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An Improved Susceptibility Weighted Imaging Method using Multi-Echo Acquisition

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Abstract

Purpose—To introduce novel acquisition and post-processing approaches for susceptibility weighted imaging (SWI) to remove background field inhomogeneity artifacts in both magnitude and phase data.

Method—The proposed method acquires three echoes in a 3D gradient echo (GRE) sequence, with a field compensation gradient (z-shim gradient) applied to the third echo. The artifacts in the magnitude data are compensated by signal estimation from all three echoes. The artifacts in phase signals are removed by modeling the background phase distortions using Gaussians. The method was applied *in vivo* and compared with conventional SWI.

Results—The method successfully compensates for background field inhomogeneity artifacts in magnitude and phase images, and demonstrated improved SWI images. In particular, vessels in frontal lobe, which were not observed in conventional SWI, were identified in the proposed method.

Conclusion—The new method improves image quality in SWI by restoring signal in the frontal and temporal regions.

Keywords

compensation gradient; field inhomogeneity induced artifact; Gaussian modeling removal; magnetic susceptibility; z-shim

Introduction

Susceptibility Weighted Imaging (SWI) is an MRI method which emphasizes the magnetic susceptibility differences between brain tissues (1-3). The susceptibility differences between water, deoxy-hemoglobin, iron, myelin, calcium, etc. induce local magnetic field perturbations which are identified in magnitude and phase signals obtained by gradient echo (GRE) imaging (2,4-10). SWI has been applied to visualize vein structures (1), to detect iron deposition in deep gray matter (5), and to identify intracranial hemorrhage in stroke and traumatic brain injury (11,12).

In conventional SWI, a flow-compensated 3D GRE sequence is used with a relatively long echo time ($TE = \sim 25$ ms at 3 T) to produce sufficient susceptibility weighting. To generate an SWI image, a reconstructed complex image is first separated into magnitude and phase images. The phase image is high-pass filtered, and subsequently scaled and thresholded to generate a phase mask. Finally, this phase mask is multiplied by the magnitude image to generate an SWI image. In certain applications (e.g., venography), neighboring slices are combined using minimum intensity projection (mIP) to better visualize veins (1).

Since the method is sensitive to all field perturbations, SWI images are not only affected by local magnetic susceptibility differences but also by non-local macroscopic field inhomogeneity such as the susceptibility differences near air and tissue interfaces. As a result, images in conventional SWI suffer from artifacts in brain regions where macroscopic (or background) field inhomogeneity is severe (e.g. frontal and temporal lobes) (Fig. 1). In these areas, identifying local susceptibility sources (e.g. vessels) is challenging, thus hindering the applicability of SWI (Fig. 1c). The spatial manifestation of these artifacts is much wider in phase than in magnitude (Figs. 1a and 1b).

A few approaches have been proposed to compensate for these artifacts in SWI. These approaches remove artifacts only in phase data (13-15), thus the resulting SWI images still have remaining artifacts from the signal dropout in magnitude data as shown in Fig. 1a.

Aside from SWI, methods have been developed to compensate for the artifacts induced by background field inhomogeneity in GRE imaging. The magnitude signal dropout has been restored by fitting a signal decay to a model (16-18) or by adding field compensation gradients (so-called *z*-shim) (19-23). Recently, signal phase in GRE has also been demonstrated to contain useful information (24,25). Several methods have been proposed to compensate for the phase artifacts caused by background field inhomogeneity in GRE images (24-27). To date, these methods have been applied to compensate either magnitude or phase images and have not been used in SWI.

In this study, we developed a new method that combines a modified *z*-shim method and a new phase artifact correction scheme to reduce image artifacts in SWI. Compared to previous compensation methods in SWI, our technique compensates for *both* magnitude and phase artifacts, thereby improving spatial coverage in the brain. The proposed method is applied *in vivo* and compared with the conventional SWI method.

