Sensitivity and Performance Time in MRI Dephasing Artifact Reduction Methods

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Although shimming can improve static field inhomogeneity, local field imperfections induced by tissue susceptibility differences cannot be completely corrected and can cause substantial signal loss in gradient echo images through intravoxel dephasing. Dephasing increases with voxel size so that one simple method of reducing the effect is to use thin slices. Signal-to-noise ratio (SNR) can then be increased by averaging over the subslices to form the final, thick slice. We call this method subslice averaging or SSAVE. Alternatively, a range of different amplitude slice select rephase gradients can be used to compensate for different susceptibility induced gradient offsets. The final image can then be formed by combining individual images in a variety of ways: summation, summation of the squares of the images, forming the maximum intensity projection of the image set, and Fourier transformation followed by summation. We show here that, contrary to previous claims, the theoretical sensitivity (i.e., SNR divided by the square root of the imaging time) of all these alternative methods is very similar. However, performance time (i.e., minimum-imaging time) of the simplest method, SSAVE, is much shorter than any of the alternatives. This is confirmed experimentally on phantoms and anesthetized mice. Magn Reson Med 45:470–476, 2001. © 2001 Wiley-Liss, Inc.

Key words: susceptibility artifacts; field homogeneity; high field MRI; functional MRI

Dephasing artifacts in MRI are caused by static magnetic field imperfections either inherent to the magnet or induced by tissue susceptibility differences which arise in vivo principally from tissue–air interfaces. The resulting signal loss can severely reduce image quality. This phenomenon is particularly problematic at higher static magnetic field strengths that are increasingly used in functional MRI or in small animal studies to improve sensitivity. Gradient echo (GE) sequences are most sensitive to dephasing artifacts, especially when heavily $T_2^*$-weighted for BOLD fMRI and contrast enhanced perfusion imaging (1–3).

Several approaches that reduce these artifacts have been suggested (4–15). The simplest method is to reduce the voxel size by reducing the slice thickness (4). The slice thickness direction is often the major source of dephasing artifacts since the voxel width is usually much larger in the slice direction. Constable (5) rejected the use of thin slices on signal-to-noise ratio (SNR) and slice coverage grounds. Instead, Constable suggested a modification of a technique, described originally by Frahm et al. (6), and later exploited by Ordidge et al. (7), that employs a set of different amplitude slice select rephasing gradients. Each rephase gradient compensates for one particular level of susceptibility-induced gradient. A composite image can then be formed by summing all images (6), by summing the squares of the images (7) or by choosing each pixel value from the optimally refocused image (i.e., performing a maximum intensity project or MIP) (5). Yang et al. (8) concluded that the Frahm-Ordidge-Constable method (FOC) was inefficient and proposed a modification, “gradient echo slice excitation profile imaging” (GESEPI) that employs a Fourier transform (FT) in the slice select direction. GESEPI is, essentially, an oversampled 3D sequence. Pixel values are averaged over the third dimension to form the final image.

Here we reconsider the possibility of averaging over thin slices, an approach we call subslice averaging, SSAVE. The use of thin slices to reduce susceptibility artifacts is not new (4) and averaging over the thin slices to increase SNR is an obvious modification (although we are unaware of any description in the literature). SSAVE has, however, been rejected in the past on the grounds of poor sensitivity. In this article, we adopt the approach of Brunner and Ernst (16) and compare the sensitivities (i.e., SNR divided by the square root of imaging time) and minimum performance (or imaging) times of the different approaches to reducing dephasing artifacts. We show that the sensitivities of the different approaches are similar but that the minimum performance time of SSAVE is much shorter than alternatives.

**THEORY**

In GE sequences, spins accumulate a phase that is a function of field inhomogeneities induced by susceptibility differences. Spatial variation of these inhomogeneities will cause a phase dispersion across a voxel and hence loss of signal. If the susceptibility-induced field is modeled by a constant gradient $g_{sus}$ along the $z$-direction then the phase dispersion within a voxel, $\phi$, at the echo time, TE, is given by:

$$\phi = \gamma \cdot TE \cdot g_{sus} \cdot r$$  \[1\]

where $\gamma$ is the gyromagnetic ratio, and $r$ is the dimension of the voxel.

