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Centering the Projection Reconstruction Trajectory: Reducing Gradient Delay Errors

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Abstract

The projection reconstruction (PR) trajectory was investigated for the effect of gradient timing delays between the actual and requested start time of each physical gradient. Radial trajectories constructed with delayed gradients miss the center of k -space in an angularly dependent manner, causing effective echo times to vary with projection angle. The gradient timing delays were measured in phantoms, revealing delays on the x, y, and z gradients which differed by as much as 5 μ sec. Using this one-time calibration measurement, the trajectories were corrected for gradient delays by addition of compensatory gradient areas to the prephasers of the logical x and y readout gradients. Effective projection-to-projection echo time variability was reduced to less than 1 μ sec for all imaging orientations. Using corrected trajectories, artifacts were reduced in phantom images and in volunteer studies. This correction should potentiate greater clinical use of the PR trajectory.

Keywords

projection reconstruction; radial imaging; gradient errors; gradient delay; compensation blips; reference scans; echo-planar imaging; magnitude back projection; linear phase

Projection reconstruction (PR) is recently a subject of renewed investigation as a valuable MR imaging trajectory. The sensitivity of PR to off-resonance and B^0 field inhomogeneity described early on (1) has been reduced with more homogeneous magnets and by imaging with high receiver bandwidths, coupled with increased gradient strengths. Gridding reconstruction methods are faster now due to improved computer hardware and the implementations are well known (2). A still serious source of image artifact in images obtained with the PR trajectory arises from miscentered k -space trajectories, to which Cartesian imaging is somewhat immune. For PR, radial lines of k -space are acquired as projections (N_p) consisting of N_r readout points. For a miscentered trajectory, the positions of the acquired data are different from the intended positions. Gridding these offset k -space points onto the locations of the intended trajectory results in a reconstruction error.

This investigation presents a new solution to the k -space miscentering artifact. The solution follows another proposed for echo-planar imaging (3). Relative delays between the requested and the actual start times of the x, y, and z gradients, shown in Fig. 1 with delays exaggerated for inspection purposes, are assumed to be a cause of k -space miscentering. The effect of different gradient delays on each physical gradient is a miscentering of k -space which varies for each projection angle. First, the relative gradient delays are measured once for a particular scanner, using a homogeneous phantom. This one-time calibration permits calculation of compensatory gradient areas needed to correct these delays for any slice plane rotation. These compensatory areas are added to the gradient readout prephasers and rephasers.

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Figure 2 shows simulated PR trajectories that result from gradient delays. The trajectories for 16 projections are plotted from 0–180°. Only the central region of k -space is shown: $(k_x, k_y) = (\pm 1, \pm 1)$ out of a 160×160 matrix. The solid dots show the position in k -space at the intended echo time. Typical delays of $t_x = -3.7 \mu\text{s}$, $t_y = +1.5 \mu\text{s}$, a 160×160 pixel field of view, and an $8 \mu\text{s}$ dwell time were used to construct the plot. The trajectories do not align at $kr = 0$. This results in an inaccurate reconstruction and reduced image quality.

THEORY

Miscentering of the PR k -Space Trajectory Due to Gradient Delays: Axial Case

There exist delay times, the origins of which were not determined, between the actual and requested start times of a gradient waveform. These delays are slightly different for the physical x , y , and z gradients used by the scanner. The 2D projection reconstruction pulse sequence employed with steady state free precession (SSFP) contrast is shown in Fig. 1. The gradients are shown for an axial imaging plane, with exaggerated delays, t_x and t_y , and t_z on each physical gradient axis. The delays cause the time integral of the gradient between the effective RF center and the intended echo time to be different than expected. Figure 3 shows the trajectory for a projection at angle θ . For an axial acquisition, the effective error in gradient area (ΔA), and therefore k -space error, is:

$$\begin{aligned}\Delta A_x(\theta) &= t_x \cdot G_{\text{read}} \cdot \cos(\theta) \\ \Delta A_y(\theta) &= t_y \cdot G_{\text{read}} \cdot \sin(\theta) \\ \Delta A_z &= -t_z \cdot G_{\text{slice}} \\ \Delta \mathbf{k} &= \frac{\gamma}{2\pi} \Delta \mathbf{A}(\theta) = i\Delta k_x + j\Delta k_y + k\Delta k_z\end{aligned}\quad [1]$$

where $G_{\text{read}} \sin(\theta)$ and $G_{\text{read}} \cos(\theta)$ are the x and y readout gradient strengths for the PR trajectory; t_x , t_y , and t_z are the gradient delay times on the x , y , and z axis, defined in Fig. 1; and Δk_x and Δk_y are defined in Fig. 3. γ is the gyromagnetic ratio for protons: $\gamma/2\pi = 4258 \text{ Hz/G}$. Equation [1] states that the gradient area errors are equal to the shaded regions shown in Fig. 1. This error results in traversal of the effective center of k -space at a k -space position shifted by (see Fig. 3):

$$\begin{aligned}\Delta k_{\parallel}(\theta) &= \Delta \mathbf{k} \cdot (\mathbf{i} \cdot \cos(\theta) + \mathbf{j} \cdot \sin(\theta)) \\ &= G_{\text{read}} \frac{\gamma}{2\pi} [t_x \cdot \cos^2(\theta) + t_y \cdot \sin^2(\theta)] \\ \Delta k_{\perp}(\theta) &= \Delta \mathbf{k} \cdot (\mathbf{i} \cdot -\sin(\theta) + \mathbf{j} \cdot \cos(\theta)) \\ &= G_{\text{read}} \frac{\gamma}{2\pi} [-t_x \cdot \sin(\theta) \cdot \cos(\theta) + t_y \cdot \sin(\theta) \cdot \cos(\theta)]\end{aligned}\quad [2]$$

where $\Delta k_{\parallel}$ is the portion of the error, $\Delta \mathbf{k}$, in the readout direction (see Fig. 3), and $\Delta k_{\perp}$ is the distance of closest approach of the trajectory to the center of k -space. The entire signal from the gradient echo is shifted from its intended position in k -space by the amount $\Delta k_{\parallel}$. Δk_z causes through-plane slice dephasing and is not a source of trajectory error for axial imaging. Below it will be demonstrated that G_{slice} is important when imaging oblique slices.

Miscentering for Oblique Imaging Planes

Using the procedure of Reeder et al. (3), the effect of and correction for gradient mistimings on oblique imaging planes can be obtained. The logical gradients (i.e., expressed in the scan plane coordinate system) requested are represented as $\mathbf{G}_{\text{log}} = [G_1, G_2, G_3] = [G_{\text{read}}(t) \cos(\theta), G_{\text{read}}(t) \sin(\theta), G_{\text{slice}}(t)]$. G_1 and G_2 are described by cosine and sine modulation with projection angle for the PR trajectory and G_3 is the slice select gradient, which must be included as a

source of delays. The rotation matrix, R , which transforms the logical gradient $\mathbf{G}_{\text{log}}$ set to a physical gradient set ($\mathbf{G}_{\text{phys}} = [G_x, G_y, G_z]$) is:

$$\mathbf{G}_{\text{phys}}(t) = \begin{bmatrix} G_x(t) \\ G_y(t) \\ G_z(t) \end{bmatrix} = \begin{bmatrix} R_{11} & R_{12} & R_{13} \\ R_{21} & R_{22} & R_{23} \\ R_{31} & R_{32} & R_{33} \end{bmatrix} \begin{bmatrix} G_{\text{read}}(t) \cdot \cos(\theta) \\ G_{\text{read}}(t) \cdot \sin(\theta) \\ G_{\text{slice}}(t) \end{bmatrix}. \quad [3]$$

Here R_{11} is the portion of G_1 which plays out on the physical x axis, R_{21} is the portion of G_1 which plays out on the physical y axis, etc.