Methods

MR Sequence

The proposed MRI pulse sequence is shown in Fig. 2. A 3D GRE sequence is modified to acquire three echoes with a field compensation gradient (G_c) applied prior to the third echo. The TE of the first echo is chosen to have a moderate susceptibility weighting ($TE_1 = 8.7$ ms). The second echo is acquired with TE ($TE_2 = 25$ ms) and bandwidth (= 114 Hz/pixel) similar to conventional SWI, whereas the first and third echoes have a higher bandwidth (= 455 Hz/pixel). Before acquiring the third echo, a field compensation gradient is applied to compensate for the field inhomogeneity across the slice. The size of the field compensation gradient ($G_c = -1.75$ G/cm and $t_c = 0.98$ ms) is similar to the optimal compensation gradient suggested in the head (22), considering slice thickness difference. This gradient is positioned immediately after the second echo and is followed by the third echo to minimize the increase in repetition time (TR) over conventional SWI. Similar to conventional SWI, 3D flow compensation is applied for the second echo. In the phase and slice encoding directions, the standard flow compensation schemes are applied (1). In readout, the flow compensation consists of two parts as follows: (i) in the first echo by using three gradients of the same duration with 1:-2:1 amplitude ratio (28), and (ii) before the second echo by applying conventional flow compensation in readout.

SWI Image Processing

For SWI image processing, the brain is separated into two distinct regions by thresholding the unwrapped phase image (ϕ_{UW} ; see below) from the 2nd echo as follows: areas that are not affected by background field inhomogeneity artifacts (Region 1; $\phi_{UW} < -4\pi$), and areas that are affected by these artifacts (Region 2; $\phi_{UW} > -4\pi$). The threshold (-4π) was empirically chosen to have slightly broader margins than the magnitude signal dropout areas for Region 2. In Region 1, an SWI image is generated from the second echo magnitude and phase images using conventional SWI post-processing. In Region 2, the signal magnitude is estimated from all three echoes and the phase mask is generated from the first echo phase. Finally, a complete SWI image is generated by combining these two regions by addition.

As in conventional SWI, phase masks are created from the phase images. However, additional processes are included to remove artifacts in the phase masks. First, the phase images are unwrapped using a Laplacian unwrapping (29,30), given as:

$$\phi_{UW} = FT^{-1} \left(\frac{FT \left(\cos \phi_{in} \cdot FT^{-1} (k^2 \cdot FT(\sin \phi_{in})) - \sin \phi_{in} \cdot FT^{-1} (k^2 \cdot FT(\cos \phi_{in})) \right)}{k^2} \right) \quad [\text{Eq. 1}]$$

where ϕ_{UW} is an unwrapped phase image, FT is a 3D Fourier transform, FT^{-1} is a 3D inverse Fourier transform, ϕ_{in} is an input phase image, and k^2 is $k_x^2 + k_y^2 + k_z^2$ in k -space. Laplacian phase unwrapping is computationally efficient (31) compared to other methods (15,32), but may introduce low-frequency phase signal (see Results and Discussion).

After Laplacian unwrapping, the background phase is modeled by a series of Gaussians as follows:

$$\phi_{BG} = G \otimes s_G^* \quad \text{where } s_G^* = \underset{s_G}{\operatorname{argmin}} \|M(\phi_{UW} - G \otimes s_G)\|_2 \quad [\text{Eq. 2}]$$

where ϕ_{BG} is a background phase image, G is a 3D Gaussian with standard deviation $\sigma_x = 0.05$, $\sigma_y = 0.05$, and $\sigma_z = 0.05$ in each axis, $\otimes$ represents a 3D convolution, s_G is the unknown distribution of the 3D Gaussians for all voxels, s_G^* is the estimated distribution for s_G , M is a weighting factor, and $\|\cdot\|_2$ is L^2 -norm. For the weighting factor (M), the magnitude image that corresponds to the phase image is used. The minimization was performed using an iterative conjugate gradient algorithm (33). Once this estimated background phase image is obtained, it is subtracted from the original phase image in complex domain (i.e. $e^{i(\phi_{in} - \phi_{BG})}$). This background field correction step removes large field inhomogeneity particularly in the frontal and temporal regions.

After the background field correction, the image (a complex image with the original magnitude and background field corrected phase) is further filtered using a high-pass homodyne filter (filter size = 55×55) to eliminate any residual low-frequency phase variation. This filter size removes objects larger than 4 mm and has been widely used in SWI (2). Finally, the phase mask is generated by scaling and thresholding the corresponding homodyne filtered phase image. These two steps are the same as in conventional SWI processing.

Different phase mask schemes are applied to the different regions. In Region 1, a negative phase mask is generated using the phase-corrected data from the second echo. In Region 2, the phase mask is calculated using a symmetric triangular mask method (34) from the first echo phase processed by the proposed method (see Discussion for the motivation of using different mask schemes). Finally, the two phase masks are combined by addition to generate a complete phase mask.