From this expression it is clear that reducing either the echo time (TE) or the voxel size ($r$) can reduce dephasing...
artifacts. The value of the echo time is usually dictated by the contrast requirements of the MR experiment being performed. For example, in BOLD fMRI TE is chosen to maximize sensitivity to changes in \( T_2^* \). The obvious strategy to reduce dephasing artifacts is therefore to decrease the voxel size in the slice direction. Averaging over groups of slices after reconstruction can then be used to increase SNR. Dephasing artifacts are minimized in the final thick slice since the phase of each subslice is discarded during the magnitude reconstruction.

**SENSITIVITY**

The sensitivities of SSAVE, GESEPI, FOC, and direct acquisition of thick slices can be compared using the following simplified expressions for minimum performance time, \( T_p \), and sensitivity, \( \eta \):

\[
T_p = T_{IM}/N_{EX}; \quad \eta = t \cdot \sqrt{N_{AV}/T_{IM}} \tag{2a}
\]

where \( T_{IM} \) is the total imaging time, \( N_{EX} \) is the number of times the imaging experiment is repeated (i.e., true signal averaging), \( t \) is the subslice thickness, and \( N_{AV} \) is the number of signal averaging steps inherent in each pulse sequence, including both phase encoding and subslice averaging steps. We assume that the repetition time, TR is equal to one in all cases. Then \( T_{IM} \) and \( N_{AV} \) can be expressed simply, as follows:

\[
T_{IM} = N_{EX} \cdot N_{PE} \cdot N_{RA}; \quad N_{AV} = N_{EX} \cdot N_{PE} \cdot N_{SS} \tag{2b}
\]

where \( N_{PE} \) is the number of phase-encoding steps, \( N_{RA} \) is the number of rephasing gradient amplitudes applied, and \( N_{SS} \) is the number of subslices used in each correction method. Substituting [2b] into [2a] yields the following simplified expressions for \( T_p \) and \( \eta \):

\[
T_p = N_{PE} \cdot N_{RA}; \quad \eta = t \cdot \sqrt{N_{SS}/N_{RA}}. \tag{3}
\]

Equation [3] will now be used to compare the sensitivities of the different methods in the absence of susceptibility-induced inhomogeneity. This is equivalent to assuming that all methods are equally effective in reducing artifacts. In the Discussion, we consider the implications of this assumption. The following results compare sequences with the same “correction factor,” \( N \). The correction factor for SSAVE and GESEPI is the ratio of the final slice width to subslice width. For FOC, the correction factor is \( N_{RA} \), the number of separate rephasing gradient amplitudes. The correction factor is not the same as the number of separate excitations since oversampling is required for GESEPI (8), which does not affect the subslice width. For simplicity, \( N_{EX} \) is assumed to be one for all pulse sequences. The different acquisition strategies are shown schematically in Fig. 1. The results and sequence parameters are summarized in Table 1.

**Direct Acquisition**

For the sake of simplicity, we assume the slice thickness is \( t = 1 \). The minimum performance time (\( T_p \)) and sensitivity (\( \eta \)) of direct acquisition are then both unity since \( N_{PE}, N_{RA}, \) and \( N_{SS} \) are all one.

**FOC**

In an FOC acquisition the final, thick slice is imaged directly but with \( N_{RA} = N \) different rephasing amplitudes. The parameters \( t, N_{SS}, \) and \( N_{PE} \) are therefore unity, the minimum imaging time is \( T_p = N \), and the sensitivity is \( \eta = 1/\sqrt{N} \).

**GESEPI**

The thick slice is divided into \( N \) subslices by phase encoding with a factor of \( n \) oversampling. In this case, \( t = 1/N, N_{SS} = N, N_{RA} = 1, \) and both \( N_{PE} \) and \( T_p = nN \). The sensitivity is again \( \eta = 1/\sqrt{N} \).

**SSAVE**

The thick slice is divided into \( N_{SS} = N \) subslices acquired within the same repetition time (TR), this time by selective excitation. Subslice thickness is \( t = 1/N \), while \( N_{PE}, N_{RA} \) and \( T_p \) are all unity. The sensitivity is again \( \eta = 1/\sqrt{N} \).

