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Improved MR Venography using Quantitative Susceptibility Weighted Imaging

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Abstract

Purpose—To remove the geometry dependence of phase based susceptibility weighting masks in SWI and to improve the visualization of the veins and microbleeds.

Materials and Methods—True susceptibility weighted images (tSWI) were generated using susceptibility based masks. Simulations were used to evaluate the influence of the characteristic parameters defining the mask. *In vivo* data from three healthy adult human volunteers were used to compare the contrast-to-noise-ratios (CNRs) of the right septal vein and the left cerebral vein as measured from both tSWI and SWI data. A traumatic brain injury (TBI) patient dataset was used to illustrate qualitatively the proper visualization of microbleeds using tSWI.

Results—Compared with conventional SWI, tSWI improved the CNR of the two selected veins by a factor of greater than three for datasets with isotropic resolution and greater than 30% for datasets with anisotropic resolution. Veins with different orientations can be properly enhanced in tSWI. Furthermore, the blooming artifact due to the strong dipolar phase of microbleeds in conventional SWI was reduced in tSWI.

Conclusion—The use of tSWI overcomes the geometric limitations of using phase and provides better visualization of the venous system, especially for data collected with isotropic resolution.

Keywords

susceptibility weighted imaging; quantitative susceptibility mapping; phase imaging

Introduction

Susceptibility weighted imaging (SWI) is a high resolution, spoiled gradient echo (GRE) magnetic resonance imaging (MRI) technique used today clinically for evaluating small veins and venous abnormalities in the brain and the presence of increased iron content and microbleeds in diseases such as dementia, multiple sclerosis, Parkinson's disease, stroke and

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traumatic brain injury (1–3). SWI's exquisite sensitivity to small tissue magnetic susceptibility changes is due to its use of phase information (3–5). Paramagnetic or diamagnetic substances relative to water such as blood products or calcium, respectively, perturb the local magnetic field proportional to their respective magnetic susceptibilities. These differences are reflected in the phase of the MR images. In conventional SWI, after applying appropriate unwrapping and/or filtering techniques on the raw phase data (1,3), a phase dependent mask is created and multiplied n times into the magnitude data to enhance the contrast/visibility of these substances.

Although SWI has been used quite successfully in clinical applications for many years, it is important to realize that it has a few weaknesses. One of them is based on the fact that the MRI phase signal is not only a function of the susceptibility, but also dependent on shape and orientation of the structure of interest. In data acquired with sufficient resolution, the phase inside veins perpendicular to the field has the opposite sign to that inside veins parallel to B_0 . This leads to variable suppression effects with the phase mask that makes SWI unique over conventional gradient echo imaging (6). Recently, quantitative susceptibility mapping (QSM) has emerged as a means to extract the source of phase information, that is, the local susceptibility distribution (7–15). QSM is known to be independent of echo time and, to a large degree, of orientation. To avoid the vessel orientation dependence in routine SWI data, instead of phase, we propose using a mask based on the susceptibility map. We refer to this approach as true-SWI (tSWI) to distinguish it from the conventional phase mask based SWI. In this work, our purpose is to compare the ability of these two methods to improve venous contrast and to show that tSWI is able to remove the geometry dependence of the phase for veins and microbleeds in SWI data.

Materials and Methods

To provide some flexibility in generating the susceptibility weighting mask W , we introduce both lower and upper thresholds for defining the mask as follows:

$$W = \begin{cases} 1 & \text{for } \chi \leq \chi_1 \\ 1 - \frac{\chi - \chi_1}{\chi_2 - \chi_1} & \text{for } \chi_1 < \chi \leq \chi_2 \\ 0 & \text{for } \chi > \chi_2 \end{cases} \quad [1]$$

where χ refers to the susceptibility value of a voxel (e.g., vein) relative to the surrounding tissue in the susceptibility map, χ_1 is the lower limit and χ_2 is the upper limit of the range of tissue susceptibility values for which we want to improve the contrast in the final susceptibility weighted image. Finally, the true-SWI is generated by multiplying the magnitude image with the mask n times similar to the usual SWI mask application:

$$tSWI = mag \cdot W^n \quad [2]$$

In order to determine appropriate values for these two thresholds, we examined different potential choices. For χ_1 , we used: 1) the mean susceptibility value in the background white matter tissue region (0 ppm) in the susceptibility map and 2) three standard deviations ($3\sigma_\chi$)

