MRI RAW DATA BASED MAGNETIC SUSCEPTIBILITY CONTRAST

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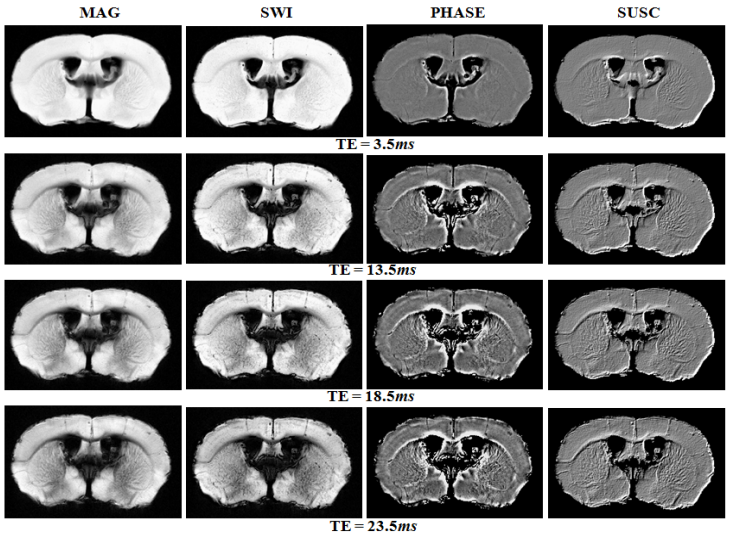
Introduction: The magnetic resonance technique of susceptibility weighted imaging is sensitive to tissue magnetic susceptibility variations that result in local inhomogeneities in the magnetic field. It generates a unique contrast, which increases with field strength and is different from that of spin density, T_1, T_2 and T_2^* [1-6]. Subtle differences in magnetic susceptibility between brain tissue types improve anatomical contrast and detail. This contrast mechanism detects diagnostic changes that may be missed by methods deriving contrast from differences in relaxation time in a range of disorders including stroke, Alzheimer's disease and cancer [1, 2]. To date, magnetic susceptibility information has been derived directly from image signal phases and indirectly from T_2^* weighted imaging sequences [5]. Local susceptibility variations lead to time delays in the raw MRI data with corresponding signal phase changes in reconstructed images. A phase mask derived from phase images has been used to enhance contrast in magnitude images by suppressing voxels within a certain range of phase values, a technique named susceptibility weighted imaging (SWI) [3]. The phase contrast imaging technique deals only with phase data corrected for macroscopic susceptibility effects [4, 6]. To obtain images reflecting susceptibility variations in tissue, filtering of the phase image is performed to remove the low-spatial frequency components of the background field. Local variations in magnetic susceptibility will also alter signal magnitude and capturing this information may improve susceptibility-derived tissue contrast. Here, we compare SWI and phase contrast imaging with a new method that uses all components of the raw data to create images sensitive to susceptibility variations.



Method: Raw MRI data can be expressed in terms of the susceptibility-free signal and the contribution from the magnetic susceptibility of the sample. In our method, we reconstruct the signal due to magnetic susceptibility after approximating the susceptibility-free signal. We recently showed that when the wave equation is used to approximate raw MRI data susceptibility information is removed. For an echo signal consisting of a sequence of raw data points f_l, l is an element of $[1,L]$, and (1) holds in the absence of biological and instrumentation noise, where θ is a change due to susceptibility, G_l is defined as $G_l(t,s)$ and c is the speed of wave propagation. To obtain a measure of localised variations in susceptibility, we approximate the left-hand-side of (1). We assume that f_l is analytic as the data are Fourier encoded. Hence, N plane waves can be used to approximate it, stated as (2). Substitution of (2) into (1) gives an exponential expansion for the function G_l. In (3), A_n and a_n are constants, c is the wave speed and $\mathbf{h} = \{A_n, a_n : n = 1 \text{ to } N\}$ are obtained by minimising (4).