It is these gradients, $\mathbf{G}_{\text{phys}}$, and not the logical gradients, that are delayed, where T is the time delay operator:

$$T[\mathbf{G}_{\text{phys}}] = \begin{bmatrix} G_x(t - t_x) \\ G_y(t - t_y) \\ G_z(t - t_z) \end{bmatrix}. \quad [4]$$

The resultant gradients played out on the logical axes are related to these physical gradients by R^T , the transpose of the rotation matrix (3):

$$\mathbf{G}_{\text{log}}^{\text{del}} = R^T T[\mathbf{R} \mathbf{G}_{\text{log}}]. \quad [5]$$

Finally, the error in gradient area is calculated as :

$$\Delta \mathbf{k}(\tau) = \frac{\gamma}{2\pi} \int_0^\tau \mathbf{G}_{\text{log}}^{\text{del}} dt - \frac{\gamma}{2\pi} \int_0^\tau \mathbf{G}_{\text{log}} dt. \quad [6]$$

Referring to Fig. 1, this $\Delta \mathbf{k}(\tau)$ is equal to the product of $(\gamma/2\pi)$ and the area shown as shaded regions, which indicate gradient area error. Eq. [6] simplifies to:

$$\Delta \mathbf{k}(\tau) = \frac{\gamma}{2\pi} R^T \begin{bmatrix} t_x \cdot (G_x(\tau) - G_x(0)) \\ t_y \cdot (G_y(\tau) - G_y(0)) \\ t_z \cdot (G_z(\tau) - G_z(0)) \end{bmatrix} \quad [7]$$

where $t = 0$ is defined as the effective center of the RF pulse, when only the logical slice gradient is nonzero, and $t = \tau$ is the time from the center of the RF to the center of the readout window (echo time), when only the readout gradients are nonzero (see Fig. 1). Equation [7] can be evaluated explicitly using matrix multiplication, employing Eq. [3] to relate physical and logical gradients. The result provides the errors in the logical gradient areas due to gradient delays (for each projection angle θ):

$$\begin{aligned}
\Delta k_1(\theta) &= G_{\text{read}} \cdot \frac{\gamma}{2\pi} \cdot \left(t_x \cdot R_{11}^2 \cdot \cos(\theta) + t_y \cdot R_{21}^2 \cdot \cos(\theta) + t_z \cdot R_{31}^2 \cdot \cos(\theta) + \right. \\
&\quad \left. t_x \cdot R_{11} \cdot R_{12} \cdot \sin(\theta) + t_y \cdot R_{22} \cdot R_{21} \cdot \sin(\theta) + t_z \cdot R_{32} \cdot R_{31} \cdot \sin(\theta) \right) \\
&\quad - G_{\text{slice}} \cdot \gamma \left(t_x \cdot R_{11} \cdot R_{13} + t_y \cdot R_{21} \cdot R_{23} + t_z \cdot R_{31} \cdot R_{33} \right) \\
\Delta k_2(\theta) &= G_{\text{read}} \cdot \frac{\gamma}{2\pi} \cdot \left(t_x \cdot R_{12}^2 \cdot \sin(\theta) + t_y \cdot R_{22}^2 \cdot \sin(\theta) + t_z \cdot R_{32}^2 \cdot \cos(\theta) + \right. \\
&\quad \left. t_x \cdot R_{11} \cdot R_{12} \cdot \cos(\theta) + t_y \cdot R_{22} \cdot R_{21} \cdot \cos(\theta) + t_z \cdot R_{32} \cdot R_{31} \cdot \cos(\theta) \right) \\
&\quad - G_{\text{slice}} \cdot \gamma \left(t_x \cdot R_{12} \cdot R_{13} + t_y \cdot R_{22} \cdot R_{23} + t_z \cdot R_{32} \cdot R_{33} \right) \\
\Delta k_3(\theta) &= G_{\text{read}} \cdot \frac{\gamma}{2\pi} \cdot \left(t_x \cdot R_{11}^2 \cdot \cos(\theta) + t_y \cdot R_{21}^2 \cdot \cos(\theta) + t_z \cdot R_{31}^2 \cdot \cos(\theta) + \right. \\
&\quad \left. t_x \cdot R_{11} \cdot R_{12} \cdot \sin(\theta) + t_y \cdot R_{22} \cdot R_{21} \cdot \sin(\theta) + t_z \cdot R_{32} \cdot R_{31} \cdot \sin(\theta) \right) \\
&\quad - G_{\text{slice}} \cdot \gamma \left(t_x \cdot R_{13}^2 + t_y \cdot R_{23}^2 + t_z \cdot R_{33}^2 \right)
\end{aligned} \tag{8}$$