above the tissue region, where σ_χ is the standard deviation of the white matter tissue region in the susceptibility map. While a threshold of 0 ppm would ensure that the susceptibility weighting mask would include smaller veins that are partial volumed, it can also lead to increased noise in tissue regions where susceptibility is supposed to be zero. On the other hand, a choice of $\sigma_\chi = 3\sigma_\chi$ would reduce inclusion of noise in the mask. For χ_2 , we used: 1) the expected mean susceptibility value in the vein, which is about 0.45 parts per million (ppm) relative to water (3,16) under normal physiological conditions; and 2) a value higher than this, in this case 1ppm. For a given set of χ_1 and χ_2 values, the contrast-to-noise ratio (CNR) between the vein and the background tissue can be optimized by choosing an appropriate n value in Eq. 2. CNR for a vein can be defined as the ratio between tSWI signal contrast for the vein and its associated uncertainty as follows:

$$CNR = \frac{|S_{ref,tSWI} - S_{vein,tSWI}|}{\sigma_t} \quad [3]$$

Where $\sigma_t = \sqrt{\sigma_{ref,tSWI}^2 + \sigma_{vein,tSWI}^2}$. Noise in either the reference (background tissue) region ($\sigma_{ref,tSWI}$) or in the vein ($\sigma_{vein,tSWI}$) in the tSWI image can be estimated using the following equation:

$$\sigma_{tSWI} = \sqrt{W^{2n}\sigma_{mag}^2 + mag^2n^2W^{2n-2}\sigma_w^2} \quad [4]$$

The noise, represented by the standard deviation for a given voxel in W , σ_w , is dependent on the noise, σ_χ in the susceptibility map in the following manner:

$$\sigma_w = \begin{cases} \frac{\sigma_\chi}{\chi_2 - \chi_1} & \text{for } \chi_1 < \chi < \chi_2 \\ 0 & \text{for } \chi < \chi_1 \text{ or } \chi > \chi_2 \\ \frac{\sigma_\chi}{2(\chi_2 - \chi_1)} & \text{for } \chi = \chi_1 \text{ or } \chi = \chi_2 \end{cases} \quad [5]$$

The factor 1/2 in the case when $\chi = \chi_1$ or $\chi = \chi_2$ in Eq. 5 is due to the discontinuity (1). For simplicity, we assume that signal from vein (or cylinder) and reference region in the original magnitude images are the same: $mag_{ref} = mag_{vein} = 5$, and their associated signal standard deviations in these two regions are also the same: $\sigma_{ref,mag} = \sigma_{vein,mag} = \sigma$. Furthermore, we assume that the mean susceptibility value of the reference region is 0, and the mean susceptibility value of the vein is χ_v .

i) When a threshold of $\chi_1 = 0$ was used to generate the susceptibility mask, in the reference region, $W = 1$, and $\sigma_w = \sigma_\chi/(2\chi_2)$. In the vein, $W = 1 - \chi_v/\chi_2$, $\sigma_w = \sigma_\chi/\chi_2$, for $\chi_1 < \chi_v < \chi_2$; but when $\chi_v = \chi_2$, $W = 0$, $\sigma_w = \sigma_\chi/(2\chi_2)$; when $\chi_v > \chi_2$, $W = 0$, $\sigma_w = 0$. Using Eq. 4,

$$\sigma_t = \begin{cases} \left(\sigma^2 + \frac{S^2 n^2 \sigma_x^2}{4\chi_2^2} + \sigma^2 \left(1 - \frac{\chi_v}{\chi_2}\right)^{2n} + S^2 n^2 \left(1 - \frac{\chi_v}{\chi_2}\right)^{2n-2} \frac{\sigma_x^2}{\chi_2^2} \right)^{\frac{1}{2}} & \text{for } \chi_1 < \chi_v < \chi_2 \\ \left(\sigma^2 + \frac{S^2 \sigma_x^2}{4\chi_2^2} + \frac{S^2 \sigma_x^2}{4\chi_2^2} \right)^{\frac{1}{2}} & \text{for } \chi_v = \chi_2, n=1 \\ \left(\sigma^2 + \frac{S^2 n^2 \sigma_x^2}{4\chi_2^2} \right)^{\frac{1}{2}} & \text{for } \chi_v = \chi_2, n>1; \text{ or } \chi_v > \chi_2 \end{cases} \quad [6]$$

