Removal of Intravoxel Dephasing Artifact in Gradient-Echo Images Using a Field-Map Based RF Refocusing Technique

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A technique is proposed to compensate for the slice dephasing artifact and improve the signal-to-noise ratio (SNR) of gradient-echo images. This method is composed of two components: mapping of the internal gradient and design of the slice-selective radiofrequency (RF) pulse. The RF pulse is designed with its phase response as the negative of the product of a chosen echo time and the intravoxel internal gradient profile in a specified region of interest (ROI). The designed RF pulse can refocus the spin phases at a selected echo time and therefore effectively recover the signal loss due to both linear and nonlinear internal gradients. Principles, implementation, and application of the method are described in this note. Magn Reson Med 42:807 – 812, 1999. © 1999 Wiley-Liss, Inc.

Key words: gradient-echo image; magnetic field inhomogeneity; slice dephasing artifact; RF refocusing technique

Spin dephasing arising from intravoxel internal gradient lowers the signal-to-noise ratio (SNR) and therefore degrades the quality of gradient-echo images, especially those acquired at long TE. This dephasing effect can be lessened with increased in-plane spatial resolution and reduced slice thickness, as suggested by Young et al. (1). However, slice thickness reduction is not always a suitable solution for image quality improvement, since it also reduces the SNR. The in-plane pixel size is therefore usually much smaller than the slice thickness in a conventional two-dimensional (2D) gradient-echo image. As a result, the intravoxel dephasing effect is mainly along the slice selection direction and is termed the slice dephasing artifact. Shimming can generally reduce the magnetic field inhomogeneity. However, the slice dephasing artifact in regions with strong susceptibility effect is difficult to compensate for by shimming alone. Several methods have been developed recently to recover the signal loss due to dephasing. These include incrementing of the slice refocusing gradient (2), application of a tailored radiofrequency (RF) pulse with a quadratic phase profile (3), use of the projection mapping technique. A slice-selective RF pulse is designed through the use of an RF pulse built on prior knowledge of the internal gradient. Based on this premise, a new algorithm, termed field-map-based RF refocusing, has been developed and is presented here. The proposed technique requires two steps: mapping of the internal gradient and design of the RF pulse. The intravoxel internal gradient within the imaged slice is first measured using the field mapping technique. A slice-selective RF pulse is designed with its phase response equal to the negative of the product of TE and the slice-direction internal gradient profile of a selected region of interest (ROI). The phase distribution along the slice-selection direction is modulated, and the magnetization vectors are refocused at a selected echo time. The SNR in the ROI can therefore be optimized at a pre-selected echo time.

THEORY AND METHODS

The phase accumulation θ within an image voxel (x₀, y₀, z₀) can be represented as a function of time and z (slice-selection direction), as shown in Eq. [1], since the slice thickness is usually much larger than the in-plane pixel size. The voxel signal decays with time due to both T₂ decay and intravoxel phase incoherence, as indicated by Eq. [2].

\[ \theta_{x_0,y_0,z_0}(z, t) = 2\pi \Delta \omega_{x_0,y_0,z_0}(z) t \]

\[ z_0 - ST/2 < z < z_0 + ST/2 \]  

\[ S_{x_0,y_0,z_0}(t) = S_{x_0,y_0,z_0}(0) \exp \left( -t/T_2 \right) \exp \left[ i \theta_{x_0,y_0,z_0}(z, t) \right] \]  

where Δω_{x_0,y_0,z_0}(z) is the Larmor frequency as a function of z, ST is the slice thickness, and S_{x_0,y_0,z_0}(0) and S_{x_0,y_0,z_0}(t) represent the voxel signal intensity at time 0 and t, respectively.

In gradient-based compensation methods (2,5–7), a linear gradient along the z-direction is added to compensate for the internal gradient. Without utilizing prior kno-
edge of the internal gradient, multiple acquisitions (e.g., 16 steps) with different compensation amplitudes \( (G_m) \) and duration \( (\tau) \) are needed. The voxel signal intensity that corresponds to multiple compensation steps is presented in Eq. [3]. The compensated voxel signal can then be calculated by taking either the maximum value (6), the average value (2), or the Fourier transform (5) of the multiple acquisitions.

\[
S_{x_0,y_0,z_0}^m(t) = S_{x_0,y_0,z_0}(0) \exp \left(-\frac{t}{T_2} \right) \times \sum z \exp \left[i(\Delta \omega_{xo,yo,zo}(z)t + 2\pi G_m z \tau)\right]
\]

\[z_o - ST/2 < z < z_o + ST/2\]

\[1 < m < 16 \]  \[\text{[3]}\]

In many in vivo studies, the nonlinear component of the internal gradient cannot be ignored (e.g., strong susceptibility effect near tissue/air interfaces). In this case, the signal loss due to the higher order internal gradient cannot be efficiently recovered by the gradient-based compensation methods, as indicated by Eq. [3].