The magnitude signal in Region 2 is estimated by modeling signal decay with field inhomogeneity gradient. Only through-plane directional field inhomogeneity is assumed and corrected because SWI is often acquired in a lower through-plane resolution compared to inplane resolution. The signal in a given voxel (z_0) is modeled as follows (22,35)

$$S(TE_n; z_0) = M_0 \cdot \exp(-TE_n \cdot R_2^*) \cdot A(G_z \cdot TE_n; z_0) \quad \text{where } A = \int_{-z_{sl}/2}^{z_{sl}/2} e^{j2\pi\gamma G_z(z-z_0)TE_n} \cdot P(z - z_0) dz \quad [\text{Eq. 3}]$$

$$P(z) = \frac{\operatorname{sinc}(\frac{z}{\Delta z})}{\operatorname{sinc}(\frac{z}{z_{sl}})}$$

where n is echo number ($n = 1, 2$, and 3), M_0 is an equilibrium magnetization, G_z is field inhomogeneity gradient in slice, z_{sl} is a 3D slab thickness, $P(z)$ is a slice profile (36), and z is the slice thickness. In the first and second echoes, G_z is the same as through-plane background field inhomogeneity ($G_{z,bg}$). In the third echo, G_z becomes $G_{z,bg} - G_{c,eff}$ where

$G_{c,eff}$ is an effective compensation gradient at TE_3 ($G_{c,eff} = G_c \times t_c / TE_3$). The three unknown parameters (M_0 , R_2^* and $G_{z,bg}$) are estimated from the measured signals from the three echoes using an iterative non-linear curve fitting algorithm (Levenberg-Marquardt algorithm) (37,38). The initial values are set as follows:

$M_0 = S(TE_1) \cdot \exp(TE_1 \cdot R_{2,init}^*)$, $R_{2,init}^* = 16.67s^{-1}$ and $G_{z,bg} = 0$ G/cm. Once these three parameters (M_0 , R_2^* , and $G_{z,bg}$) are estimated, the magnitude signal at the second echo in Region 2 is estimated using Eq. 3. A final magnitude image is generated by adding the resulting magnitude estimation in Region 2 and the second echo magnitude in Region 1.

Finally, an SWI image is generated by multiplying the region-combined magnitude image with the fourth power of the region-combined phase masks. Minimum intensity projection (mIP) images are obtained by finding the voxel-wise minimum value over eight adjacent slices of the resulting SWI images. In Region 2, reduced SWI contrast is expected because data are not acquired with optimal SWI weighting.

Experiments

For *in vivo* experiments, 17 subjects (12 males) were recruited (age: 29.6 ± 3.7). All subjects signed a consent form approved by institutional internal review board prior to the experiment. All MR data were collected using a 3 T MRI system (ISOL Technology, South Korea) at Korea Advanced Institute of Science and Technology (KAIST). A quadrature birdcage RF coil was used for signal transmission and reception. The scan parameters were TR = 50 ms, TE = 8.7, 25.0 and 32.1 ms, Matrix size = $220 \times 220 \times 48$, FOV = $220 \times 220 \times 57.6$ mm³, resolution = $1 \times 1 \times 1.2$ mm³, and flip angle = 16° . The bandwidth was 455 Hz/pixel for the first and third echoes and 114 Hz/pixel for the second echo. The total scan time was 8 min 48 sec. These acquisition parameters are summarized in Table 1. All data were processed using MATLAB (ver. 8.0, Mathworks, Natick, MA, 2012). An SWI image was generated using the proposed method. For comparison, a conventional SWI image was generated from the second echo magnitude and phase images using standard SWI reconstruction techniques. Lastly, to further validate our proposed method, an additional intermediate SWI image was constructed by applying our phase compensation pipeline to the conventional SWI data collected from the second echo. The purpose of this intermediate SWI image was to demonstrate the necessity of the additional magnitude signal compensation in our proposed method.