**MATERIALS AND METHODS**

All experiments were performed on a horizontal 7 T micro-imaging system (SMIS, Surrey, UK) with a 200-mm access horizontal magnet (Magnex Scientific, Abingdon, UK) equipped with 250-mT/m actively shielded gradients (ID = 120 mm). The RF coil was a 35-mm inner diameter quadrature birdcage tuned to 301 MHz.

**Phantom Experiments**

Two phantom experiments were performed. The first experiment was to compare sensitivity in images reconstructed by FOC, GESEPI, and SSAVE with the same correction factor (\( N \) in Table 1). The second experiment was to compare the efficacy of the three techniques in reducing dephasing artifacts.
TR was set to 300 ms, which was long compared to the selective excitation pulse was an 8-ms, five-lobe sinc pulse. Imaging parameters were identical for each approach. The sequences were GE and unless stated otherwise all imagings were as follows: echo time, TE = 50 ms; flip angle, $\alpha = 90^\circ$; field of view, FOV = 30 mm; matrix size = $128 \times 128$; bandwidth = $50$ kHz. For each method, SNR was calculated as the ratio of the mean of the phantom signal in a homogeneous ROI over the standard deviation of the background noise in an ROI in air.

**Direct Acquisition**

The 2-mm thick slice of the phantom was acquired directly with a number of signal averages, $N_{\text{ex}} = 16$.

**FOC**

Eight sets of data were acquired from the 2-mm slice with different slice select rephase gradient amplitudes ($N_{\text{ra}} = 8$). The minimum and maximum amplitudes were identical to those used in the GESEPI acquisition described below. The number of signal averages was $N_{\text{ex}} = 2$. Each dataset was processed separately with a 2D Fourier transform. In FOC the final image is formed from the maximum intensity projection (MIP) through the set of eight images. However, SNR was computed using the noise measured in a single image since the MIP process biases the noise estimate, increasing the mean value of the background noise.

**GESEPI**

An oversampled 3D sequence was used to image the 2-mm slice. Sixteen phase encoding steps were applied over an FOV of 4 mm, resulting in $2 \times$ oversampling ($N_{\text{pe}} = 16, n = 2$). No signal averaging was used for the GESEPI experiment ($N_{\text{ex}} = 1$). The data were Fourier-transformed to give $16 \times 0.25$ mm subslices. The central eight slices, covering the 2 mm excited slice, were averaged to form the final image ($N_{\text{ss}} = 8$).

**SSAVE**

Data were acquired from eight contiguous $0.25$ mm slices using a multislice sequence. The number of signal averages was $N_{\text{ex}} = 16$. The eight images were then averaged to form the final 2-mm slice image ($N_{\text{ss}} = 8$).

**Artifact Reduction Comparison**

In a second experiment, the ability of the three methods to reduce dephasing artifacts was assessed qualitatively by imaging a susceptibility phantom, created by placing a 5-mm diameter, air-filled cylinder perpendicularly across a 27-mm diameter, doped water-filled cylinder.

Imaging parameters were identical to those used in the sensitivity experiment (correction factor $N = 8$). The imaging time for the direct acquisition and each of the correction methods was $614.4$ sec ($N_{\text{ex}}$: Direct, 16; FOC, 2; GESEPI, 1; SSAVE, 16). Additional datasets were also acquired with correction factors of $N = 4$: 4 $\times$ 0.5-mm SSAVE subslices; $8 \times 0.5$-mm GESEPI subslices (including $2 \times$ oversampling); and four FOC rephasing amplitudes. The number of signal averages ($N_{\text{ex}}$) for these acquisitions were again chosen so the imaging times were equal ($614.4$ sec) for all methods ($N_{\text{ex}}$: FOC, 4; GESEPI, 2; SSAVE, 16).

**Mouse Brain Images**

The ability of the three correction methods to reduce dephasing artifacts was also evaluated in vivo. Images of mouse brains were acquired in regions of artifact that occur in the frontal lobe near the nasal cavities.