Δk_1 , Δk_2 , and Δk_3 are the logical k -space errors in the readout (1,2), and slice-encoding (3) directions. G_{read} is the magnitude gradient strength of the readout gradient; G_{slice} is the magnitude of the slice select gradient. These logical k -space offsets must be subtracted from the k -space trajectory, by correcting the prephaser gradient areas of logical gradients. Note that Eq. [8] indicates that the slice gradient causes in-plane trajectory errors for oblique planes only.

Note that Δk_1 and Δk_2 indicate an error in the PR trajectory. The offset Δk_3 indicates phase across the slice direction, causing signal modulation which varies with projection angle, but does not disturb the in-plane k -space trajectory. The amplitude modulation is very small, with an amplitude of roughly 0.034 cycles/cm ($\Delta k_3 = 2\text{G/cm} \cdot 4258\text{ Hz/G} \cdot 4\text{ }\mu\text{s}$), or 0.034 cycles for a typical 10 mm slice. Slice dephasing from Δk_3 is small, so the correction of Δk_3 was not employed.

MATERIALS AND METHODS

Data Acquisition and Reconstruction

Imaging was performed on a General Electric (Waukesha, WI) CV/i 1.5 T system, equipped with CRM (cardiac resonance module) gradient coils and SGD (scaleable gradient design) gradient amplifiers rated at a slew rate of 150 mT/m/s and 4 G/cm maximum gradient strength. A commercial fast gradient echo sequence was modified to collect projections and provide gradient echo or steady state free precession contrast (SSFP) contrast with short TRs. Full echoes were collected at evenly spaced projections over 180°. Initial pulse sequence development included manual adjustment of the data acquisition window in time to provide the best visual image quality in phantoms. Typical scan parameters were: 160 Nr $\times$ 128 Np, 24 cm FOV, 8 mm slice thickness, flip angle 60°, $\pm 62.5\text{ kHz}$ (8 $\mu\text{s/sample}$) receiver bandwidth, TR = 3.7 ms. The head coil was used for phantom experiments. Volunteer scans were performed with the head coil and torso and cardiac phased array coils. Phantoms and human volunteers were imaged, before and after gradient correction by addition of compensation gradient area to the pre- and rephasers. Volunteers provided informed consent in accordance with our institutional review board. Gridding was performed using a triangular convolution function of width 2 pixels (2). The k -space projections were filtered as described by Joseph (4). Fourier transforms were performed using FFTW software (www.fftw.org). Because the goal was to produce a k -space trajectory that was centered, it was essential to assign the k -space data points to locations for a centered projection, i.e., $k_r = -128, \dots, -1, 0, 1, \dots, 127$.

Gradient Delay Measurements

Delays were measured by imaging a 4-inch diameter spherical CuSO_4 phantom in a standard MR quadrature birdcage coil (head coil), placed at isocenter, using the SSFP scan parameters

above, and many slice orientations. SSFP is valuable because of the recently reported insensitivity of the *phase* of the collected SSFP signal to off-resonance (5): i.e., little linear phase will be present on each projection due to shim imperfections. The SNR of the phantom images was about 130, estimated as the mean signal in an ROI within the magnitude image of the phantom, with noise estimated as the standard deviation in air-space, multiplied by 1.52 (6). Each echo from the raw k -space data (i.e., each projection) was Fourier-transformed and then processed using the algorithm described by Ahn and Cho (7). This algorithm provides the shift ($\Delta k_{||}$) of the center of k -space for each projection by measuring the linear phase angle of the image space projections; in practice, any algorithm which provides the k -space shift can be employed. These are related by the well-known Fourier shift theorem:

$$DFT[S(k - \Delta k_{||})] = DFT[S(k)] \cdot \exp\left(-i \frac{2\pi \Delta k_{||} \cdot x}{N}\right) \quad [9]$$

where DFT is the discrete Fourier transform and FOV is the field of view. The Ahn algorithm provides an estimate of the angle $(2\pi \Delta k_{||} \cdot x)/N$, thereby providing $\Delta k_{||}$ for each projection. This reflects gradient mistiming if there are no other sources of linear phase shift due to the object or the imaging procedure. The k -space pixel shift, $\Delta k_{||}$, can be converted to delay time:

$$t_d = (FOV \cdot \Delta k_{||}) / 2 \cdot BW. \quad [10]$$

where $2 \cdot BW$ is the receiver bandwidth of acquisition (inverse dwell time). Combining Eq. [2] and Eq. [10], and using the identity $2 \cdot BW = (\gamma/2\pi) \cdot G_{read} \cdot FOV$, provides the dependence of measured delays on projection angle (θ):

$$t_d(\theta) = [t_1 \cdot \cos^2(\theta) + t_2 \cdot \sin^2(\theta)] \quad [11]$$

where t_1 and t_2 are delays on the x, y, or z axis, depending on the imaging plane. The delays were measured for each projection over 180° . Then the delay vs. θ curve was fit to Eq. [11]. These fits provided t_x , t_y , and t_z . In summary, physical gradient delays are measured by acquisition of SSFP data with frequency encoding direction in all three physical gradient directions, using a homogeneous phantom and processing the raw data to determine the k -space position shifts of the echoes and interpreting the shifts as delays using Eq. [10]. Ideally, these delays values could be obtained very precisely by the scanner manufacturer during scanner installation.

Gradient Delay Correction

The error in area, ΔA , is related to k -space position offsets, Δk , using the gyromagnetic ratio: $\Delta k = (\gamma/2\pi)\Delta A$. The logical readout prephaser and rephaser areas were corrected by addition of delay compensatory areas, given by Eq. [8]:

$$A_{1,2}^{corr}(\theta) = A_{1,2}(\theta) \pm \frac{\Delta k_{1,2}(\theta)}{\gamma / 2\pi} \quad [12]$$

where $A(\theta)$ is the area of the prephaser (or rephaser) before scaling. $\Delta k_{1,2}(\theta)$ is given by Eq. [8]. The + corresponds to the correction for the rephasers and – for the prephasers in Eq. [12]. The adjusted amplitudes for each projection angle were placed in a look-up table for use during the scan. Images were analyzed visually and quantitatively by measuring the maximum artifact signal intensity as a percent of the object signal intensity on phantom images, using a large ROI in the phantom, and an ROI selecting the entire background which contained artifacts.

RESULTS

Figure 4 shows the delays measured for each projection angle around 180° for coronal, axial, sagittal, and oblique and doubly oblique views, with and without correction, smoothed by a two-point mean filter. The least-squares fits of Eq. [11] to the measurements of Fig. 4a-c

provided delay times of $t_x = -1.7 \mu\text{s}$, $t_y = +1.35 \mu\text{s}$, and $t_z = -3.7 \mu\text{s}$. Note that the gradient timing delays for each physical gradient can be estimated by direct inspection of Fig. 4a-c. The delays measured from four very different oblique slices with and without correction are plotted in Fig. 4d. For oblique imaging the shape of the delay vs. θ curve was asymmetric. For this case, the process of slice selective excitation leaves the spins at an offset position in the logical readout plane, which is independent of projection angle (see Eq. [8]). The delays measured in Fig. 4d reflect the component of this offset which is in the projection direction for each projection angle. Using the correction of Eq. [12], the delays were almost completely removed, as shown in Fig. 4, demonstrating that the general framework of Eq. [12] is successful for removing delays for an arbitrary obliquity.