ii) When a threshold of $\chi_1 = 3\sigma_\chi$ was chosen, in the reference region, $W = 1$, and $\sigma_w = 0$ due to the fact that most pixels in the reference (background tissue) region have susceptibility values less than $3\sigma_\chi$. In the vein, $W = 1 - (\chi_v - 3\sigma_\chi)/(\chi_2 - 3\sigma_\chi)$, $\sigma_w = \sigma_\chi/(\chi_2 - 3\sigma_\chi)$, for $\chi_1 < \chi_v < \chi_2$; but when $\chi_v = \chi_2$, $W = 0$, $\sigma_w = \sigma_\chi/2(\chi_2 - 3\sigma_\chi)$; when $\chi_v > \chi_2$, $W = 0$, $\sigma_w = 0$. Thus,

$$\sigma_t = \begin{cases} \left(\sigma^2 + \sigma^2 \left(1 - \frac{\chi_v - 3\sigma_\chi}{\chi_2 - 3\sigma_\chi}\right)^{2n} + S^2 n^2 \left(1 - \frac{\chi_v - 3\sigma_\chi}{\chi_2 - 3\sigma_\chi}\right)^{2n-2} \frac{\sigma_x^2}{(\chi_2 - 3\sigma_\chi)^2} \right)^{\frac{1}{2}}, & \text{for } \chi_1 < \chi_v < \chi_2 \\ \left(\sigma^2 + S^2 n^2 \frac{\sigma_x^2}{4(\chi_2 - 3\sigma_\chi)^2} \right)^{\frac{1}{2}}, & \text{for } \chi_v = \chi_2, n=1 \\ \sigma, & \text{for } \chi_v = \chi_2, n>1; \text{ or } \chi_v > \chi_2 \end{cases} \quad [7]$$

Thus the CNR can be written as:

$$CNR = S \left(1 - \left(1 - \frac{\chi_v - \chi_1}{\chi_2 - \chi_1} \right)^n \right) / \sigma_t \quad [8]$$

for $\chi_1 < \chi_v < \chi_2$, with σ_t given in Eqs. 6 and 7.

When $\chi_1 = 3\sigma_\chi$ and χ_v is slightly less than χ_2 , CNR approaches SNR in the magnitude images as n approaches infinity. In this case, the optimal n was chosen to be the value where CNR reaches 90% of the maximal CNR for a certain vein.

Simulations

To evaluate the theoretical predictions, the optimal choice of n for generating tSWI images for different threshold values and vessel susceptibility values, and the influence of high-pass filtering on the final CNR for veins in tSWI images, simulations were performed using cylinders as surrogates to veins. A series of cylinders with radii ranging from 2 pixels to 16 pixels was used to simulate the associated phase images in a 512×512 matrix at $B_0 = 3T$, and $TE = 10ms$. The cylinders were taken to be perpendicular to the main magnetic field. The input susceptibility of the cylinders was set to be 0.45ppm and the susceptibility value of the background region was set to zero. In order to simulate a more realistic response of the field perturbation, the complex data of a cylinder with radius 16 times of the final radius was first created on an 8192×8192 matrix. The magnitude signal for the cylinder and background region were taken to be unity. The central 512×512 k-space points generated from the larger matrix were then used to reconstruct the complex images of the cylinders. Gaussian noise was added to both real and imaginary channels of the data to simulate the SNR in the magnitude images to be 10:1. The simulated phase images were processed using a homodyne high-pass filter with a k-space window size of 64×64 . Two sets of susceptibility

maps, one from unfiltered and the other from filtered phase images, were generated for each cylinder size, using truncated k-space division with a k-space threshold of 0.1 (7). This is to evaluate the influence of high-pass filtering on the final CNR in tSWI images. The tSWI images were generated using Eqs. 1 and 2 for different values of χ_1 and χ_2 as mentioned in the previous section. The standard deviation of the susceptibility maps was measured from a reference region outside the cylinder. The susceptibility mask was multiplied into the magnitude image n times with n ranging from 0 to 10 ($n=0$ refers to the case of no mask multiplication). The local CNRs between cylinders and the background reference were measured from the tSWI data using:

$$CNR_{measured} = |S_{vein} - S_{ref}| / \sigma_t \quad [9]$$

where S_{vein} and S_{ref} are the mean intensity values inside the cylinder (vein) and inside a reference region of interest (ROI) adjacent to the cylinder directly from tSWI image, respectively. In order to estimate the overall noise σ_t directly from tSWI images, the standard deviations inside the cylinder ($\sigma_{vein,tSWI}$) and the reference region ($\sigma_{ref,tSWI}$) in tSWI were measured and σ_t was again calculated as the square root of $\sigma_{vein,tSWI}^2 + \sigma_{ref,tSWI}^2$. The theoretically predicted CNRs from Eq. 8 using different thresholds for generating the susceptibility mask were compared with those measured from the simulations and the appropriate value for χ_2 for processing *in vivo* data was determined. CNRs of the cylinders with different susceptibility values ranging from 0.2ppm to 0.45ppm were calculated to evaluate the influence of the susceptibility value of the object on the optimal choice of n .

