In-vivo Rat Brain Tissue Characterization by Susceptibility Weighted Imaging at 9.4 T

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INTRODUCTION

Magnetic heterogeneity of brain tissue is caused by the venous vascular system (paramagnetic blood deoxyhemoglobin), by different tissue iron concentrations and tissue myelin content [1]. These heterogeneities are the origin of contrast in susceptibility weighted imaging (SWI) since they induce an offset in the resonance frequency that can be detected in the signal phase. In this study SWI was used to improve contrast in rat brain regions of different magnetic susceptibilities by combining magnitude and phase information [2] and to estimate, for the first time on animal models, frequency shifts occurred within and between white matter (WM) and gray matter (GM).

MATERIALS AND METHODS

Wistar rats were imaged on a Bruker 9.4 T Biospec System with 600 mT/m gradient set. 3D gradient echo flow compensated images were acquired with the following parameters: TR = 53 ms, TE_{opt} = 24 ms, flip angle = 14 degrees, FOV = 30×30×16 mm^3, matrix = 256×256×30, leading to a 0.117 × 0.117 × 0.533 mm^3 spatial resolution and an acquisition time of 6m 46s. In order to obtain optimum phase contrast, TE_{opt} = 24 ms, was chosen based on T_2* values [1] of GM (26.5±1.7ms) and WM (19±1.6ms) at 9.4 T. A four channel surface array was used as receiver only. Iron based contrast agent VSOP (2 µl/gram body weight, Ferropharm, Teltow, Germany) was injected to characterize rats' cerebral vascular architecture enabling the identification of small vascular density regions along the WM/GM boundary. In this region, (dotted line, Fig. 1b, c) the frequency shift corresponding to the observed phase differences between WM/GM was estimated. Consequently, the phase shifts induced by paramagnetic deoxyhemoglobin of venous blood are negligible. Post-contrast agent injection images were acquired at shorter TE (11 ms) in order to compensate T_2* blood signal decay caused by the contrast agent. Images were processed offline with an in-house developed software written in Matlab language to calculate the magnitude and phase maps. Background field inhomogeneities and coil sensitivities induce low spatial frequency phase variation within the image. These effects were removed by applying a high-pass filter. As suggested by Haacke *et al* [2] images obtained with each channel were low-pass filtered using a Hanning filter (7×7 pixels). High-pass filtered phase images are obtained through a complex division of the original images with the low-pass filtered images. Magnitude images are calculated using the standard sum of squares reconstruction (Fig. 1a). Phase effects caused by different channel orientation are effectively removed by the high-pass filter. Since the phase standard deviation is inversely proportional to the signal-to-noise ratio (SNR) of the magnitude MR-image [3], one phase image is calculated using the following relation:

$$I_{\phi} = \sum_{i=1}^{n_{channel}} I_{\phi i} \cdot I_{Mi} \Big/ \sum_{i=1}^{n_{channel}} I_{Mi} \qquad (1)$$

where I_{Φ} represents the resulting phase image, $I_{\Phi i}$ and I_{Mi} the phase and the magnitude images obtained with each individual channel respectively.

An intensity mask is computed from the magnitude image and low intensity pixels are not evaluated in the obtained phase image. The resulting phase is used to create a phase mask. Since the background field inhomogeneities were effectively removed by the high-pass filter, the resulting phase values are close to zero. Values between $-\pi/2$ and $\pi/2$ were linearly normalized between -1 and 1 such that -1 corresponds to $-\pi/2$ and 1 to $\pi/2$ (M_{init}). Phase values not included in the interval [$-\pi/2$, $\pi/2$] were not computed since they were removed by the high-pass filter. The resulting mask, M_{final}, was calculated using the following equation:

$$M_{final} = \mathrm{sign}(M_{init}) \cdot |M_{init}|^{1/n} + 1 \qquad (2)$$

The factor n=3 was empirically found to produce the best result.

Final values of M_{final} are between 0 and 2, in the resulting magnitude image (Fig.1b) voxels having negative phase values are attenuated and those with positive phase values are amplified. The frequency shift estimation between WM/GM was carried out on the phase images, the resulting magnitude images after phase mask multiplication as well as the resulting masks, M_{final} were only used for better visualization of anatomical structures.

