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Partially-parallel, susceptibility-weighted MR imaging of brain vasculature at 7 Tesla using sensitivity encoding and an autocalibrating parallel technique

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Abstract—**Susceptibility-weighted magnetic resonance imaging is a powerful tool for high resolution imaging of the vasculature, aiding in the diagnosis of many pathologic conditions. The technique is especially beneficial at higher field strengths where traditional sequences that measure cerebral blood volume suffer from severe distortions, rendering them inapplicable at 7T. However, conventional susceptibility-weighted imaging (SWI) sequences involve long scan times, on the order of 10 minutes for a 2 cm slab of coverage. This work implemented two partially parallel imaging reconstruction methods, 1) an autocalibrating parallel technique based on GRAPPA algorithm, and 2) sensitivity encoding (SENSE) for accelerating SWI of the brain at 7 Tesla. By employing two-fold under-sampling in the phase-encoding direction for both techniques, a two-fold reduction in scan time was simulated. Analysis of contrast ratios in large and small vessels compared to the surrounding brain parenchyma showed close agreement between the full and GRAPPA reconstructed datasets for both vessel sizes, while a decrease in the small vessel contrast was observed with SENSE.**

I. Introduction

Susceptibility weighted imaging (SWI) in an emergent technique for high resolution, distortion-free imaging of brain vasculature that is recently gaining importance in the clinical setting. The weighting relies on changes in phase that result from signal loss due to partial volume effects near venous vessels, typically requiring long echo times to obtain sufficient weighting [1]. This phase-sensitive imaging technique has recently been shown to improve the diagnosis of brain neoplasms, neurological trauma, and vascular malformations, in addition to a variety of cerebrovascular and neurodegenerative diseases [2]. As high field MR systems with multi-channel coil capabilities become readily available for routine clinical use, traditional techniques to measure relative cerebral blood volume in brain tumors such as dynamic susceptibility-contrast perfusion MRI are challenging due to increased B_0 inhomogeneity and differences in magnetic susceptibility at air-tissue interfaces that lead to signal drop out and large geometric distortions in echo planar imaging. An alternative approach is to use these susceptibility effects through the phase information contained in conventional gradient-echo sequences to create high resolution susceptibility-weighted venograms, which take advantage of the heightened susceptibility-contrast inherent with increased field strength [1].

In order to achieve sufficient susceptibility contrast from vessels, while simultaneously minimizing contrast among gray matter, white matter, and ventricles, long echo and repetition times are required. Although the echo time needed to visualize the large phase changes in venous vessels due to the magnetic susceptibility differences between oxygenated and deoxygenated blood is reduced with increasing field strength due to decreased T2* relaxation, long acquisition times are still compulsory to avoid T1-weighting because of longer T1 relaxation times. As a result, the total acquisition time for SWI remains long at higher field strengths (>10 minutes for only a 2 cm slab of tissue at 7T), which can result in patient discomfort and motion induced artifacts. Thus, the need for faster acquisition times and efficient ways of combining multi-channel coil data without losing the phase information as a result of the reconstruction becomes apparent. Partially parallel imaging (PPI) acquisitions and reconstruction algorithms can be employed to speed up these long acquisition times as long as the concomitant decrease in SNR does not significantly affect the contrast between vessels and brain parenchyma [1,3,4].

In PPI the MR signal is received by an array of coils and the measurement time is reduced by undersampling the signal in the Fourier or k-space domain, in the phase-encoding direction. Undersampling by a factor of R accelerates acquisition time by R-fold, with the tradeoff of reduced signal-to-noise ratio (SNR). The reduction in the sampling density in k-space is equivalent to reduction of the image field-of-view (FOV) by R times. The missing data points are synthesized post-acquisition by incorporation of the spatial information from the individual coil array elements. The "unaliasing" or parallel reconstruction can be performed either in the spatial or Fourier domain, by direct inversion or indirect reconstruction techniques. Direct techniques like Sensitivity Encoding (SENSE) [5] form the encoding or reconstruction matrix by estimating the localized sensitivity of each coil element from a low resolution coil calibration scan and solve the reconstruction by direct inversion of the encoding matrix. Indirect techniques, such as Generalized Autocalibrating Partially

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Parallel Acquisitions (GRAPPA) [6], acquire an additional small set of phase-encoding lines at the Nyquist sampling frequency from the center of k-space that act as training lines for estimating the interpolation weights and then use them to synthesize the skipped phase encoding lines from the acquired lines. Since the coil calibration is built into the actual acquisition in these methods, they are also known as autocalibrating (AC) techniques.