In vivo data

To evaluate the efficacy of tSWI in *in vivo* neuro-imaging, we compared the CNR obtained in tSWI data with that obtained in conventional SWI images in three healthy adult volunteers. The study was approved by the local institutional review board and informed consent was obtained from all subjects before the MRI scan. The volunteers were imaged on a 3T Verio system (Siemens, Erlangen) using a 3D SWI sequence with isotropic voxel size of $0.5\text{mm} \times 0.5\text{mm} \times 0.5\text{mm}$. Imaging parameters are given in Table 1. Data were acquired in the transverse orientation. In one case (volunteer 1), the SWI sequence was performed twice using two different echo times ($TE = 14.3\text{ms}$ and 17.3ms). To evaluate the influence of voxel aspect ratio on the CNR, lower resolution images of $0.5\text{mm} \times 0.5\text{mm} \times 2\text{mm}$ (anisotropic voxel size) from all 4 volunteer datasets were generated by taking the central portion of the original k-space along the transverse direction.

The quantitative susceptibility maps were generated for the isotropic and the anisotropic data, as follows: tissues outside the brain were removed using the Brain Extraction Tool (BET) in FSL (17), a homodyne high-pass filter with a k-space window of 64×64 was applied to remove the background field induced phase artifacts (1), and the inversion process to create susceptibility maps was accomplished using a single orientation truncated k-space division approach with a k-space truncation threshold of 0.1 (7). Similar to the case of the simulated data, two sets of tSWI images for each of the SWI datasets were generated using Eqs. 1 and 2 with: a) $\chi_1=0$ and b) $\chi_1 = 3\sigma_\chi$, where σ_χ is the standard deviation of the

susceptibility in the background white matter region close to the vein for which the CNR was measured. The threshold χ_2 was kept at 0.45ppm. The susceptibility mask was multiplied into the magnitude image n times, with n ranging from 1 to 10. To generate conventional SWI images, phase masks were created using the high-pass filtered phase images and then multiplied four times into the magnitude images (1). To investigate the impact of newer data processing methods, we also applied phase unwrapping (18) and SHARP (9) to remove the background field, and applied a geometry constrained iterative algorithm (8) to reconstruct the susceptibility maps for one dataset (Dataset 2). Then, tSWI images were generated using $\chi_1=0$ and $\chi_2=0.45$ ppm. Local CNRs of two selected veins, the left internal cerebral vein (LICV) and the right septal vein (RSV), were measured from both tSWI and SWI using Eq. 9. Each vessel's ROI was selected from the susceptibility maps and copied onto the tSWI or SWI images for CNR evaluation. The reference ROI adjacent to each vein was taken from the same slice. To demonstrate the advantages of using tSWI over conventional SWI, we also analyzed one SWI dataset from a TBI patient. For this patient dataset, susceptibility maps were generated through homodyne high-pass filtering and truncated k-space division. tSWI images were then obtained with susceptibility weighting masks generated using $\chi_1=0$, $\chi_2=0.45$ ppm and $n=2$. All the processing was done using Matlab (R2010a, Natick, MA).

Results

Simulations

The simulated phase images of cylinders of different sizes, their corresponding susceptibility maps and tSWI images are shown in Fig. 1. The measured CNRs for cylinders with different radii, but with a constant input susceptibility of 0.45ppm, are plotted as a function of n in Fig. 2, and the theoretically predicted CNRs are plotted in Fig. 3. Since no T2* effects are considered here, the CNRs shown in Figs. 2 and 3 reflect contrasts from only phase/susceptibility differences between the cylinders and the background reference region. The optimal choice of n and, correspondingly, the value of CNR in the tSWI image, are influenced both by (a) the choices of χ_1 and χ_2 and (b) the high pass filter. For $\chi_1=0$, CNRs reach maximum when $n=4$ (Figs. 2a, 2c, 2e, 3a and 3c). When χ_2 is larger with $\chi_1=0$, it also takes a larger n value to reach the optimized CNR. Meanwhile, the choices of χ_1 and χ_2 can also affect the rate at which optimal CNR is approached as a function of n . For $\chi_1=3\sigma_\chi$, CNRs in general increase as n increases (Figs. 2b, 2d, 2f, 3b and 3d). The optimal n may be chosen when CNR reaches 90% of the maximum of CNR. The optimal n was 4 for $\chi_2=0.45$ ppm, and greater than 10 for $\chi_2=1$ ppm.