All animals were maintained under protocols approved by the Institutional Animal Care and Use Committee of New York University School of Medicine. Adult Swiss Webster mice (Taconic, Germantown, NY) were anesthetized with ketamine: xylazine (120:20 mg/kg) injected intramuscularly and body temperature was maintained above $30^\circ$C during the experiment with a homemade circulating water pad and monitored with a mouse rectal probe (YSI, Yellow Springs, OH). Mouse brain images

<table>
<thead>
<tr>
<th>Method</th>
<th>Subslice thickness</th>
<th>Phase encoding steps</th>
<th>Rephasing amplitudes</th>
<th>Averaged subslices</th>
<th>Sensitivity</th>
<th>Minimum performance time</th>
</tr>
</thead>
<tbody>
<tr>
<td>Direct</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>FOC</td>
<td>1</td>
<td>1</td>
<td>N</td>
<td>1</td>
<td>1</td>
<td>$1/\sqrt{N}$</td>
</tr>
<tr>
<td>GESEPI</td>
<td>$1/N$</td>
<td>$nN$</td>
<td>1</td>
<td>N</td>
<td>$1/\sqrt{N}$</td>
<td>$nN$</td>
</tr>
<tr>
<td>SSAVE</td>
<td>$1/N$</td>
<td>1</td>
<td>1</td>
<td>N</td>
<td>$1/\sqrt{N}$</td>
<td>1</td>
</tr>
</tbody>
</table>

Sensitivity and performance time are calculated using Eq. [3]. For FOC, the number of separate refocusing pulses is $N$. For GESEPI and SSAVE the number of subslices is $N$. The GESEPI oversampling factor is $n$. TR is assumed to be equal in all cases.

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**Table 1**

<table>
<thead>
<tr>
<th>Method</th>
<th>Subslice thickness</th>
<th>Phase encoding steps</th>
<th>Rephasing amplitudes</th>
<th>Averaged subslices</th>
<th>Sensitivity</th>
<th>Minimum performance time</th>
</tr>
</thead>
<tbody>
<tr>
<td>Direct</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
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</tr>
<tr>
<td>FOC</td>
<td>1</td>
<td>1</td>
<td>N</td>
<td>1</td>
<td>1</td>
<td>$1/\sqrt{N}$</td>
</tr>
<tr>
<td>GESEPI</td>
<td>$1/N$</td>
<td>$nN$</td>
<td>1</td>
<td>N</td>
<td>$1/\sqrt{N}$</td>
<td>$nN$</td>
</tr>
<tr>
<td>SSAVE</td>
<td>$1/N$</td>
<td>1</td>
<td>1</td>
<td>N</td>
<td>$1/\sqrt{N}$</td>
<td>1</td>
</tr>
</tbody>
</table>

Summary of Sensitivity and Performance Time Values for the Different Methods of Reducing Dephasing Artifacts

SNRs were compared in images of a uniform 16-mm diameter cylinder filled with gadolinium-doped water (3 mM Magnevist; Berlex Laboratories, Wayne, NJ). Four different methods were used to form images of 2-mm thick slices (Fig. 1). For the three correction methods, the correction factor was $N = 8$: $8 \times 0.25$ mm SSAVE subslices, $16 \times 0.25$ mm GESEPI subslices (including $2 \times$ oversampling), eight FOC rephasing gradient amplitudes. This correction factor is arbitrary, since the purpose of the experiment is to compare sensitivity in the absence of artifacts. The number of signal averages ($N_{\text{ex}}$) for each method was chosen to equalize imaging time ($T_{\text{im}} = 614.4$ sec). All sequences were GE and unless stated otherwise all imaging parameters were identical for each approach. The selective excitation pulse was an 8-ms, five-lobe sinc pulse. TR was set to 300 ms, which was long compared to the $T_1$ of the phantom ($T_1 \sim 50$ ms). Other sequence parameters were as follows: echo time, TE = 20 ms; flip angle, $\alpha = 90^\circ$; field of view, FOV = 30 mm; matrix size = $128 \times 128$; bandwidth = 50 kHz. For each method, SNR was calculated as the ratio of the mean of the phantom signal in a homogeneous ROI over the standard deviation of the background noise in an ROI in air.
were acquired with the following sequence parameters: 5.3 ms, 3-lobe sinc excitation pulse; repetition time, TR = 300 ms; echo time, TE = 20 ms; flip angle, $\alpha = 45^\circ$; matrix size = $128 \times 128$; field of view, FOV = 30 mm; slice thickness = 1 mm; bandwidth = 50 kHz. The resulting image resolution was $234 \mu m \times 234 \mu m \times 1000 \mu m$. SNRs were calculated as the image intensity measured in an ROI in a homogeneous region of brain tissue, divided by the standard deviation of the noise in an ROI in air.

