase of the corrected \( k \)-space data line of the reference scan with the spin echo data line of the reference scan. Any additional errors in the previously corrected EPI \( k \)-space data line were then corrected.

CORRECTION TECHNIQUE 3 corrected the amplitude in addition to applying the phase corrections described in CORRECTION TECHNIQUE 2. After the echo centers were lined up following the procedure in CORRECTION TECHNIQUE 2, both the phase and amplitude errors were calculated simultaneously. These calculations were then applied to the previously phase corrected EPI data. In calculating amplitude correction factors in \( k \)-space based on a multireference scan, special care was taken to avoid infinite amplitude (division by zero) in any point where noise dominated. First, a search was made for the maximum amplitude of each echo that appeared in the denominator of Eq. [15] and a lower threshold of 0.01 of
Reduction of Geometric and Intensity Distortion in EPI

FIG. 9. Axial images of the patient brain acquired before (a, d, g) and 30 s after Gd-DTPA injection (b, e, h). Images are corrected using CORRECTION TECHNIQUE 1 (a, b, c), CORRECTION TECHNIQUE 2 (d, e, f), and CORRECTION TECHNIQUE 3 (g, h, i). Images (c), (i), and (j) are the differences from before and after Gd-DTPA injection.

The maximum amplitude was set. If the amplitude of any point was less than this threshold, an offset value equal to the threshold was added to the denominator to avoid spikes. In addition, a two-dimensional median filter was applied to the amplitude correction factors before the final correction.

For the EPI-CG sequence, interpolation was performed before any corrections to map the readout data to an equidistant 128 × 64 grid. CORRECTION TECHNIQUES 1, 2, and 3 were applied to the entire reordered k-space data. Since only a total of 64 echoes in $k_x$ were acquired, zero padding in the phase encoding direction ($k_y$) was performed to obtain a 128 × 128 filtered data matrix. The final 128 × 128 spin density image was obtained by performing a 2D Fourier transform on this data matrix and taking the magnitude of the Fourier transformed image.

For the EPI-ANMR sequence, interpolation was not needed because k-space was sampled uniformly. However, a baseline calibration was required because the hardware did not automatically provide it as in the EPI-CG sequence. A simple average of real and imaginary data from several high frequency k-space lines was used as the baseline values. These values were subtracted from all real and imaginary data, respectively. CORRECTION TECHNIQUE 1 was not applied to EPI-ANMR data since an echo time shift technique (16) was automatically implemented in the ANMR software to reduce the N/2 ghost. CORRECTION TECHNIQUES 2 and 3 calculated from a multireference scan were both applied to the entire reordered k-space data. Since 128 echoes were obtained, no zero padding in the phase encoding direction was applied. The final 128 × 128 spin density image was obtained by performing a 2D Fourier transform on the filtered data and taking the magnitude.

RESULTS

EPI-CG Pulse Sequence

Figure 6 demonstrates the effectiveness of CORRECTION TECHNIQUE 1 in reducing N/2 ghost in an EPI-CG image. Figure 6a shows the image of a selected axial slice of the plastic bottle filled with water. The N/2 ghost with a half field of view shift can be seen. Figure 6b shows the image corrected using the nonlinear phase correction based on a single reference scan. The ghosting artifact is effectively reduced.

Figure 7 demonstrates the effectiveness of correcting for geometric and intensity distortions in EPI-CG images of the coaxial cylindrical phantom filled with water, which was positioned with its axis perpendicular to the main field. Figure 7a shows a reconstructed coronal slice of the phantom without any correction (original image). As suggested by Coxon and Mansfield (17), the distortions are always worse in the nonaxial EPI images because of greater field inhomogeneity in these planes. In this experiment the field inhomogeneity was made even worse by placing a nonsymmetric object perpendicular to the main magnetic field. The geometric and intensity distortions are severe in Fig. 7a. Figure 7b shows the image corrected using CORRECTION TECHNIQUE 1. The N/2 ghost is reduced. However, the image is still distorted after the correction. Figure 7c shows the image using CORRECTION TECHNIQUE 2, and Figure 7d shows the image using CORRECTION TECHNIQUE 3. The geometric and intensity distortions are significantly reduced.
healthy volunteer and Fig. 8b shows the image of the same slice reconstructed using CORRECTION TECHNIQUE 1. Severe susceptibility artifacts are observed because of the existence of air/tissue and bone/tissue interfaces. The N/2 ghost is reduced with the nonlinear phase correction. Figure 8c shows the same image reconstructed using CORRECTION TECHNIQUE 2 and Figure 8d illustrate the same image reconstructed using CORRECTION TECHNIQUE 3, which are both based on a multireference scan. In both cases, geometric and intensity distortions are dramatically reduced.

Figure 9 shows results from a patient who received the contrast agent Gd-DTPA. Figures 9a and 9b are images before and 30 s after the Gd-DTPA injection both reconstructed using CORRECTION TECHNIQUE 1. Figure 9c is the subtraction of Figs. 9a and 9b showing that the image intensity decreased when the contrast agent passed through the brain. Figures 9d–9f and 9g–9i show the corresponding images reconstructed using CORRECTION TECHNIQUE 2 and CORRECTION TECHNIQUE 3, respectively. Geometric and intensity distortions can be recognized in the images in Figs. 9a and 9b. With the proposed correction technique, geometric and intensity distortions are reduced even when the image intensity distribution changes between the acquisition of the reference scan and the EPI scan (Figs. 9e and 9h). The subtraction images (Figs. 9f and 9i) show a similar change in the image intensity due to the contrast agent.