When a high-pass filtered phase was used, the optimal choice of n was not affected much for $\chi_1=0$ and $\chi_2=0.45$ ppm (Fig. 2e), but was slightly bigger for $\chi_1=3\sigma_\chi$ and $\chi_2=0.45$ ppm for all the cylinders except for the smallest one (Fig. 2f). For both $\chi_1=0$ and $\chi_1=3\sigma_\chi$, the maximal CNR was reduced for bigger cylinders, when the high-pass filter was used. This is also partly evident in Fig. 1. The behavior in the case when the high pass filtered phase was used, agreed with the pattern observed in the theoretically predicted CNRs for objects with low susceptibility in Fig. 3. Given the simulated results shown in Fig. 2, we chose $\chi_2=0.45$ ppm in the *in vivo* studies for a consistent choice of n for the maximal CNR.

In vivo tSWI

The local CNRs of the two veins, 1) the right septal vein and 2) the left internal cerebral vein, normalized by the SNRs, are plotted in Figs. 4 and 5 for all the *in vivo* datasets. The normalized CNRs in the SWI images and in the original magnitude images, as well as the SNRs in the original magnitude images (from the background tissues) are shown in Table 2. Compared to the original magnitude images, SWI improves the local CNR of the right septal vein in the anisotropic data, but not in the isotropic data. The CNR of the left internal cerebral vein is not improved in SWI in either isotropic or anisotropic data, due to the amplification of the noise in the background tissue region. Compared with the CNRs in magnitude images, the local CNRs in tSWI were improved by roughly a factor of 2 in both isotropic and anisotropic cases for $\chi_1 = 0$. Compared with conventional SWI, the CNRs were improved by a factor of greater than three for datasets with isotropic resolution and greater than 30% for datasets with anisotropic resolution in tSWI. The local CNRs were further improved when $\chi_1 = 3\sigma_\chi$. Considering all cases, when $\chi_1 = 0$ was used, $n = 2$ was a reasonable practical choice for both isotropic and anisotropic datasets; when $\chi_1 = 3\sigma_\chi$ was used, $n = 4$ was a reasonable choice for isotropic datasets and $n = 8$ for anisotropic datasets. When SHARP along with an iterative algorithm (8) was used to generate susceptibility maps, the CNRs of the two veins of interest in the corresponding tSWI image were improved, as shown in Figs. 4c, 4d, 5c and 5d. For isotropic resolution, at $n=2$ with $\chi_1 = 0$, the relative improvements for the left internal cerebral vein and the right septal vein were 23% and 14%, respectively. For anisotropic resolution and at $n=2$, the relative improvements for the two veins were less than 5%. However, this improvement in CNR was more significant for grey matter structures than for veins, as can be seen from Figs. 6d and 6h.

In Fig. 6, we compare the tSWI and SWI minimal intensity projections for both the isotropic and anisotropic cases. The tSWI appears to have higher CNR than the conventional SWI in both isotropic and anisotropic data. For tSWI, isotropic data provided a better delineation of the venous structures, compared to the anisotropic data. This is consistent with the results shown in Figs. 4 and 5, in which the normalized maximal CNRs are higher for the isotropic data than those for the anisotropic data. When $\chi_1 = 3\sigma_\chi$ was used, the visibility of some tiny veins and the grey matter structures was reduced compared to the case when $\chi_1 = 0$ was used.

As an example of this process with $\chi_1 = 0$, $\chi_2 = 0.45\text{ppm}$ and $n = 2$, Fig. 7 shows a case that demonstrates the problems with the conventional SWI processing: one of the veins has a trajectory roughly at the magic angle (54.7°) with respect to the direction of the main magnetic field B_0 . The black arrow shows the vein which is clearly seen in the tSWI (Fig. 7f). In SWI (Fig. 7e), the vein actually shows a dark structure which is in fact more associated with its edges. This makes the veins appear much bigger in the SWI than in the tSWI data, as can be seen from the minimal intensity projections (mIPs) in Figs. 7g and 7h. This non-local phase information used in SWI can lead to an inaccurate estimation of the geometry of microbleeds, as demonstrated in Fig. 8. In this TBI case, tSWI has more faithfully represented the microbleeds.

Discussion

Quantitative susceptibility mapping offers an additional means to recognize veins and microbleeds and other tissues with high iron content as phase imaging does. However, the phase images are dependent on each object's shape and orientation while the susceptibility values of the structures are not, at least in principle. Therefore, to produce better susceptibility weighted images, we have investigated the use of susceptibility maps for the masking process.

