Susceptibility Weighted Imaging and Susceptibility Mapping (SWIM): A New Means to Visualize Veins and Quantify Susceptibility

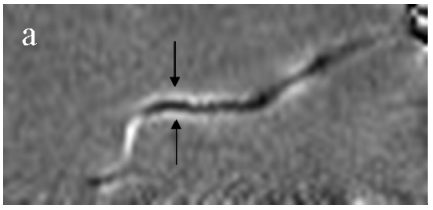
E. M. Haacke[1,2], J. Tang[3], Y-C. N. Cheng[1], and J. Neelavalli[2,4]
[1]Academic Radiology, Wayne State University, Detroit, Michigan, United States, [2]The MRI Institute for Biomedical Research, Detroit, Michigan, United States, [3]McMaster University, Hamilton, Ontario, Canada, [4]Nuffield Department of Surgery, University of Oxford, Oxford, United Kingdom

Introduction: Susceptibility weighted imaging (SWI) has been used for some time as a means to enhance venous vessels using high pass filtered phase images [1,2]. However, efficacy of SWI processing relies especially on an asymmetric voxel aspect ratio [3] and to a slight extent on the relative orientation of the head and the main magnetic field. Therefore, when SWI is collected as high resolution isotropic data, the conventional processing will fail. To overcome this problem and make the venous vessel depiction independent of these factors, we propose using a form of susceptibility mapping to produce an image of veins from SWI data [1] which we refer to here as SWIM (susceptibility weighted imaging and mapping).

Materials and methods: The method uses a direct inverse approach of the Green's function in k-space relating susceptibility distribution and the corresponding magnetic field perturbation [4, 5]. To reconstruct a susceptibility map with minimal artifacts using the inverse approach, following steps were followed: i) collect an isotropic high resolution SWI data set, ii) high pass filter the phase images, iii) interpolate k-space, iv) remove spurious phase noise sources from the phase images and v) regularize the data. It should be noted that "k-space" in this abstract refers to the Fourier transform of the high pass filtered phase images, and not the usual acquired k-space data in MRI. We collected images with 0.5mm isotropic resolution at 4T with three echo times of 11.6ms, 15ms, and 19.2 ms. The bandwidth was 80 Hz/pixel. A total of 88 slices were collected. Noise in the phase images was first removed, by using a complex threshold [6] approach, followed by a skull stripping algorithm. In order to minimize aliasing of the Gibbs ringing in the image, k-space was interpolated by zero filling the phase images to a 512 × 512 × 128 matrix size. This 3D image set, $\varphi_{zf\text{-}proc}(r)$, was then Fourier transformed to $s_{zf\text{-}proc}(k)$. A regularized inverse filter, $g^{-1}(k)$, was applied to the Fourier transform of the processed high pass filtered phase image. The forward filter g(k) is defined by $g(k) = 1/3 - k_z^2/k^2$ where $k^2 = k_x^2 + k_y^2 + k_z^2$. Regularization of $g^{-1}(k)$ was carried out as follows: first, for any k where $g(k) < 0.1$, g(k) is set to 0.1 (i.e., $g^{-1}(k)$ is set to a maximum of 10). Second, the inverse is brought smoothly to zero as k approaches the singularity k_o, to remove the discontinuity at $k = k_o$ by multiplying $g^{-1}(k)$ by the constant $\alpha^2(k_z)$ where $\alpha(k_z) = (k_z - k_{zo})/(b\Delta k_z)$ for $(k_z\text{-}k_{zo}) < b\Delta k_z$. Here, k_{zo} is the point at which the function $g^{-1}(k)$ becomes undefined, Δk_z is the k-space sampling interval along z and b was empirically set equal to 30. The susceptibility map for the 3D data set is calculated via: $\chi(r) = FT\text{-}1(g^{-1}_{reg}(k)s_{zf\text{-}proc}(k))/(\gamma B_o TE)$ where the subscript "zf-proc" refers to the high pass filtered, zero filled phase images.