Figure 5 shows measurements of delays with and without correction for an oblique plane using a wide range of scan parameters. The delay correction method was effective (within $\pm 0.7 \mu\text{s}$) with change in FOV (24→32 cm), receiver bandwidth ($\pm 62.5\text{kHz}$ → $\pm 125\text{kHz}$), slice-thickness (8→12 mm, even for oblique scans), readout resolution (160→256), and using gradient echo instead of SSFP contrast, employing only a single calibration of physical gradient delays. The delays at $\pm 32 \text{ kHz}$ (not shown) were only partially removed using the correction method. This is less consequential because at $\pm 32 \text{ kHz}$ each frequency encoding requires $16 \mu\text{s}$ per sample, so that even large delays (e.g., $4 \mu\text{s}$) result in trajectory changes by only a small fraction of a k -space pixel. At receiver bandwidths of $\pm 32 \text{ kHz}$, shim and off-resonance play a larger role in trajectory errors than gradient delays. Delays were measured on a 1.5 T Siemens Sonata scanner and were found to have similar variation and magnitude.

Figure 6 shows a sphere imaged using coronal, axial, and sagittal planes, and also the doubly oblique plane before and after gradient area correction. The doubly oblique plane is the same as Rotation 4 of Fig. 4d, described by the rotation matrix: $[R_{11}, R_{12}, R_{13}, R_{21}, R_{22}, R_{23}, R_{31}, R_{32}, R_{33}] = [0.0, 0.610, 0.792, 0.825, -0.447, 0.345, 0.565, 0.652, -0.504]$. The improvement due to correction for gradient delays is evident. The maximum artifact level as a percent of the object signal are given, without (average 24%) and with the correction (average 4%).

Figure 7a,b shows a sagittal slice of the head, using 2D spoiled gradient echo and $\text{TR}/\theta = 100 \text{ ms}/40^\circ$, $\pm 62.5 \text{ kHz}$ bandwidth, $160 \times 160 \text{ Np}$, without and with correction. The improvement is similar to improvement in phantoms.

DISCUSSION

This investigation provided a successful analytical method of reducing the trajectory errors for any slice orientation and any imaged object using a single set of calibration measurements of physical gradient timing delays. Although gradient hardware improvements to remove these gradient delays would be valuable, the use of compensatory gradient areas provides a precise and simple correction method. Compensatory gradient areas are valuable, since employment of software correction of timing delays on the physical gradients themselves is not possible for any scanner with gradients that are software-programmable with a temporal resolution coarser than roughly $1 \mu\text{s}$. Furthermore the actual physical gradients, where these timing delays must be placed, are not easily accessible (3).

The present method differs from that used to correct Cartesian multiecho trajectories in (3) because it incorporated sources of trajectory errors arising from delayed *slice select gradients*. For oblique scans generally, gradient delays during slice selection result in an excitation that leaves the magnetization with an in-plane offset. This additional correction term identified in the present investigation will be valuable when applied to the Cartesian multiecho trajectories and oblique planes.

Other methods have been employed for reducing the effect of timing delays. A recently proposed solution to PR trajectory errors involves measurement of trajectories followed by accurate gridding (8). Low receiver bandwidths help, since a given gradient delay results in larger k -space trajectory errors with higher bandwidths than with lower bandwidths, due to the larger gradient amplitudes employed. For example, a 4 μ s delay corresponds to a k -space sample point shift of 0.5, using ± 62.5 kHz receiver bandwidth, compared to a sample shift of 1.0 using ± 125 kHz. A commonly used solution is magnitude back projection (9), which removes all phase (including linear) from the image space projections, although it is known to provide an inaccurate reconstruction in some situations. Another involves first-order phase-correction of each projection, which has some of the challenges of magnitude back projection, unless judiciously applied (10).