There are a number of key observations that can be made from the data presented herein. First, the results presented in this paper demonstrate that the object shape and orientation can be reasonably accounted for by using susceptibility maps and, hence, the inability of SWI processing to enhance veins at different orientations can be overcome. Besides, the blooming artifact due to the dipolar phase of microbleeds in conventional SWI was avoided. This leads to potential applications of this technique, for example, the evaluation of microbleeds in TBI studies. Second, tSWI can be used to process isotropic data, whereas SWI processing has relied on anisotropic data for its best results due to the direct use of phase information which is orientation dependent (6). In the past, SWI data have usually been collected with anisotropic resolution with 2mm slice thickness (6). However, modern segmented echo planar approaches are becoming viable and one expects to see more data being collected with isotropic voxel sizes (19,20). The high isotropic resolution also helps to reduce the error caused by the partial voluming effects in susceptibility quantification, and thus leads to improved quality for tSWI. Note that, high image resolution will also lead to lower SNR within a given image time and thus lower CNR. Generally speaking, tSWI is most advantageous with the isotropic datasets. Third, the use of a susceptibility mask is not restricted to the paramagnetic venous blood, but it could also be designed to study the diamagnetic materials (e.g., calcifications, which have negative susceptibility in the susceptibility maps). Fourth, the effects of the upper and lower thresholds used in creating the susceptibility masks have been studied for two reasonable values, and the optimal number of multiplications, n has been determined. When the lower threshold was set to zero, the fact that a continuous mask from zero to unity is generated makes it possible to enhance contrast even in smaller veins, or larger veins that have had their phase artificially suppressed by using the high-pass filter, or in structures that have lower iron contents. The use of $\chi_1 = 3\sigma_\chi$ helps to avoid amplifying noise in regions of low susceptibility and hence leads to a higher CNR. However, at the same time it can prevent small veins or structures with very low susceptibility from being enhanced. In addition, different datasets require different optimal n values. To avoid this problem, it may be more practical to choose χ_1 to be 0. We choose the upper threshold χ_2 to be 0.45ppm, as it corresponds to the theoretical susceptibility of venous blood when the oxygen saturation is 70% and the hematocrit is 45%. Increasing this upper threshold may lead to a slightly larger value for the optimal n when the susceptibility value of the vein is much smaller than χ_2 . In most cases, $n = 2$ gave optimal results, for $\chi_1 = 0$ and $\chi_2 = 0.45$ ppm. In order to capture smaller veins or structures with lower susceptibility values such as the basal ganglia, a slightly larger n can be used for either isotropic or anisotropic datasets. The lower susceptibility values are due to a combined effect of partial voluming and high-pass filtering. Fifth, the predicted CNRs

slightly deviate from measured CNRs in simulations, as the mean value of the susceptibility mask W in the background reference region is slightly less than 1. As a result, this creates slight differences between the prediction and measurements at large n values ($n > 4$). Lastly, conventional SWI uses phase information which is dependent on echo time and usually a relatively long echo time is used in SWI data collection. Although the phase mask could be redefined as a function of echo time to accommodate the loss of phase information as echo times are reduced, no such modification needs to take place for tSWI since the susceptibility map does not change with echo time. However when echo times are reduced the phase SNR used to generate the susceptibility map will decrease. On the other hand, if echo times become too long, phase aliasing occurs and the apparent size of the vessel will increase. Thus, tSWI makes the use of short echo times possible as long as the SNR is high enough to create a reasonable estimate of the local susceptibilities. The selection of a shorter TE has several major advantages, including reducing background field induced phase artifacts, shorter scan time and better overall image quality.

There are several limitations to this method. First, we are using susceptibility maps generated from a single orientation dataset to create the mask for tSWI. These susceptibility maps can have streaking artifacts which are caused by the singularities in the inverse kernel (7–15). The streaking artifacts could permeate the tSWI data causing artifacts which did not exist before or decrease the CNR of grey matter structures. Some newer techniques such as nonlinear regularization (10,13,14) and iterative algorithms (8) will reduce the streaking artifacts and the latter is particularly time-efficient. Another common problem of the single orientation QSM method is the systematic under-estimation or bias of the susceptibility. However, this can be compensated by the thresholds used to generate the tSWI weighting masks. Second, we used the traditional homodyne high-pass filter to remove the background phase artifacts in the *in vivo* data. Even though homodyne high-pass filtering could be applied without phase unwrapping, it leads to an underestimation of the susceptibility, especially for large objects. This can be improved by using newly developed background field removal methods (9,21). But for relatively small structures such as veins, homodyne high-pass filtering already gives satisfying results. Given the fact that homodyne high-pass filtering is still being widely used, the proposed algorithm can be directly added to the current SWI data processing scheme.