Figure 8 demonstrates the effectiveness of the correction techniques applied to images from a human study acquired using the EPI-CG pulse sequence. Figure 8a shows an axial slice through the cerebellum and temporal lobes in a EPI-ANMR Pulse Sequence

Figure 10 demonstrates the effectiveness of the correction techniques applied to images of the coaxial cylin-
DISCUSSION

A new method based on acquiring phase encoded multireference scans was developed to correct for the geometric and intensity distortions in single shot EPI. The long echo readout time coupled with magnetic field inhomogeneities with single-shot EPI results in geometric and intensity distortions in the reconstructed images. These distortions are especially severe at interfaces with large susceptibility changes. The proposed correction technique requires obtaining a separate reference scan for each phase encoding to estimate the phase and amplitude errors in the measured EPI data. The technique was compared with two other variations of the technique using echo planar images of phantoms and human volunteers. These images were acquired using conventional gradients and \( \pm 62.5 \text{ kHz} \) sampling rates and using a commercially available resonating gradient system with much faster A/D converters. For both systems the correction technique was shown to significantly reduce the image distortions.

The correction based on a single reference scan (CORRECTION TECHNIQUE 1) reduced N/2 ghosting artifacts but did not correct the geometric and intensity distortions. This is not too surprising since a single reference scan cannot measure the additional pixel-to-pixel phase differences accumulated during the phase encoding blips.

The phase only multireference scan correction (CORRECTION TECHNIQUE 2) reduces the geometric and intensity distortions but increases the noise in the image compared with the original uncorrected image and the one obtained after application of CORRECTION TECHNIQUE 1. In this technique, the phase is corrected in both \( p \)-space and \( k \)-space by multiplying the data by the complex exponential of the estimated phase error on a point by point basis. The corrected image is the result of a convolution of the uncorrected image with the inverse Fourier transform (2D for \( k \)-space correction and 1D for \( p \)-space correction) of the phase correction factors. The phase error is corrected at all points in both \( p \)-space and \( k \)-space. Points with low magnitudes will have significant uncertainty in the measured phase errors. Thus, large errors in the high frequency phase terms contribute to the noise that is spread throughout the image. These phase errors result in a significant amount of energy distributed throughout the 2D point spread function and causes a significant increase in noise in the phase corre-
rected images. To reduce the noise caused by high frequency phase errors, a test reconstruction was performed using only the central 32 lines of the phase correction matrix with the outer phase correction values effectively set to zero. In this correction, the noise external to the corrected object was significantly reduced. The distortion of the object was also significantly reduced. However, there remained a significant amount of noise and ghosting due to spread of the object in the phase encoding direction.

The phase and amplitude multireference scan correction (CORRECTION TECHNIQUE 3) has better noise properties. The improved noise performance is probably due to the care taken to avoid the effects of small denominator values in the filter in Eq. [15]. As described above, a small offset was added to the denominator where the values approached zero, and a two-dimensional median filter was used to remove any residual noise spikes. We hypothesize that the offset correction could also be used in CORRECTION TECHNIQUE 2 to remove spurious phase values and, thus, improve the resulting noise characteristics.

The current implementation of this distortion correction technique does not make good use of all the reference scan measurements. Only two out of each 64 measured echoes are used in the determination of the phase and phase/amplitude correction factors for each single line of k-space. It is likely that a more efficient use of the reference scan measurements can be developed such that these correction coefficients can be determined with a significant reduction in noise errors.

Although the discussion in this paper was limited to spin-echo EPI, the proposed multireference technique could be used in gradient-echo EPI. In gradient-echo EPI, the first echo always has the minimum amplitude and phase errors. The correction factors can be calculated by comparing all echoes to the first echo instead of the echo nearest the spin-echo center, as is done in the spin-echo EPI application. However, it is not expected that the technique could improve on any residual susceptibility distortions present in standard gradient echo imaging.

The multireference correction technique proposed in this paper is especially useful with dynamic studies. The technique requires $N_e$ excitations to acquire a multireference scan. Although the acquisition of the multireference scan is relatively long compared with the acquisition of one echo-planar image, it is not so long compared with the large number of images acquired of the same slice in dynamic studies involving the use of contrast agents and in functional MRI studies. To reduce motion sensitivity during acquisition of the multireference scan, it might be desirable in the future to implement an additional navigator echo in addition to the multireference scan.

The correction technique presented in this paper will be affected by patient motion. However, to the extent that patient motion is eliminated, we have found that the proposed correction technique can adequately correct for susceptibility induced geometric and intensity distortions and that the correction seems to work when applied to the images in dynamic sequences where the image intensity is changing due to contrast bolus passage or activation studies. We believe that the corrected images provide more accurate anatomical information that is very important for clinical diagnosis. As the demand increases for the application of diffusion and perfusion imaging, it becomes more and more important to solve the image quality issues that are present with echo-planar imaging.

REFERENCES