The feasibility of several PPI techniques has been proposed in the literature for accelerating SWI acquisitions at varying field strengths. Sedlacik et al. initially simulated elliptical and GRAPPA k-space undersampling and reconstruction regimes with factor of two reductions for SWI at 1.5T and found reduced contrast of small vessels compared to the fully sampled case [4]. However, no noticeable difference in the contrast resolution of small vessels was observed when we applied SENSE to SWI with the same reduction factor at 3T [7]. Both the gain in SNR and increased susceptibility contrast achievable at higher field strengths should facilitate the implementation of PPI techniques for SWI. At 7T, Banerjee et al. demonstrated the ability to implement an autocalibrating PPI technique for SWI, but quantitative analysis of vessel contrast was not assessed [8]. The goal of this paper is to assess the effects of both GRAPPA and SENSE based reconstructions on vessel contrast for accelerating susceptibility-weighted imaging of the brain at 7 Tesla.

II. Methodology

A. Data acquisition

High resolution T2*-weighted brain MR imaging was performed on six healthy volunteers with a 7T GE Signa scanner (GE Healthcare, Milwaukee, WI) using uniform excitation by a volume transmitter and reception by an eight channel phased-array head coil (Nova Medical, Wilmington, MA). Informed consent was obtained from the volunteers prior to scanning. The susceptibility weighted imaging employed a 3D flow compensated, GRE sequence with TE/TR=16/80ms, flip angle=20^o, BW=62.5 kHz, a 24x24x2.8 cm^3 FOV, and a 512x256x28 image matrix. The total data acquisition time was 11 minutes. The imaging protocol also included the acquisition of a low resolution, proton-density weighted, fast gradient echo sequence (TE/TR=2.1/150 ms, flip angle=20°, a 30x30 cm^2 FOV, 64x64 image matrix, and 3 mm slice thickness) for coil sensitivity estimation.

B. SWI post-processing of magnitude and phase data

The complex data from all coils was transferred off-line to a Sun workstation and post-processing was performed using Matlab 6.5 (The Mathworks Inc., Natick, MA) on a Linux cluster. For the conventional full FOV multi-channel acquisitions, phase masks were constructed from the raw complex data of each individual coil element through complex division by a low-pass filtered image to remove the effects of field inhomogeneity, and scaling the resulting negative phase values between zero and one [1], as illustrated in Figure 1. The phase masks were then multiplied into the magnitude image from each coil four times and the resulting susceptibility-weighted images were combined by the traditional square root of sum of squares method. The sum of squares method was preferred to summing the images weighted by coil sensitivity profiles because the weighted sum of coil sensitivities gave decreased uniformity in the combined images. A low pass filter correction with edge completion algorithm [9] was then applied to the combined images to remove any residual intensity variation across the image. Minimum intensity projections (mIPs) through 15 mm thick slabs of similar anatomical locations were generated to obtain the final SWI images used in the subsequent analysis.