In conclusion, we have proposed a data processing scheme which we refer to as true SWI or tSWI to generate SWI like images using susceptibility maps. This helps to avoid the orientation dependence related problem in SWI, especially in data with isotropic resolution and, in the future, possibly to allow the use of short TE SWI data collection. This tSWI data provide better and more consistent visualization of the venous system and thus have potential clinical applications in the study of neurodegenerative diseases.

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Figure 1.

Phase images (*a* and *b*), susceptibility maps (*c* and *d*) and tSWI images (*e*, *f*, *g* and *h*) for simulated cylinders with and without homodyne high-pass filtering. Images in the first and third columns are generated using the original phase images without high-pass filtering, while images in the second and fourth columns are generated using high-pass filtering. The tSWI images *e* and *f* were generated using $\chi_1=0$, $\chi_2=0.45\text{ppm}$, $n=2$; while *g* and *h* were generated using $\chi_1=3\sigma_\chi$, $\chi_2=0.45\text{ppm}$, $n=4$. σ_χ is the standard deviation of a reference region measured from the susceptibility maps shown in *c* and *d* ($\sigma_\chi=0.05\text{ppm}$ for both *c* and *d*). The SNR in the original magnitude image was set to be 10:1 and the CNR between the cylinders and background in the original magnitude images was basically zero.

Figure 2.

Measured CNRs of cylinders from simulated tSWI images. Figures in different rows were generated using different χ_2 values, while figures in different columns were generated using different χ_1 values. **a)** $\chi_1=0$, $\chi_2=1\text{ppm}$; **b)** $\chi_1=3\sigma_\chi$, $\chi_2=1\text{ppm}$; **c)** $\chi_1=0$, $\chi_2=0.45\text{ppm}$; **d)** $\chi_1=3\sigma_\chi$, $\chi_2=0.45\text{ppm}$; **e)** $\chi_1=0$, $\chi_2=0.45\text{ppm}$; and **f)** $\chi_1=3\sigma_\chi$, $\chi_2=0.45\text{ppm}$. To evaluate the effect of high-pass filtering, **e** and **f** were generated using high-pass filtered phase images.

Figure 3.

Theoretically predicted CNRs of cylinders with different susceptibility values. Figures in different rows were generated using different χ_2 values, while figures in different columns were generated using different χ_1 values. **a)** $\chi_1=0, \chi_2=1\text{ppm}$; **b)** $\chi_1=3\sigma_\chi, \chi_2=1\text{ppm}$; **c)** $\chi_1=0, \chi_2=0.45\text{ppm}$; and **d)** $\chi_1=3\sigma_\chi, \chi_2=0.45\text{ppm}$.

Figure 4.

Local CNRs of the right septal vein (**a**, **c**, and **e**) and the left internal cerebral vein (**b**, **d**, and **f**) from different datasets with isotropic resolution. **a**, **b**, **c** and **d** were generated when threshold $\chi_1=0$ was used to create the susceptibility weighting masks, while **e** and **f** were generated when $\chi_1=3\sigma_\chi$ was used. The CNRs were normalized by the corresponding SNRs listed in Table 2. **c** and **d** show the CNRs of the two veins in Dataset 2 with isotropic resolution, when different data processing methods were used for susceptibility mapping (see Figure 6 for examples of the tSWI images).

Figure 5.

Local CNRs of the right septal vein (**a**, **c**, and **e**) and the left internal cerebral vein (**b**, **d**, and **f**) from different datasets with anisotropic resolution. **a**, **b**, **c** and **d** were generated when threshold $\chi_1=0$ was used to create the susceptibility weighting masks, while **e** and **f** were generated when $\chi_1=3\sigma_\chi$ was used. The CNRs were normalized by the corresponding SNRs listed in Table 2. **c** and **d** show the CNRs of the two veins in Dataset 2 with anisotropic resolution, when different data processing methods were used for susceptibility mapping (see Figure 6 for examples of the tSWI images).

Figure 6.