Results and Discussion: An example of the successful inversion process for a vein whose phase behavior changes along it's course due to its changing orientation w.r.t the main magnetic field is shown in Figure 1 where after the SWIM processing, the vein appears as one contiguous object. In Figure 2, we present an example set of SWIM images from the 0.5mm isotropic resolution data with echo times of 11.6ms, 15ms, and 9.2ms. Several observations can be made here: first, the image with the shortest echo time is the noisiest as it has much less phase information. Second, the contrast between gray matter and white matter improves at longer echo times. Third, the smaller vessels become more visible and better defined at longer echo times because there is more phase information. There are six major sources of error in creating venous susceptibility maps: i) errors due to $g^{-1}(k)$, ii) errors caused by high pass filtering the input phase data, iii) discretization errors, iv) partial voluming, v) aliasing caused by longer echo times and vi) thermal noise. Quantitatively, high pass filtering removes the unwanted air-tissue interface effects but also modifies the measured susceptibility of the vessels making absolute quantification difficult. Although quantification was not the primary goal of this work, through simulations, we find that the susceptibility values are highly affected by both the high pass filtering and partial voluming of the vessels.

Although regularization is an important part of the susceptibility mapping approach in SWIM, the following points are also critical: 1) properly handling the k-space interpolation process; 2) spatial resolution must be high enough to allow for as many pixels as possible to give useful phase information outside the source; 3) echo time must be long enough for enough phase to develop so as to achieve sufficient phase SNR; 4) not only should the noisy pixels from the magnitude image be thresholded, but also areas of wildly varying phase coming from, for example, the skull where out-of-phase fat can cause a serious problem. Setting the phase inside spherical or cylindrical regions to be a constant also helps avoid phase noise effects which lead to a new type of Gibbs ringing that we refer to as the inverse dipole rippling effect; and 5) partial volume effects should be avoided.



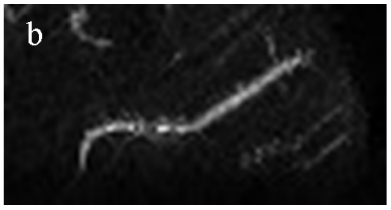


Figure 1: (Top Left) a) A sagittal cut demonstrating the change in phase as a vein courses through the brain for the TE = 19.2ms case. In the original phase image, the dipole effect is clearly seen with the vessel appearing dark in the phase image when perpendicular to the field and bright when parallel to the field. The distance from the top of the vessel (upper arrow) to the bottom of the vessel (lower arrow) is 7 slices (3.5mm). The fact that the phase is clearly seen in one slice with the correct sign suggests that the vein is on the order of one or two pixels in diameter (i.e., 0.5mm to 1mm). b) The same vein shown from the susceptibility map (SWIM image) now shows no orientation dependence.

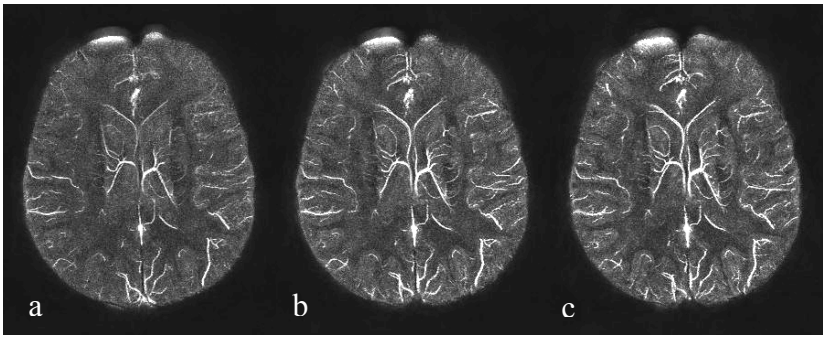


Figure 2: MIPped SWIM images from three different echo times: a) 11.6ms, b) 15ms and c) 19.2ms. Note the improvement in image quality and vessel visibility as the echo time increases. These images were MIPped over 16mm (32 slices).

Conclusions: SWIM processing can successfully create venograms of the brain with varying levels of contrast-to-noise depending on the size of the vessel and the echo time used. These images should be independent of the head orientation in the magnet. Quantitatively, in the future, it should be possible to use this approach to evaluate microbleeds, calcifications and map oxygen saturation from veins of all angles automatically. Finally, These SWIM images can serve as a new form of MR venography with the potential of directly depicting variations in magnetic susceptibility, i.e., indirectly mapping oxygen saturation.

References: [1] Reichenbach JR, Venkatesan R, et al. Radiology (1997) 204: 272. [2] Haacke EM, Xu Y et al. MRM (2004) 52:612. [3] Xu Y, Haacke EM MRI (2006) 24 : 155. [4] Marques JP, Bowtell R et al. Concepts Magn Reson Part B (Magn Reson Engineering) (2003):198:26. [5] Cheng YC-N, Neelavalli J et al. Phys Med Biol (2006); 51(24): 6381. [6] Pandian DS, Ciulla C et al. JMRI (2008): 28: 727.