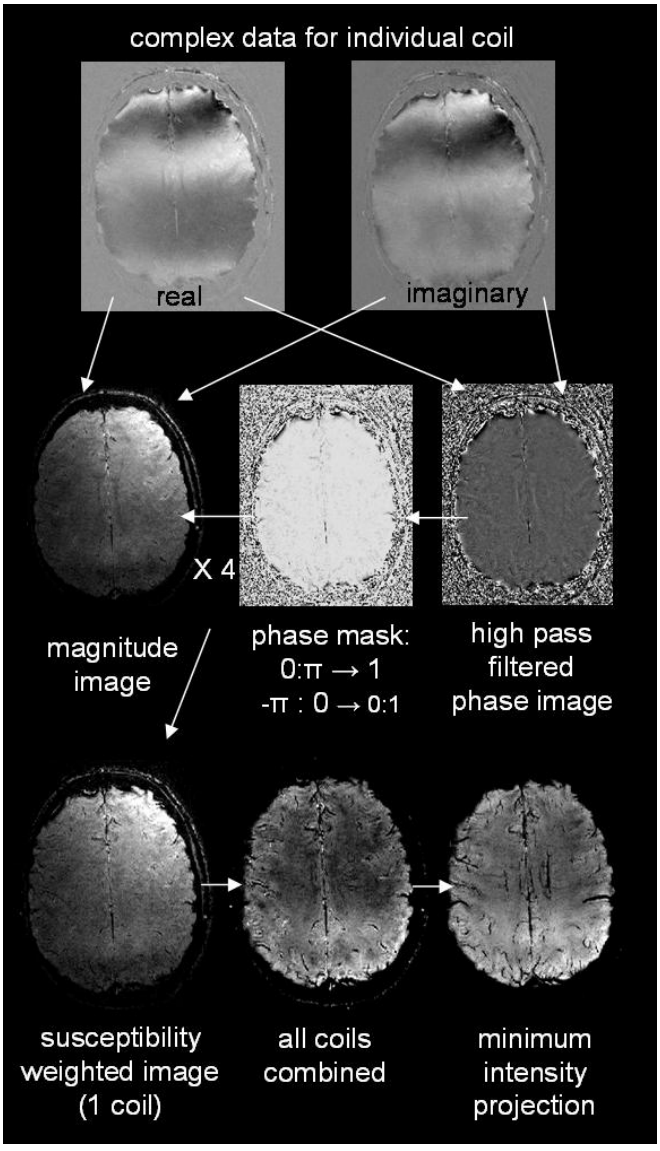

Figure 1: Depiction of processing steps involved in multi-channel SWI.

C. GRAPPA based reconstruction of SWI

GRAPPA is a robust parallel reconstruction technique operating in the Fourier domain that reconstructs the full FOV data from each individual coil allowing subsequent array combination. It employs a block-wise reconstruction in which 1 block consists of 1 acquired line and R-1 skipped lines for a reduction factor of R. The interpolation weights for an individual coil are obtained by least square fitting of the acquired lines from all the coils to the AC lines of that coil. Several variations of the data fitting described above have been presented in the literature, such as performing the fitting piecewise for segments along the unaccelerated direction. Wang et al proposed a floating node fitting (FNF) which allows additional data fits compared to conventional GRAPPA during the calibration. The authors also proposed a multi-column multi line interpolation (MCMLI) that uses the nearest neighboring points in the unaccelerated direction as well for synthesis of missing lines [10].

The full FOV complex k-space data from each coil element was decimated in the phase encoding direction to simulate a reduction factor of 2, while retaining 16 central AC lines for the FNF with MCMLI algorithm employed to synthesize the missing lines of k-space. The full FOV complex image data for each coil element generated by the GRAPPA reconstruction then underwent the same post-processing method as the initial full FOV dataset to create magnitude susceptibility-weighted images for each coil. The SWI images were then combined by taking the square root of the sum of squares, low pass filter corrected, and projected.

D. SENSE based reconstruction of SWI

In SENSE reconstruction, an aliased or reduced FOV image is first obtained for each coil array element. The construction of the combined full FOV image is based on the fact that each pixel in the reduced FOV includes the signal contributions from a number of positions in the full FOV. Separation of superimposed pixels is then achieved using the local coil sensitivities to determine different weights needed to unfold the signal superposition of each coil image.