In the proposed RF refocusing technique, an RF pulse with a specified phase response profile is designed and used for slice-selection excitation. Utilizing prior knowledge of the internal gradient, the RF pulse is designed with its phase response \( (\Phi) \) as the negative of the predicted phase evolution. The dispersed phases due to both linear and nonlinear internal gradients can therefore be refocused at a pre-selected TE, as indicated by Eq. [4].

\[
S_{x_0,y_0,z_0}(t) = S_{x_0,y_0,z_0}(0) \exp \left(-\frac{t}{T_2} \right) \times \sum z \exp \left[i(\Delta \omega_{xo,yo,zo}(z)t + \Phi_{TE}(z))\right]
\]

\[
\Phi_{TE}(z) = -\Delta \omega_{xo,yo,zo}(z)TE \]  \[\text{[4]}\]

Phantom and animal experimental studies were performed on a GE Omega 4.7 T imager fitted with actively shielded gradient coils, capable of generating the maximum amplitude of 18 Gauss/cm. The phantom consisted of an air-filled glass sphere (2 mm radius) embedded within 5% gelatin, as shown in Figure 1a. The parameters of single-slice x-y-plane gradient-echo images included slice thickness 4 mm, field of view (FOV) 51.2 x 51.2 mm, TR 200 msec, flip angle 30°, and matrix size 256 x 256. Using a birdcage RF coil, 12 gradient-echo images were acquired at different TE values (10–21 msec) with a sinc RF pulse. A spin-echo reference image was acquired with TR 500 msec, TE 18 msec, and the same FOV and matrix size as the gradient-echo images. The intravoxel local field gradients within five pre-selected ROIs (indicated in Fig. 1b, consisting of five voxels in each) were measured with 2D x-z plane echo-shifted spin-echo images of slice thickness 1 mm. Other parameters were the same as those used in the x-y plane spin-echo imaging.

FIG. 1. a: Schematic of a phantom used in the evaluation of slice dephasing compensation methods. The phantom consists of an air-filled glass sphere embedded within a 5% gelatin-containing vial. The selected slice is indicated by two white lines, and five ROIs selected for comparison are indicated in red (ROI 1–5, from left to right). b: The x-y plane spin-echo reference image of the slice defined in a, with five ROIs indicated in red. c: The x-z plane internal gradient field map. d: Internal gradient profiles along the z-direction for ROI 1 (solid line), ROI 2 (dotted line), ROI 3 (dashed line), ROI 4 (dashed line with one dot), and ROI 5 (dashed line with three dots) with zero indicating the center of each ROI. The degree of field non-linearity is higher in regions close to the air sphere (e.g., ROI 1 and 2).
Five RF pulses were designed with their phase responses as the negative of the product of the measured intravoxel internal gradient (corresponding to five ROIs) and TE 16 msec. Based on the small tip angle assumption (10), the time-domain waveforms of the RF pulses were obtained by taking the inverse Fourier transform of the desired response profiles. This calculation is valid for gradient-echo imaging with a short TR, in which the flip angle is usually small. Twelve images were then acquired at different TEs (10–21 msec) using the designed RF pulses to demonstrate that spin phases can actually be refocused at a selected echo time. A 16-step gradient-based compensation method was also implemented. The gradient-based compensation method and the described RF refocusing technique were compared in terms of the image SNR in the same five pre-selected ROIs.

The proposed technique was also tested on a transverse-plane slice of a rabbit brain containing strong nonlinear internal gradients in the vicinity of the tissue/air interface. Using a surface RF coil, a gradient-echo image was acquired at TE 16 msec with a sinc RF pulse. Other MR imaging parameters include: matrix size 256 × 256, slice thickness 4 mm, FOV 51.2 × 51.2 mm², and TR 200 msec. A spin-echo reference image of the same slice was then acquired with same parameters except for TR of 500 msec. The 3D field inhomogeneity map with a matrix size 64 × 64 × 32 of the selected slice was obtained (16 min scan time) for RF pulse design. Two 3D echo-shifted spin-echo images were acquired with a readout gradient in the x-direction and phase encoding gradients in both the y and z-directions, after selection of a 4 mm slice slab along the y-direction. A 3D field map of the selected 4 mm slice slab was then derived from the two 3D images. An excitation RF pulse was designed with its slice profile phase response as the negative of the product of 16 msec TE and the internal gradient profile within the voxel. The designed RF pulse was then used to acquire a gradient-echo image at TE 16 msec for comparison.