Start times of the physical gradients differ by as much as 5 μ s and correction for these relative delays results in qualitatively and quantitatively improved images for *any* slice plane rotation (Fig. 4) and a wide range of scan parameters (Fig. 5). The delays need to be measured only once for a particular gradient set. Gradient delays have been shown in this investigation to be an important cause of miscentering and the image quality improvement after correction is visible in high SNR situations. While this investigation highlights the tolerance and resilience of the PR acquisition to k -space trajectory errors, the correction potentiates greater clinical use of this trajectory by reducing one of the most important sources of trajectory error.

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FIG 1.

Logical x, y, and z gradients for a PR SSFP acquisition are delayed with respect to each other. This results in a center of the data acquisition window that does not coincide in time with the centers of the gradient echoes. The shaded area shows the extra gradient area that has accumulated at the intended echo time, which is equivalent to a k -space shift.

FIG 2.

Result of a simulation of the trajectory for 16 projections in the presence of gradient delays of 3.7 and $-1.5 \mu\text{s}$, and an $8 \mu\text{s}$ per sample dwell-time. Only the central region, $(k_x, k_y) = (\pm 1, \pm 1)$, out of a 160×160 matrix, is shown. The circles indicate the position on the trajectories at the intended echo time for each projection. Note that the effective echo time, the point of closest approach to $(k_x, k_y) = (0, 0)$, varies with projection angle, and that the k -space trajectory is miscentered.

FIG 3.

The effect of gradient delays on projection at angle θ . This shows the intended and the actual k -space trajectory of a nonoblique slice for a very large delay. The two points (●) represent the k -space position at time of the center of the data acquisition window for the intended $(k_x, k_y) = (0,0)$ and actual $(k_x, k_y) = (\Delta k_x, \Delta k_y)$ trajectory. This position difference can be resolved into a portion which is parallel to the intended trajectory, given by $\Delta k_{||}$, and portion perpendicular, $\Delta k_{\perp}$. $\Delta k_{\perp}$ is the distance of closest approach to $(0,0)$ of the actual trajectory. $\Delta k_{||}$ is significant, because it is the shift in each collected gradient echo which can be measured in each projection and can be directly related to gradient delays.

FIG 4.

Results of measurement in a phantom of the effective gradient delays t_x , t_y , and t_z on the x, y, and z axis, using axial, coronal, sagittal, and doubly oblique imaging planes. **a–c:** Measured delays (●) smoothed with a two-point window and the best-fit curve to these raw data are shown. After correction, the delays are reduced greatly (solid lines). Note that the gradient timing delays for each physical gradient can be estimated by direct inspection of these plots. For example, the axial acquisition (**a**) uses only the physical RL (right–left or y) gradient at $\theta = 0^\circ$. The echo time delay at $\theta = 0^\circ$ is $\sim 1.2 \mu\text{s}$. This indicates the the y gradient starts $1.2 \mu\text{s}$ later than intended (see Fig. 1). **d:** Four very different oblique scan plane rotations without (solid lines) and with correction.

Measured Delays

FIG 5.

Measurement of delays for an oblique scan without and with correction over a range of scan parameters. Standard (std) scan parameters were: 24 cm FOV, 160 Nr, 128 Np, ± 62.5 kHz receiver bandwidth, SSFP, 8 mm slice. In each additional scan a single parameter such as readout resolution (256 Nr), receiver bandwidth (± 125 kHz), slice thickness (12 mm slice), FOV (32 cm), or gradient echo contrast (GRE), was varied.

FIG 6. Images of the sphere, showing axial (or the x-y plane), sagittal (y-z plane), coronal (x-z plane), and doubly oblique views before (a) and after correction (b). Images visually improved in all cases. The percent maximum artifact signal is provided for each image.

FIG 7.
2D T_1 -weighted sagittal images of the head, without (**a**) and with (**b**) correction. Improvement is evident everywhere.