Removal of air/tissue interface field effects in Susceptibility Weighted Imaging

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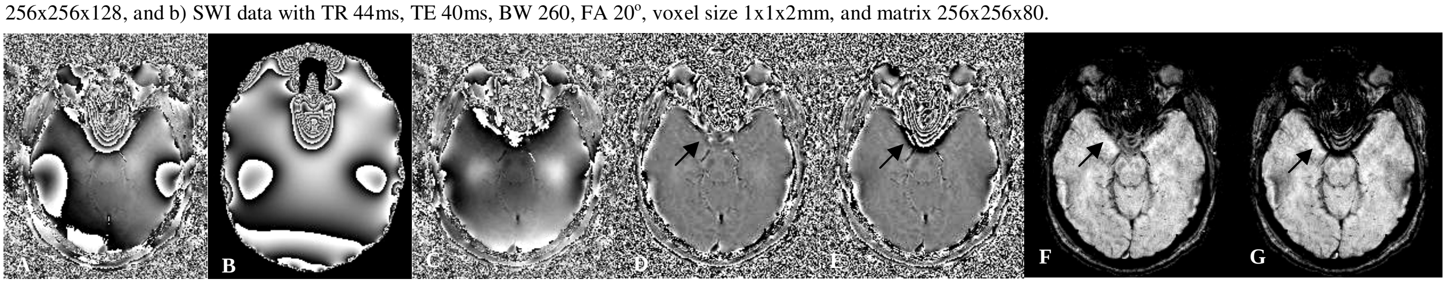
Introduction: Susceptibility weighted imaging (SWI) [1] is a method that uses phase information to reveal local susceptibility changes between tissues and as such relies on the removal of unwanted background field affects from phase images. Usually a high pass filter is used to remove the low spatial frequency phase variations. However, a very strong filter is often needed to remove the rapid phase aliasing near the air/tissue interface, which results in a concomitant loss of important local phase information. A weaker filter, on the other hand, makes it difficult to study the forebrain and midbrain regions due to strong remnant aliasing. We present here a novel method for removing these background phase effects. We use the 3D structural information of the air/tissue interface to estimate the geometry induced local field inhomogenieties and remove their contribution from the SWI phase images. The field estimation method we use is based on a Fourier transform method [2-5].

Theory, Materials & Methods: The induced magnetic field deviation distribution B(r) due to an object can be calculated through the Fourier transformation of its spatial susceptibility distribution matrix, $\chi(r)$, which includes the size and shape information of the object. Specifically, we take $B(r) = B_o \cdot FT^{-1}$**[** $FT[\chi(r)] \,.\, FT[G(r)]$ **] (Eq. 1)**, where B_o is the main magnetic field, FT stands for Fourier transform, and G(r) is the Green's function [5]. We can obtain the geometry of the object from a fast gradient echo, short echo time magnitude dataset. For using Eq. 1 to calculate the field inhomogenieties from air/tissue interface, the corresponding $\Delta\chi$ between the brain tissue and the sinus is to be found out. We postulate that using the phase images to estimate the field and the geometry (from magnitude), we can estimate the susceptibility by: a) a semi-quantitative approach: varying the susceptibility, calculating the difference between the predicted field and the measured field, and plotting the standard deviation (σ) of this remnant field (and choose χ corresponding to the result with the least σ); and b) a quantitative approach: regions within the object field map with significant SNR can be used to fit the predicted field distribution to that in Eq. 1, through a least squares fit.

First, both the methods are evaluated in a homogeneous phantom with a complicated geometry containing distilled water (with known $\Delta\chi_{water/air}$=9.4ppm) to see whether the measured χ, using both methods, agree with theory. Then we apply the standard deviation plotting method to human brain images to find the $\Delta\chi$ between two major interfaces influencing the SWI phase images: a) brain/air-filled sinus and b) brain/mastoid region (containing bone and mastoid air cells). We extract the geometry of the head from a T1 weighted dataset using a combination of complex thresholding [6] and anatomical knowledge. Since the Fourier transform is linear in nature, a complicated geometry can be expressed as a sum of multiple sub-structures, each with different χ values. The head geometry was modeled as the combination of the frontal, ethmoid, sphenoidal, maxillary sinuses, the brain tissue, and the mastoid cavities with different $\Delta\chi$. The head geometry and $\Delta\chi$ estimates from the σ-plot analysis were then used to calculate the field and hence phase at TE 40ms which was subsequently used to correct the original SWI phase images.