RESULTS AND DISCUSSION
Results of the phantom studies are presented in Figs. 1–4. A reference x-y plane image acquired with a spin-echo imaging sequence is shown in Fig. 1b, with five pre-selected ROIs (five voxels each) indicated in red. Local field gradients within each ROI are obtained from an x-z plane field map (Fig. 1c), and are plotted in Fig. 1d. As shown, the local field gradients are highly nonlinear in areas close to the water/air interface (e.g., ROI 1 and 2), and the degree of non-linearity progressively decreases in areas distant from the air sphere (e.g., ROI 4 and 5). A gradient-echo image of the same phantom acquired at TE 16 msec using a sinc RF pulse is shown in Figure 2a. In comparison to the spin-echo image, severe signal loss originating from intravoxel internal gradient is observed. Five gradient-echo images acquired using RF pulses designed for the five

![Figure 2](image-url)

**FIG. 2.** a: The x-y plane gradient-echo image acquired with a sinc RF excitation pulse at TE 16 msec. Severe signal loss due to the intravoxel dephasing effect is observed. b–f: Gradient-echo images acquired with RF pulses designed to refocus the spin phases in each of the five ROIs at TE 16 msec. SNR is optimized in ROIs (indicated by cross-hairs) and the corresponding radially symmetric areas with similar field inhomogeneity.
selected ROIs are presented in Fig. 2b–f. High signal intensities in ROIs (indicated by cross-hairs) illustrate that signal loss can be locally recovered at a selected TE with appropriately designed RF pulses. We note that signal in other areas with similar internal gradient will also be refocused with an ROI-based RF pulse. Since the internal gradient originating from an air sphere is radially symmetric in the x-y plane, as indicated by Eq. [5] \((11,12)\), the recovered signal in Fig. 2 also shows the radial symmetry.

\[
\Delta \omega(x, y, z) = \frac{\Delta \chi R^3 \omega_0(2z^2 - x^2 - y^2)}{3(x^2 + y^2 + z^2)^{5/2}} \tag{5}
\]

where \(\Delta \chi\) is the susceptibility difference between air and water, \(R\) is the radius of the air sphere, and \(\omega_0\) is the main field strength along the z-direction.

The SNR in ROI 3 is plotted in Figure 3 as a function of TE for gradient-echo images acquired with a sinc and a designed RF pulse, for comparison. As shown, the signal decays with increasing echo time when a sinc RF pulse is used. The signal intensity, however, can be recovered with a field-map-based RF pulse, as indicated by Eq. [4]. Optimum SNR occurred at a pre-selected TE (16 msec), which corresponds to the maximum of the refocusing curve shown in Fig. 3. Efficiencies of the described RF refocusing technique and the 16-step gradient-based method are also compared in terms of the SNR for each ROI.

Table 1 lists the SNR of five ROIs measured in a) gradient-echo image (TE 16 msec) without compensation, b) images acquired with the field-map-based RF pulses, and c) 16-step gradient increment compensation with three different post-processing algorithms (2,5,6). As shown, the highest SNR for each ROI can always be obtained in a single step with an ROI-based RF pulse. Among the three post-processing algorithms for the gradient-based compensation, the average of the 16 images acquired with different compensation gradients (2) provides the best in-plane signal uniformity. However, the SNR is lowest in this approach. A higher SNR can be obtained through the use of the maximum-signal-map approach (6). However, its efficiency is comparable to the described RF refocusing technique only in areas with linear internal gradient (e.g., ROI 4.)