Results

Figure 3 shows phase images from each step of the conventional SWI method (Figs. 3b and 3c) and the proposed post-processing pipeline (Figs. 3d – 3g). The original phase image from the second echo shows large background field inhomogeneity in the frontal lobe (Fig. 3a). The conventional method results in substantial phase artifact after the homodyne filtering (Figs. 3b and 3c). By comparison, the phase variation artifact is substantially reduced after phase correction using Laplacian unwrapping (Fig. 3d) and background phase removal using Gaussian modeling (Fig. 3e). Laplacian unwrapping introduces unwanted low frequency phase variations (Fig. 3e). This artifact, in addition to other low frequency variations, is successfully removed by the homodyne filter (Fig. 3f). The resulting filtered

phase image (Fig. 3f) demonstrates that the proposed method compensates for background field inhomogeneity in most of the frontal lobe region. Despite successful compensation of the phase image from the second echo, a small region in the frontal lobe still shows erroneous phase values (Fig. 3f; red arrow). This artifact originates from signal dropout in the magnitude image (Fig. 4b), which results in unreliable phase values. In the proposed method, this area is classified as part of Region 2 (the areas substantially affected by field inhomogeneity), and therefore the phase mask is substituted by the processed phase signal from the first echo. The final phase mask from the summation of the two phase masks shows little background field artifact (Fig. 3g), demonstrating substantial improvement compared to from the phase mask from the conventional method (Fig. 3c).

Figure 4 shows the proposed magnitude image processing. The magnitude image from the second echo is used in Region 1 (Fig. 4d) whereas the magnitude image is estimated from all three magnitude images in Region 2 (Figs. 4a – 4c). The region-combined magnitude image (Fig. 4f) demonstrates signal restoration in the frontal lobe.

Figure 5 shows representative SWI and mIP images using the conventional (left column), the phase-artifact corrected (middle column) and the proposed (right column) methods. The SWI and mIP images generated by the conventional method show substantial artifacts in the frontal and temporal brain regions (left column of Fig. 5). Once phase artifacts are removed using the proposed phase processing, artifact regions are reduced. However, signal dropout in the magnitude images persists resulting in the artifacts in the SWI and mIP images (middle column of Fig. 5). These artifacts are successfully removed in the SWI and mIP images generated by the proposed method with magnitude compensation as shown in the right column of Figure 5. Outside of these artifact-prone regions, all three methods show similar SWI contrast. In the proposed method, large veins that are lost in the conventional method are clearly visible (red arrow). Similar improvement was observed in all subjects. The SWI images generated by the proposed method show slight signal intensity variation within Region 2 and at the boundary of the two regions. This may originate from insufficient magnitude correction (see Discussion). These variations are less conspicuous in the mIP images.

Discussion

The method proposed in this article successfully compensates for the susceptibility artifacts caused by background field inhomogeneity in SWI. In particular, the method reduces both magnitude and phase artifacts in the frontal and temporal brain regions and allows for successful delineation of vessels in the frontal lobe area.

In the proposed method, the SWI contrast can be reduced in Region 2 as compared to Region 1. The estimated magnitude from the multi-echo signals may have a reduced level of local magnetic susceptibility contrast (e.g. in vessels) as shown in a similar *z*-shim method (Fig. 6c in Nam *et al.* (22)). Additionally, the phase mask in Region 2 is generated using sub-optimal susceptibility weighting ($TE_1 = 8.7$ ms) to avoid magnitude signal dropout. Hence, the resulting SWI image is expected to have reduced SWI contrast. Despite the reduction in contrast, large veins in the frontal lobe are still observed demonstrating the

usefulness of the method. Additional susceptibility weighting in the phase signal can be obtained by increasing TE_1 or increasing the number of the phase mask multiplications to the magnitude image in Region 2.

Two different phase mask schemes were applied to the two regions. In Region 1, a negative phase mask, which is commonly used in venography, was applied. In Region 2, a triangular phase mask was used. The triangular phase mask generates a slightly blurred phase mask but is advantageous in capturing vessels in a wide range of orientations (39). In our experiment, this mask provided better vessel conspicuity in the frontal region of the brain potentially due to the orientation of the vessel.

In the phase image processing, Laplacian unwrapping was used as a computationally efficient 3D phase unwrapping approach. The method, however, did not produce an exact phase unwrapping result and introduced low frequency phase variations (Fig. 3e). This low frequency variation should not be considered as a limitation as it was successfully removed by the homodyne high-pass filter. Hence, Laplacian unwrapping is a good compromise between computational efficiency and unwrapping accuracy. As an alternative, other 3D phase unwrapping techniques could be applied (32,40).