$$G_l(s,t) \approx \frac{\partial^2 f_l}{\partial t^2} + c^2 \frac{\partial^2 f_l}{ds^2} = \frac{\partial^2 f_l}{\partial \theta^2} + c^2 \frac{\partial f_l}{\partial \theta} \nabla^2 \theta, \qquad (1)$$

$$f_l(t,s) \approx \sum_{n=1}^{N} \hat{A}_n e^{ja_n\left(t+s/c\right)}. \qquad (2)$$

$$G_l \approx \frac{\partial^2}{\partial t^2}\left\{\sum_{n=1}^{N} \hat{A}_n e^{ja_n\left(t+s/c\right)}\right\} + c^2 \frac{\partial^2}{\partial s^2}\left\{\sum_{n=1}^{N} \hat{A}_n e^{ja_n\left(t+s/c\right)}\right\} = -2\sum_{n=1}^{N} a_n^2 \hat{A}_n e^{ja_n\left(t+s/c\right)} = \sum_{n=1}^{N} A_n e^{ja_n\left(t+s/c\right)} \qquad (3)$$

$$\alpha = \left\| G_l - f_l(t,s) \right\| \qquad (4)$$

Data acquisition: Raw MRI data was acquired using a Bruker 16.4T Biospin small animal MRI instrument running Paravision version 5. An *ex vivo* mouse brain was imaged using the standard gradient echo sequence (*i.e.* FLASH) with the following MRI acquisition parameters: matrix size = 256 by 256, repletion time (TR) = 1.5*s*, flip angle = 30°, bandwidth = 50,000*Hz*, field-of-view = 10.2*mm* by 10.2*mm*, slice thickness and separation = 0.5*mm*, rare factor = 1, number of slices = 209 and zero filling was not used. Multiple echo times (TEs) were set to achieve strong T_1 to T_2^*-weightings. The TEs set were (*ms*): 3.5, 8.5, 13.5, 18.5, 23.5, 28.5, 33.5, 38.5, 43.5, 48.5, 53.5 and 58.5. Mouse brain relaxation times were measured using built-in relaxometry sequences: $T_1 = 2.4s$, $T_2 = 51ms$ and $T_2^* = 24ms$. These numbers were found to vary across different regions of the brain. No additional filtering of the raw data was performed, other than what is used to record *k*-space data.

Image reconstruction: All computations were performed using in-house functions written in 32-bit MATLAB version 7.10.0.499 (R2010a). Four different images were reconstructed from the variable TE mouse brain raw MRI data. Magnitude images were reconstructed by taking the Fourier transform of *k*-space data to obtain image voxel complex signals. The SWI reconstructions have been optimized as outlined in [8]. The microscopic phases were used to reconstruct a phase mask to enhance magnitude image detail using correlation between magnitude and phase information. Phase contrast images have been reconstructed according to the method provided in [4, 6]. Essentially, phase contrast images illustrate microscopic phase variations. The final image provided is our susceptibility contrast reconstruction, obtained through the steps outlined in the previous section. We approximated raw MRI data using 128 basis functions, which is one half of the matrix size in the readout direction. The method of least squares was used to approximate basis function coefficients. The four different reconstructions are illustrated side-by-side to allow comparison between them.

Results: Images of a brain slice correspond to TE = 3.5, 13.5, 18.5 and 23.5*ms*. At TE=28.5*ms*, significant deterioration of signal is present, and hence, we do not illustrate large TE results. The magnitude of voxel signal intensities has been normalized for all images depicted. As the TE is increased, larger effects due to susceptibility are expected in the images. This is evident by comparing the SWI images from small to large TEs. The contrast achieved in the PHASE images is also greater with an increase in TE. Our proposed reconstruction, as illustrated under SUSC, also shows an increase in detail, which is consistent with the other two susceptibility based methods. We observe that after TE=18.5*ms* susceptibility contrast does not appear to increase.

References

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