<table>
<thead>
<tr>
<th>ROI</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
</tr>
</thead>
<tbody>
<tr>
<td>No compensation</td>
<td>43.0</td>
<td>108.4</td>
<td>18.0</td>
<td>33.76</td>
<td>73.89</td>
</tr>
<tr>
<td>RF refocusing technique</td>
<td>84.0</td>
<td>189.7</td>
<td>212.2</td>
<td>203.0</td>
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<td>16-step gradient compensation</td>
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<td>45.1</td>
<td>48.1</td>
<td>41.8</td>
<td>30.6</td>
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<tr>
<td>Average</td>
<td>48.2</td>
<td>110.1</td>
<td>115.4</td>
<td>155.9</td>
<td>197.8</td>
</tr>
<tr>
<td>Maximum</td>
<td>22.1</td>
<td>82.7</td>
<td>94.3</td>
<td>82.5</td>
<td>65.2</td>
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</table>
and 5). In comparison, the Fourier transform approach (5) exhibits a better balance between signal uniformity and SNR than the other two processing algorithms for the gradient-based compensation. However, its recovery efficiency in selected ROIs is still lower than the proposed RF refocusing method. A higher efficiency of the ROI-based RF refocusing technique is achieved due to a) the optimized RF pulse design based on the field map, and b) compensation for both linear and nonlinear internal gradients, as indicated by Eq. [4].

The RF pulse designed to refocus the spin phases in ROI 3 is shown in Figure 4. With the small flip angle assumption, the time-domain waveform of the designed pulse (with real and imaginary components shown in Fig. 4a) is the inverse Fourier transform of the desired response (with magnitude and phase components shown in Fig. 4b). The desired magnitude response (dashed line in Fig. 4b) is the square waveform with thickness of 4 mm, and the desired phase response (solid line in Fig. 4b) is the negative of the product of the field profile in ROI 3 and the selected TE 16 msec. Actual magnitude and phase responses are also presented (in Fig. 4b) for comparison. They were measured experimentally with a 1D gradient-echo sequence with both slice-selection and readout gradients along the z-direction. Background phase terms were eliminated by subtraction of the sinc RF pulse phase response from that of the designed pulse. As shown in Fig. 4b, the measured and calculated phase responses are nearly identical, which validates the use of the small flip angle assumption.

In the above implementation, we have assumed 20 isochromat units in an image voxel (with 4 mm slice thickness and 0.2 mm field map resolution along the slice direction). The presented technique compensates for phase incoherence among those 20 isochromat units and disregards the dephasing within each unit. This assumption is appropriate to achieve high compensation efficiency for four out of five ROIs shown in Table 1. However, as the internal gradient increases, the signal degradation due to the intra-unit dephasing effects becomes apparent. This is clearly illustrated with the ROI 1 example in which <50% of the signal is recovered. Computer simulation (not shown) quantitatively explains the observed difference in compensation efficiency between ROI 1 and the other ROIs. Based on this simulation, a 2.5 times decrease in the isochromat unit size will result in an additional 50% increase in the ROI 1 SNR. Experimentally, this can be accomplished by increasing the resolution of the field map and the RF pulse waveform.

Application of this technique to imaging of a rabbit brain in the presence of strong nonlinear internal gradients is shown in Figure 5. In comparison to the spin-echo reference image (Fig. 5a), the gradient-echo image (Fig. 5b) shows a severe slice dephasing artifact near the tissue/air interface. Using the RF pulse designed to compensate for the internal gradients in a region indicated by an arrow (Fig. 5b), the observed signal loss is effectively recovered in the acquired image (Fig. 5c). Summation of the two gradient-echo images (Fig. 5b and c) is presented in Fig. 5d, which illustrates that the image quality can be greatly improved in two steps.

The described ROI-based technique can be applied to functional (f)MRI to recover the signal loss efficiently in critical brain regions with strong internal gradients, such as tissue/air interface. An important consideration is that the algorithm designed to compensate for the intravoxel dephasing effect would not eliminate the blood oxygenation level-dependent (BOLD) contrast. Susceptibility changes associated with hemodynamic response can still be present when the internal gradients in both baseline and activation images are compensated for by the same amount. The proposed RF refocusing technique can therefore preserve the BOLD contrast of functional response. This argument has been validated (6,9), and also applies to the present RF refocusing technique. Ongoing studies in our lab are directed toward the fMRI application of this new technique.

CONCLUSIONS

A technique has been proposed to compensate for the slice dephasing artifact due to both linear and nonlinear internal gradients in gradient-echo images. A slice-selective RF pulse based on a field map is applied to refocus the spin phases at a selected echo time. The proposed RF refocusing technique is more efficient than the incremental gradient-based methods in terms of the SNR in a specified ROI as well as the time required. The new method designed for
ROI-based signal recovery will be useful for fMRI studies of critical brain regions with a strong susceptibility effect (e.g., tissue/air interface).

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