Comparison between minimal intensity projections (mIP) of tSWI and SWI data over 16mm for isotropic (top row) and anisotropic data (bottom row) for Dataset 2. For **b**, **c**, **f** and **g**, susceptibility maps were generated using homodyne high-pass filtering and thresholded k-space division; while for **d** and **h**, susceptibility maps were generated using SHARP and geometry constrained iterative algorithm. **a**) isotropic SWI mIP; **b**) isotropic tSWI mIP ($\chi_1=0$, $\chi_2=0.45\text{ppm}$, $n=2$); **c**) isotropic tSWI mIP ($\chi_1=3\sigma_\chi$, $\chi_2=0.45\text{ppm}$, $n=4$); **d**) isotropic tSWI mIP ($\chi_1=0$, $\chi_2=0.45\text{ppm}$, $n=2$); **e**) anisotropic SWI mIP. **f**) anisotropic tSWI mIP ($\chi_1=0$, $\chi_2=0.45\text{ppm}$, $n=2$). **g**) anisotropic tSWI mIP ($\chi_1=3\sigma_\chi$, $\chi_2=0.45\text{ppm}$, $n=8$). **h**) anisotropic tSWI mIP ($\chi_1=0$, $\chi_2=0.45\text{ppm}$, $n=2$).

Figure 7.

A sagittal view showing a vein near the magic angle (54.7° relative to the main magnetic field) as indicated by the black arrows. **a)** Phase image (from a left-handed system) showing effectively zero phase inside the vein, with outer field dipole effects also visible; **b)** susceptibility maps showing the vein as uniformly bright; **c)** susceptibility weighting mask obtained from the phase image ($n=4$); **d)** susceptibility weighting mask obtained from the susceptibility maps ($\chi_1=0$, $\chi_2=0.45\text{ppm}$, $n=2$); **e)** SWI showing unsuppressed signal inside the vein; and **f)** tSWI showing a clear suppression of the vein even at the magic angle. **g)** mIP of SWI in the sagittal direction. **h)** mIP of tSWI in the sagittal direction. Note the vessels near the magic angle are now well delineated in the tSWI data.

Figure 8.

Sagittal views of SWI (*a* and *c*) and tSWI images (*b* and *d*) in a TBI case. The microbleeds appear much bigger on the SWI images than on the tSWI images, as indicated by the white arrows. This is due to the non-local phase information used in the conventional SWI weighting mask. For better visualization, the images were interpolated in through-plane direction from a resolution of $0.5\text{mm} \times 0.5\text{mm} \times 2\text{mm}$ to 0.5mm isotropic resolution.

Imaging parameters for three volunteers and one patient for *in vivo* studies. Dataset 5 was collected on a TBI patient.

Dataset No.	1	2	3	4	5
Volunteer No.	1	1	2	3	-
B ₀ (T)	3	3	3	3	3
TR (ms)	26	26	24	24	29
TE (ms)	14.3	17.3	17	15.3	20
FA (degrees)	15	15	15	12	15
BW (Hz/px)	121	121	181	121	120
Voxel size (mm ³)	0.5×0.5×0.5	0.5×0.5×0.5	0.5×0.5×0.5	0.5×0.5×0.5	0.5×0.5×2
Matrix Size	512×368×256	512×368×256	512×368×224	512×368×192	512×416×64

Local CNRs of the two selected veins on tSWI images, SWI images and the original magnitude images, as well as the SNRs in the original magnitude images from different datasets. The CNRs were normalized by the corresponding SNRs. The SNRs were measured from the reference regions close to the right septal vein in the magnitude images. The tSWI images were generated using $\chi_1=0$, $\chi_2=0.45\text{ppm}$, $n=2$.

Table 2

	Isotropic case			Anisotropic case		
	Vein 1	Vein 2	Vein 1	Vein 2	Vein 1	Vein 2
Dataset 1	tSWI	0.6	0.5	0.4	0.4	0.4
	CNR	SWI	0.2	0.1	0.3	0.1
		Mag	0.2	0.1	0.2	0.2
Dataset 2	SNR	10.1		18.8		
		tSWI	0.7	0.5	0.5	0.4
	CNR	SWI	0.1	0.1	0.3	0.2
Dataset 3		Mag	0.2	0.2	0.2	0.2
	SNR	9.7		17.0		
		tSWI	0.7	0.6	0.4	0.5
Dataset 4	CNR	SWI	0.1	0.2	0.3	0.2
		Mag	0.2	0.3	0.2	0.2
	SNR	9.1		17.0		
Dataset 5		tSWI	0.6	0.5	0.4	0.4
	CNR	SWI	0.2	0.1	0.3	0.1
		Mag	0.3	0.2	0.2	0.1
Dataset 6	SNR	9.4		18.6		