SENSE reconstruction with a reduction factor of 2 was simulated to generate complex reduced FOV images. The complex coil sensitivity images for each coil element were up-sampled to match the high resolution dataset and normalized by the combined coil sensitivity map to remove any additional weighting from the underlying brain anatomy. The unfolding of superimposed pixels was accomplished by direct inversion of an encoding matrix created from the sensitivity profiles. Susceptibility weighting was then performed on the combined full FOV image generated by the SENSE reconstruction, followed by low pass filter correction and minimum intensity projection.

E. Data Analysis

Minimum intensity projections of the susceptibility weighted images were analyzed to determine four main regions from which to calculate vessel contrast, as demonstrated in Figure 2. Brain parenchyma and large vessel masks were generated by thresholding the mIP images, while small vessel and adjacent white matter regions were manually identified. Contrast ratios were defined as the mean signal intensity within the background tissue region divided by that within the vessel. The brain parenchyma region was used as the background tissue for the contrast ratio calculation of large vessels, while the sensitivity of detecting small vessels was evaluated using adjacent white matter as the background tissue.

To evaluate the performance of the GRAPPA and SENSE acquisitions, contrast ratios of the subsampled datasets were compared to the full FOV case. Statistical significance was determined through the use of a Wilcoxon signed rank test.

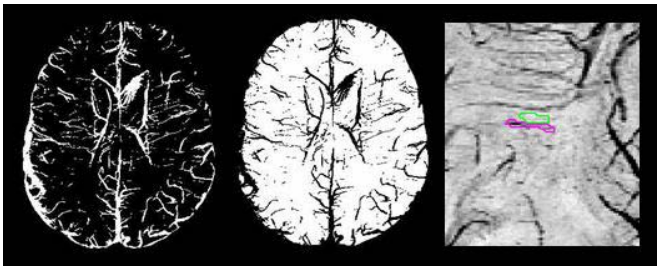

Figure 2: Depiction of brain regions used to generate contrast ratios. In a) large vessel mask, b) whole brain region mask, c) small vessel and neighboring white matter regions.

III. RESULTS

Figure 3 displays the reconstructed magnitude images before and after susceptibility weighting by the phase data for the three acquisitions types. Image uniformity after low pass filter correction was consistent among acquisitions.

The mean contrast ratios and corresponding p-values are displayed in Table 1. No statistically significant difference was found between the GRAPPA and full FOV acquisitions for both vessel sizes. SENSE performed similarly to GRAPPA and the full FOV acquisitions in terms of large vessel contrast, but showed significantly reduced contrast for the delineation of small vessels. Figure 4 depicts these differences in contrast on a subject by subject basis. A decrease in small vessel contrast was observed with the SENSE reconstruction across all volunteers compared to both the GRAPPA and full FOV cases. Although there was no statistically significant difference in large vessel contrast among acquisitions for the overall population, the large vessel contrast ratio of the SENSE acquisition deviated on average 14% from the full FOV data, compared to the 3% deviation exhibited by GRAPPA.

TABLE 1

Mean contrast ratios (CR) for large and small vessels and scan time

ACQUISITION	FULL FOV	GRAPPA	SENSE
LARGE VESSEL CR	3.07 ± .38	3.32 ± .48	3.19 ± .37
P-VALUE	--	P = .7	P = .15
SMALL VESSEL CR	1.25 ± .15	1.32 ± .14	1.08 ± .05
P-VALUE	--	P = .7	P = .03*
SCAN TIME	11 MIN	6.16 MIN	5.5 MIN