**RESULTS**

**Phantom Images**

**Sensitivity Comparison**

The sensitivities of the four methods were compared in 2-mm slice images of a uniform phantom. Table 2 shows the measured SNR and sensitivity values for direct acquisition and the three correction methods. The results show that the measured sensitivities for all correction methods are very similar.

**Artifact Reduction Comparison**

Images of the susceptibility phantom were acquired by each method with correction factors of four or eight (Fig. 2). The dephasing seen in the directly acquired slice (Fig. 2a) is greatly reduced by all three methods using a correction factor ($N_{rephasing \, amplitudes} / subslices$) of 4 (Fig. 2b–d), and nearly eliminated using $N = 8$ (Fig. 2e–g). FOC (Fig. 2b,e) leaves more residual artifact than GESEPI (Fig. 2c,f) or SSAVE (Fig. 2d,g). We confirmed the observation of Yang et al. (8) that oversampling is necessary to avoid artifacts due to signal wrap in GESEPI, although we found

<table>
<thead>
<tr>
<th>Method</th>
<th>Subslice thickness t (mm)</th>
<th>Number of phase encoding steps $N_{PE}$</th>
<th>Number of rephasing amplitudes $N_{RA}$</th>
<th>Number of averaged subslices $N_{SS}$</th>
<th>Number of signal averages $N_{EX}$</th>
<th>SNR in final slice</th>
<th>Sensitivity $\eta = \frac{SNR}{\sqrt{T_{IM}}}$ (1/\sqrt{s})</th>
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</thead>
<tbody>
<tr>
<td>Direct</td>
<td>2</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>16</td>
<td>838.8</td>
<td>33.8</td>
</tr>
<tr>
<td>FOC</td>
<td>2</td>
<td>1</td>
<td>8</td>
<td>1</td>
<td>2</td>
<td>322.7</td>
<td>13.0</td>
</tr>
<tr>
<td>GESEPI</td>
<td>0.25</td>
<td>16</td>
<td>1</td>
<td>8</td>
<td>1</td>
<td>297.7</td>
<td>12.0</td>
</tr>
<tr>
<td>SSAVE</td>
<td>0.25</td>
<td>1</td>
<td>1</td>
<td>8</td>
<td>16</td>
<td>285.5</td>
<td>11.5</td>
</tr>
</tbody>
</table>

FOC, 8 rephasing steps; GESEPI, 16 phase encoding steps over a 4-mm FOV (2× oversampling); SSAVE, 8 × 0.25-mm subslices. The imaging time for each method was the same (614.4 sec).
that 2×-oversampling was sufficient to eliminate these effects.

Mouse Brain Images

Figure 3 shows 1-mm thick coronal brain images (Fig. 3b–f) acquired at the level of the frontal lobes, indicated in the sagittal scout image (Fig. 3a). The imaging time for the directly acquired slice (Fig. 3b) was 38.4 sec ($N_{EX} = 1$). SNR = 22.4 was measured in an ROI centered on the caudate putamen (striatum). The minimum imaging time with SSAVE was the same as the directly acquired slice (Fig. 3c; 38.4 sec, $N_{EX} = 1$, $N_{SS} = 4$), resulting in an average of four 250-μm subslices that clearly reduces the dephasing artifacts at the cost of SNR loss (SNR = 18.2). SSAVE images were also acquired with signal averaging (Fig. 3f; 38.4 sec, $N_{EX} = 8$, $N_{SS} = 4$) in the same time (307.4 sec) as FOC (Fig. 3d; $N_{EX} = 4$, $N_{SS} = 4$) and GESEPI (Fig. 3e; $N_{PE} = 8$ including 2×-oversampling, $N_{EX} = 1$). In each case the dephasing artifacts were clearly reduced and SNRs were similar, as expected: $\text{SNR}_{\text{FOC}} = 49.8$, $\text{SNR}_{\text{GESEPI}} = 49.5$, $\text{SNR}_{\text{SSAVE}} = 49.1$.