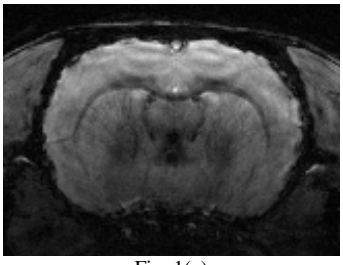
Fig. 1(a)

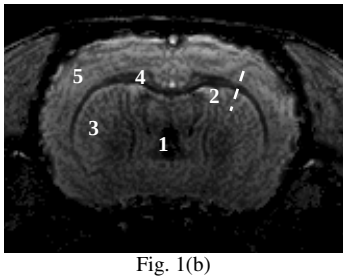

Fig. 1(b)

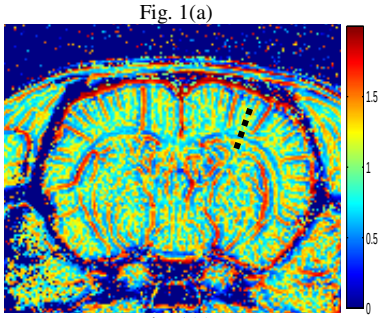

Fig. 1(c)

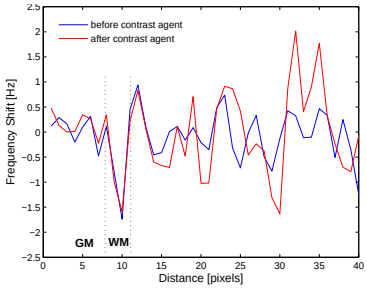

Fig. 1(d)

Fig 1: Magnitude data of a 0.5 mm slab (1a) and the resulting magnitude image after phase mask multiplication (1b). Note the anatomical details visible at this resolution, fornix (1), ventricle (2), striatum with spreaded myelinated fibers (3) and the high contrast between WM (4) and GM (5). The frequency shift between WM and GM in a region of weaker vascular density (dotted line) was found close to 1.7 Hz (1d). Phase mask computed after contrast agent injection is displayed in Fig 1c and give a comprehensive insight of brain vascular system.

RESULTS/DISCUSSION

In most of the brain regions contrast between WM and GM is nicely resolved in both magnitude and phase data. The resulting magnitude image 1b shows clear evidence of anatomical structures like fornix, ventricles, striatum (with spreaded myelinated fibers) and venous vasculature. In the phase data the frequency differences between WM/GM was calculated along their boundary and estimated at 1.9±0.65 Hz. The GM frequency was higher than that of WM, corresponding to a larger (and more positive) magnetic susceptibility (Fig. 1d). The relatively high vessels density (Fig 1c) in GM relative to WM, variations of blood oxygenation and tissue composition (i.e. degree of myelination) could be possible causes of these phase differences noticed between and within both tissue types [1]. Zhong *et al* [4] suggested that water-molecule exchange processes could also result in phase differences between WM/GM. The frequency shift between WM/GM was estimated (Fig 1d) before (blue line) and after (red line) contrast agent administration in areas of smaller vascular density in order to neglect susceptibility effects of paramagnetic deoxyhemoglobin in vessels. Frequency shifts up to 1.7 Hz are obtained. The reduced vessels contribution is noticed since negligible frequency shifts were seen before and after bolus.

CONCLUSIONS

In this study it is demonstrated that SWI can provide enhanced image contrast by combining phase and magnitude information. A strong contrast was observed between WM and GM with a frequency shift estimated at 1.7 Hz in specific regions of negligible susceptibility effects caused by paramagnetic deoxyhemoglobin. The resulting improved contrast makes our method very suitable for application on animal models of demyelination to detect subtle changes induced by myelin loss, gliosis or axonal damage. This method is also adapted for checking the localization of demyelinated lesions in areas with high vascular density as observed in multiple sclerosis patients [5] and to differentiate between different types of plaques.

REFERENCES

[1] Duyn JH *et al*, PNAS, 2007; 104:11796-11801. [2] Haacke EM *et al*, MRM, 2000; 12:525-533. [3] Conturo TE and Smith GD, MRM, 1990; 15:420-437. [4] Zhong K *et al*, Neuroimage, 2008; 40:1561-1566. [5] Hammond KE *et al*, Neuroimage, 2008; 1682–1692.