*denotes significant difference from full FOV acquisition

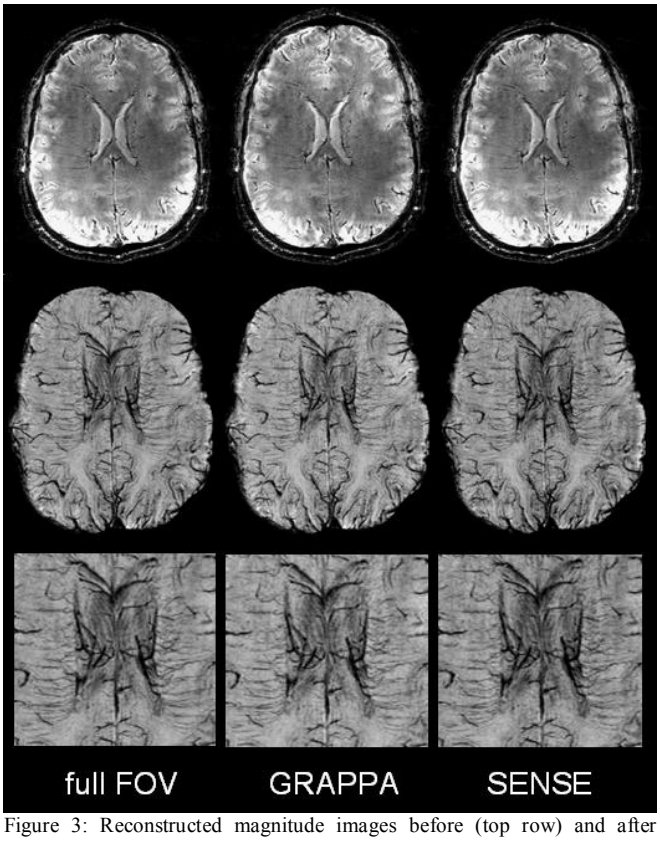

Figure 3: Reconstructed magnitude images before (top row) and after (middle row) SWI processing for the full FOV, GRAPPA, and SENSE acquisitions. The bottom row displays magnified SWI images for the visualization of small vessels.

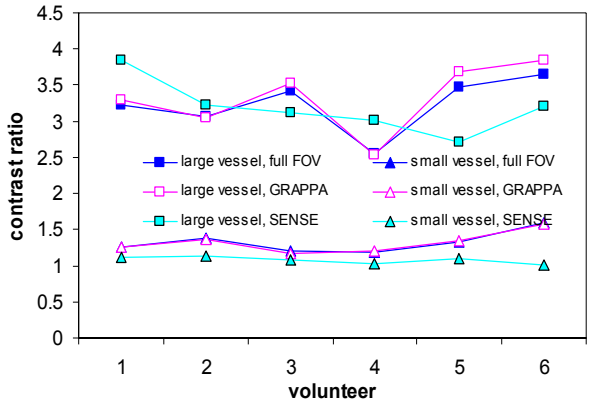

Figure 4: Contrast ratios for small and large vessels for full FOV, GRAPPA, and SENSE reconstructions, plotted individually for each volunteer.

IV. DISCUSSION AND CONCLUSION

Susceptibility-weighted imaging is a promising high resolution technique for high field systems beyond 3T. Implementation of a multi-channel coil array and parallel imaging can allow at least a 2-fold reduction in scan time without compromising the contrast between veins and surrounding brain tissue, which is especially attractive at 7T where long repetition times would prevent SWI from being routinely used in a clinical setting.

Although both PPI techniques implemented in this study were simulated for a reduction factor of two, GRAPPA exhibited overall improved vessel contrast compared to the SENSE reconstruction. The fact that SENSE showed less contrast in small vessels was surprising based on previous results at 3T that showed no difference in small vessel detection between SENSE and the full FOV acquisition [8]. This decreased contrast is most likely due to imperfect estimation of coil sensitivity maps. The arrangement and quality of the receive coil elements of the 8-channel coil at 7T were not necessarily optimized for SENSE and thus further evaluation of coil sensitivity profiles and geometry factor maps is necessary to confirm the accuracy of the contrast observed with the SENSE reconstruction.

In this work we showed the feasibility of accelerating SWI acquisitions two-fold by using partially parallel imaging techniques. Indirect techniques like GRAPPA may have an advantage over direct techniques such as SENSE in terms of robustness in detecting small vessels, despite the slight increase in scan time required to acquire the extra AC lines. Future work will explore higher reduction factors to determine the acceleration limit for these PPI techniques on maintaining the same level of sensitivity to small vessel detection.

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