DISCUSSION

We have shown both theoretically and experimentally that when the number of SSAVE and GESEPI subslices and the number of FOC rephase gradient amplitudes are equal, sensitivity is also equal in the absence of dephasing. This leads to the obvious question of the relative efficacy of the techniques in reducing artifacts when susceptibility dephasing is present. In other words, does each method require the same number of correction steps to achieve adequate artifact reduction? A partial answer can be given if we make the simplifying assumption that susceptibility differences induce linear gradients across the voxel. GESEPI defines the subslices by phase encoding. With linear gradient offsets, it is straightforward to show that no signal is lost until a cutoff gradient value is reached at which point all signal is lost (17). GESEPI subslice thickness would normally be reduced so that the cutoff value is greater than the largest offset gradient. By contrast, SSAVE defines subslices by selective excitation. Signal response to a susceptibility gradient then follows a sinc function and even small dephasing gradients will cause some signal loss. “Optimum” subslice thickness then depends on how much nonuniformity one is prepared to tolerate. In the comparison above, we made the assumption that GESEPI and SSAVE subslice thicknesses were equal. With this assumption, it is straightforward to show that SSAVE signal loss at the GESEPI cutoff is 34% (19). FOC methods selectively excite the thick slice directly. The sensitivity of each acquisition to offset gradients is again a sinc function. Each acquisition exactly compensates for some values of dephasing gradient and is less effective for others. The
images are combined to form a more uniform response across a range of offset values. Again, it can be shown that a maximum 34% reduction in signal will occur at the GESEPI cutoff value (19).

An optimized GESEPI acquisition may, then, have a sensitivity advantage where susceptibility gradients are close to the GESEPI cutoff. However, optimizing GESEPI so precisely would be time-consuming and, in practice, it is likely that some standard "safe" parameters would often be used. Under those circumstances, GESEPI advantages would be minimal. In addition, if the GESEPI parameters are set incorrectly, SSAVE may actually have an advantage since the method loses signal only slowly with increasing inhomogeneity.

The discussion above assumes linear offset gradients. This assumption is reasonable if the voxels are much smaller than the structures causing the susceptibility differences. With larger voxels the patterns of inhomogeneity will be more complex and the exact responses of GESEPI, FOC, and SSAVE are more difficult to predict. It seems, however, that nonlinear offsets are more likely to change the shape of the response than its width. This is confirmed in Fig. 2, which shows that artifacts for both SSAVE and GESEPI are almost identical at both subslice thicknesses. With 0.5-mm subslices SSAVE and GESEPI both show residual artifacts (Fig. 2c,d). It is clear that differences in the efficacy of artifact suppression between SSAVE and GESEPI are small. In both cases the artifact is almost entirely eliminated with 0.25-mm subslices (Fig. 2f,g). There is more residual artifact in the FOC images (Fig. 2b,e) that results from the variation in correction efficacy with susceptibility gradient, but again correction efficacy is similar.

At first sight it might appear odd that the 2D SSAVE method should give the same sensitivity as the 3D GESEPI technique, since it is widely accepted that 3D sequences provide better sensitivity than 2D sequences. This belief is based on the work of Brunner and Ernst (16). However, although Brunner and Ernst anticipated multislice imaging, their original comparison was between 3D and sequential 2D slice imaging. We showed recently that when 2D multislicing is considered, the sensitivities of 2D and 3D sequences are usually very similar (18).

The apparent increase in background noise seen in the FOC images in Fig. 3 is due to the MIP algorithm used but does not indicate a decrease in true SNR. With the MIP algorithm, each pixel value is derived from a single image, without any kind of averaging. SNR within the pixel must, therefore, be equal to that in the original image. The mean background is increased, however, since the MIP selects the maximum noise value from all images. Measuring SNR as signal relative to background noise within a single image, as we did in the artifact reduction comparison in phantom and in vivo, avoids the problem of biasing the noise estimate. The sensitivity of other FOC reconstruction schemes may be slightly different to the MIP used here. Summation is likely to reduce sensitivity since noise from all images is added into the final image, whereas substantial signal only exists in one or two images. Sum of squares effectively weights the addition so that only images with substantial signal contribute significantly to the final image. If substantial signal only exists in one image, the SNR will then be identical to that of the MIP image. If signal exists in two images there might be an increase in sensitivity, but we believe it will be small.