Phantom Imaging: a cylindrical polypropylene container (D=124mm,H=180mm) with a small hollow cylindrical cavity (D=12mm,H=70mm) in the middle was filled with distilled water and imaged at 1.5T (Siemens Sonata) using the SWI sequence [1] with TR 15ms, FA 6^o, voxel 0.78 x 0.78 x 0.78mm, matrix 256x256x192, BW 610 Hz/pixel, at TEs of 6.58ms and 9.58ms to obtain the field map. Prior to imaging, shimming was performed on a spherical phantom (D=170mm, $NiSO_4$ solution), to within 6Hz FWHM, and these settings used for imaging the water phantom. ***Imaging Volunteers***: A volunteer was imaged at 1.5T with the following parameters: a) 3D gradient echo at TE 5ms and 6ms (to obtain the field map) and TR 15ms, FA 20^o, BW 400 Hz/pixel, voxel 1x1x2mm, matrix 256x256x128, and b) SWI data with TR 44ms, TE 40ms, BW 260, FA 20^o, voxel size 1x1x2mm, and matrix 256x256x80.



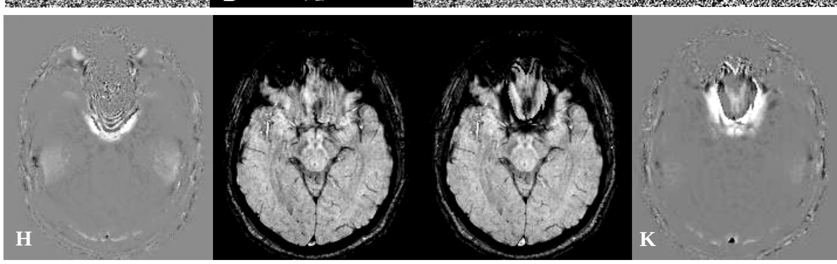


***Fig1**: (**A**) original phase image; (**B**) estimated TE=40ms phase image using method described; (**C**) complex dividing **B** into **A**; (**D**) high pass filtered image of C; (**E**) high pass filtered image of **A**; (F) SWI processed magnitude image using **D**; (**G**) SWI processed magnitude image using original high pass filtered phase in **E**; (**H**) difference between **F** and **G** showing the signal retained in the midbrain region. (**I**) another SWI processed magnitude using the corrected phase image generated using our method; (**J**) corresponding SWI processed magnitude using original high pass filtered phase; (**K**) Difference between **I** and **J** showing the signal retained in the forebrain region.*

Results: Both χ estimation methods in the water phantom agreed with theory. The standard deviation plots indicate that χ lies between 9 and 11ppm and the least squares estimate gives a value of χ = 9.337 ± 0.004 ppm (the expected value is 9.4ppm). Although the least squares method is highly promising, appropriate algorithm for its use in human phase images is yet to be optimized. The standard deviation plots for the human images indicate that the $\Delta\chi_{brain/Mastoid}$ is between 4 and 7 ppm and the $\Delta\chi_{brain/Other\_Sinus}$ is between 12 and 15 ppm. The images in the accompanying figure show the results of phase estimation and further processing using $\Delta\chi_{brain/Mastoid}$=4ppm and $\Delta\chi_{brain/Other\_Sinus}$=12ppm. There is a dramatic improvement in the estimates of the pristine phase behavior prior to high pass filtering, when comparing the corrected phase in Fig. C to the original phase in Fig. A. The old version of SWI shows terrible artifacts near the sinus region (Figs. G and J) and now most of these are removed (Figs. F and I).

Discussion and Conclusion: The ability of the field estimation method to remove the rapid aliasing near air/tissue interfaces and generate better SWI images is of utmost importance for studying mid and forebrain regions as well as diseases such as subarachnoid hemorrhages. Although most aliasing is completely gone, the new corrected phase now requires only a mild high pass filter to operate with the SWI processing. Compared to phase unwrapping techniques in 2D or 3D, this is a much more robust approach. A limitation of this method is the requirement of the geometry information to produce the correcting field map. In summary we present here a powerful method for removal of background field affects from SWI phase images to produce an artifact free phase and hence better, "cleaner" SWI processed images.

References: [1] Haacke et al., MRM, 52:612, 2004. [2] Deville et al., Phys. Rev. B, 19:5666, 1979. [3] Salomir et al., MR Eng. 198:26, 2003. [4] Marques and Bowtell, MR Eng. 25:65, 2005. [5] Neelavalli et al., ISMRM, 1016, 2007. [6] Pandian D., et al., A complex threshold method for identifying noisy background pixels in MRI. MS Thesis, CSC Dept., WSU 2007.