The homodyne filter size (55×55) used in the proposed method has been widely used in SWI and has shown a good susceptibility weighting (2,3). The size of the homodyne filter may be increased to reduce phase artifact areas. However, the change is not favorable because it reduces the SWI contrast (2).

Different from the homodyne filtering that removes phase variation based on the spatial frequency, the Gaussian model fitting can be designed to remove phase variation in the frontal lobe as shown in Fig. 3e. The Gaussian modeling may decrease local magnetic susceptibility contrast particularly if the size of Gaussian is of comparable size to the susceptibility source. In our processing, a relatively large size Gaussian ($\sigma = 0.05$) was used to avoid over-fitting. Thus, little contrast is lost by this process. As a result, the conventional SWI images show similar SWI contrast to those from the proposed method in Region 1 (Fig. 5).

The total scan time increase from the proposed method is marginal. The field compensation gradient and the third echo (combined ~ 3.2 ms) only increase the minimum TR by approximately 10 %. This modest increase in scan time is outweighed by the substantial improvements in the magnitude signal compensation and quality of SWI images.

In our experiment, TR was set to 50 ms rather than the minimum TR (~ 33 ms) because of the slow analog-to-digital conversion speed in the system. The low SNR of the birdcage head coil used in this experiment limited the image resolution to $1 \times 1 \times 1.2$ mm³.

The SWI images in the proposed method (Fig. 5) show slight signal intensity variation within Region 2 and at the boundary of the two regions. This may have originated from imperfect parameter estimation using a single z-shim gradient. Additional z-shim gradients and echoes may be included to improve the estimation at the cost of an increased minimum TR.

Compensating in-plane field inhomogeneity may be necessary if large in-plane B_0 field inhomogeneity exists (41). In SWI, an anisotropic voxel with higher in-plane resolution is common, thereby making the impact of through-plane dephasing more significant than in-plane effects. Hence, our method focuses on through-plane compensation.

In our method, the brain was divided into two regions based on an empirically determined threshold in phase (-4π). Other approaches using magnitude images or local field information (13) may also be applied or combined with our approach to more accurately delineate the regions.

Conclusions

In conclusion, the proposed method reduces image artifacts in SWI, and thereby improves the spatial coverage. Compared to other SWI approaches suggested to compensate for field inhomogeneity artifacts, our method compensates not only phase artifacts but also magnitude artifacts. This technique may have important applications in detecting hemorrhage and cerebrovascular disease in the frontal or temporal lobe.

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Figure 1. Conventional SWI: (a) magnitude image, (b) phase mask (negative phase mask after homodyne filter of the original phase image), and (c) SWI image

Figure 2.

Proposed SWI sequence: a 3D GRE sequence with three echoes. A field compensation gradient (amplitude: G_c and duration: t_c) is applied before the third echo. Flow is compensated in the second echo.

Figure 3.

Images from each step of post-processing for phase. (a) Second echo phase image. (b) Phase image after homodyne filtering of the second echo phase signal. (c) the resulting phase mask (conventional SWI method). (d) Laplacian unwrapped phase image. (e) Phase image after removing the fitted phase signal by Gaussian modeling. (f) Homodyne filtered phase image. Red arrow indicates phase artifact from reduced magnitude signal. (g) Combined phase mask.

Figure 4.

Images from each post-processing step for magnitude (a) First echo magnitude image. (b) Second echo magnitude image. (c) Third echo magnitude image. (d) Magnitude image in Region 1. (e) Estimated magnitude image in Region 2. (f) Combined magnitude image.

Figure 5.

SWI and mIP images from three consecutive slices. SWI image (a) and mIP image (d) from the conventional method. Phase artifact corrected SWI image (b) and mIP image (e). The proposed phase post-processing method was used for the phase artifact correction. SWI image (c) and mIP image (f) generated by the proposed method. Additional veins are visible in the proposed method (red arrow)

Acquisition parameters**Table 1**

TR	50 ms
TE ₁ , TE ₂ , TE ₃	8.7, 25.0, 32.1 ms
Flip Angle	16°
Total scan time	8 min 48 sec
Matrix Size	220 × 220 × 48
FOV	220 × 220 × 57.6 mm ³
Resolution	1 × 1 × 1.2 mm ³
Bandwidth for 1 st and 3 rd echoes	455 Hz/pixel
Bandwidth for 2 nd echo	114 Hz/pixel
3 rd echo z-shim gradient	534 μT/m