SSAVE requires narrow slices, which are obtained by either increasing the gradient strength or using longer RF pulses. For a given repetition time, the latter is often the only possible alternative due to gradient strength limitations and therefore increases the amount of time "wasted." Therefore, in some cases there may be a slight time and sensitivity disadvantage for SSAVE relative to the other techniques. Furthermore, slice overlap (cross-talk) in multislice sequences such as SSAVE is a potential problem, especially if spin-echo sequences are employed. We did not observe any SNR loss or artifacts that could be ascribed to cross-talk with the GE sequences used in these studies.

SSAVE can suffer a sensitivity disadvantage when the minimum TR needed to accommodate the required number of subslices is greater than would otherwise be chosen. This is not generally true in T1-weighted gradient echo sequences since the enhanced recovery with the longer TR of the 2D sequence compensates for the larger number of averages in the 3D sequence (18). In T2- and T2*-weighted sequences, however, if a large number of subslices is required, TR, and hence imaging time, would have to be increased with the result of a reduction in sensitivity.

SSAVE offers several advantages over alternative techniques. First, SSAVE is easily implemented on any MR scanner using any interleaved, multislice sequence. Second, SSAVE minimum performance time will often be much shorter than FOC or GESEPI, even when TR has to be increased to accommodate the subslices. Both FOC and GESEPI require imaging times that are several times the single slice imaging time since each acquisition has to be repeated with multiple slice select refocusing/phase encoding gradients. GESEPI can be particularly time-consuming since oversampling is necessary to overcome aliasing. The original authors used oversampling factors of four and above (8). We found that oversampling was indeed necessary to avoid artifacts, but found an oversampling factor of two to be adequate. With both GESEPI and FOC the imaging time is prohibitively long for dynamic studies such as contrast enhanced perfusion imaging. A further advantage of SSAVE is that slice coverage can be extended by increasing the slice separation, albeit with the obvious disadvantage of interslice gaps. Finally, EPI implementations of SSAVE and FOC may be less prone to movement artifacts than GESEPI. A GESEPI-EPI sequence is a multislice refocusing/phase encoding technique so that subject movement between shots will produce artifacts in the slice-select/phase-encoding direction. EPI versions of FOC and SSAVE are single-shot techniques so that movement will cause misregistration of the subslices (which is in principle correctable by postprocessing) rather than artifacts.

We have used SSAVE to correct dephasing artifacts in mouse brain images, but it could also be employed in imaging other regions of the body with susceptibility-induced gradients, such as abdominal imaging, cardiac imaging, and in imaging protocols where catheters, implants, probe temperature, and other clamps induce local inhomogeneities.

Fernandez-Seara and Wehrli (20) have recently introduced a very different approach to correcting dephasing
artifacts. Their method fits a modeled function to a multiecho sequence and may thus be more appropriate for $T_2^*$ measurements than the single echo acquisitions considered here.

In conclusion, all three methods investigated reduce dephasing artifacts at the cost of lower sensitivity, relative to direct acquisition. SSAVE has the advantages of speed (minimum performance time) and flexibility. The method can be implemented with any multislice sequence without modification. SSAVE will be particularly advantageous in situations where short imaging times are required to follow dynamic changes (MRI and perfusion imaging). Clearly, the sequence used in Fig. 3 is not appropriate for dynamic imaging. However, single-shot, single average EPI images of the human head give adequate SNR, even with narrow slices and could be utilized in a SSAVE acquisition. Minimum imaging times of GESEPI and FOC are, however, much longer than that of SSAVE, so that the former approaches generally cannot be used for dynamic imaging. On the other hand, if averaging is necessary to provide acceptable SNR GESEPI and FOC will be as good as SSAVE.

REFERENCES