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Modified human contrast sensitivity function based phase mask for susceptibility-weighted imaging

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ABSTRACT

The aim of the work is to increase the visual information in magnetic resonance imaging based susceptibility weighted images. Our approach is to amplify spatial frequency information of the phase mask used to increase susceptibility weighting using a modified version of the human contrast sensitivity function. Thereby, we propose a modified version of the human contrast sensitivity function for use in phase mask creation. Comparison with conventional susceptibility-weighted imaging was undertaken on a qualitative basis and quantitatively with a number of established image quality metrics on *ex vivo* mouse brain magnetic resonance images obtained at 16.4 T at various echo times. Four experts also compared the quality of *in vivo* 1.5 and 3 T human brain magnetic resonance images generated with traditional susceptibility weighted imaging and with the new method. We found that parameters of the modified human contrast sensitivity function can be chosen to improve delineation of structural detail of mouse and human brains. Information contained in susceptibility-weighted images generated using the modified human contrast sensitivity function based phase mask corresponds to that in the conventional method, however the visual range over which it is depicted has improved visual perception. Hence, qualitative evaluation of information contained in susceptibility-weighted images can be improved by amplifying spatial frequencies where human contrast sensitivity is reduced.

1. Introduction

Magnetic resonance imaging (MRI) signals have magnitude and phase which may provide complementary information. Clinical diagnostic imaging most commonly relies on information based on signal magnitude, which is dominated by proton density and the influences of T_1 and T_2^* relaxation times. Signal phase also contains information about the structural properties of tissue with contrast arising from changes in the magnetic field induced by variations in magnetic susceptibility (Barnes and Haacke, 2012; Deistung et al., 2008; Duyn et al., 2007b; Haacke et al., 2004b). This effect scales with the static field strength of the scanner, and contrast differences in 7*T* phase images are sufficient to enable cortical structure to be distinguished in fine detail (Duyn et al., 2007b). The combination of both signal magnitude and phase are increasingly performed across a range of MRI studies (e.g. Beauchamp et al., 2011; de Champfleur et al., 2011; Jagadeesan et al., 2011; Zivadinov et al., 2012).

In susceptibility-weighted imaging (SWI), a phase mask derived from gradient recalled echo (GRE) MRI phase images is multiplied with the magnitude image (Barnes and Haacke, 2012; Deistung et al., 2008; Haacke et al., 2004b; Haacke et al., 2009). SWI has been used in cortical parcellation and in studying brain iron levels and calcification and blood oxygenation of the cerebral venous system (Barnes and Haacke, 2012; Deistung et al., 2008; Duyn, 2010; Haacke et al., 2004b; Haacke et al., 2005; Haacke and Ye, 2012; Robinson and Bhuta, 2011).

The distribution of voxel values achieved in SWI can be altered by changing the echo time of the GRE-MRI sequence, by changing the level of phase filtering prior to phase mask creation or by altering phase mask contrast by raising voxel values to a different power. A change in echo time requires additional data acquisition, whereas the latter two image processing techniques can be applied on already acquired data. Increased echo time leads to increased signal phase but also increases noise and signal magnitude decay. The exponentially decaying modulation of the signal magnitude with time imposes the condition that echo time should not be greater than T_2^* to maximize image contrast and to preserve detail (Duyn et al., 2007a; Vegh et al., 2012).

Phase masks may accentuate either negative and positive phase effects (Haacke et al., 2009; Mittal et al., 2009). Masks are made up

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of values between 0 and 1, with most values falling around 1. The power to which phase mask entries are raised alters the distribution of values. For example, if adjacent voxels in the phase mask have values of 0.9 and 0.909 (1% larger), raising voxel values to the power of 4 amplifies the difference to 4% ($0.9^4 = 0.6561$ versus $0.909^4 = 0.6827$). Hence, the distribution of voxel values changes with the power applied to the phase mask. Notably, the use of large powers leads to values concentrated around zero, diminishing the utility of the technique. Simulation studies have shown that contrast due to veins is maximized when a power between 3 and 5 is applied to the phase mask (Haacke et al., 2004b). Subsequently, the power of 4 has been employed in qualitative studies (Barnes and Haacke, 2012; Haacke et al., 2004b).

Our work focuses on changing the distribution of values in the phase mask to enhance SWI contrast. In human visual perception, contrast sensitivity varies with the spatial frequency of visual stimuli and if the latter is too high, the pattern of stimuli can no longer be perceived. For example, for an image consisting of vertical black and white stripes, if the stripes are very thin individual stripes cannot be distinguished and all that is seen is a gray image. If the width of the stripes is progressively increased, there is a threshold width above which stripes are perceptible. The relationship between visual contrast sensitivity and spatial frequency of visual stimuli is described by the Human Contrast Sensitivity Function (HCSF) (Farrell, 1999; Kelly, 1975; Mannos and Sakrison, 1974) (see Fig. 1(a)) and has been attributed to the eye's optical properties, cone photoreceptor sampling aperture and both passive and active neuronal connections (Daly, 1992). In addition to the spatial frequency of image components, the size of the image, its eccentricity and the level of background noise may also affect perception. Taking into account the HCSF, we sought to enhance the perception of image contrast across the spectrum of spatial frequencies by increasing contrast at spatial frequencies where contrast sensitivity is low. Fig. 1(b–d) provides a simple illustration of how perception of image content can be enhanced by applying a one-to-one mapping between the original image value and a new image with elevated contrast using a contrast sensitivity function. We developed a method of combining band-pass filtered phase images to amplify information in the low sensitivity region of the HCSF spectrum, resulting in a novel approach of creating phase masks which were multiplied with the magnitude image to generate a new SWI image. We evaluated the performance of the method across a range of echo times and parameter settings. HCSF-modified SWI results were qualitatively and quantitatively compared with traditional SWI reconstructions.

2. Experimental

2.1. MRI data acquisition — 16.4 T ex vivo mouse brain

MRI raw data were acquired on a 16.4 T Bruker Biospin® animal MRI instrument running Paravision® (V5). An *ex vivo* mouse brain was imaged with the standard gradient recalled echo (GRE) sequence. The imaging acquisition parameters applied were: matrix size = 256 × 256, repetition time (TR) = 1.5 s, flip angle (α) = 30°, bandwidth = 50,000 Hz, field-of-view (FOV) = 10.2 mm × 10.2 mm, slice thickness and separation = 0.5 mm and number of slices = 20. Zero filling was not applied. Multiple echo times (T_E) were set to obtain different T_2^*-weighting levels in resultant images. The range of T_E selected is from 3.5 to 58.5 ms, in steps of 5 ms. The T_2^*-relaxation time of the mouse brain was calculated through the standard Paravision relaxometry sequence, which was 24 ms. The filtering of raw data beyond what is standard in Paravision was not performed.

2.2. MRI data acquisition — 1.5 and 3 T in vivo human brain

The research was approved by the hospital's Human Research Ethics Committee. Forty-three patients admitted with a clinical diagnosis of acute ischemic stroke to Royal Brisbane and Women's Hospital were recruited between May 2011 and April 2012. Patients underwent an MRI examination at admission from which five MRI data sets were randomly selected for this study. MRI susceptibility weighted images were acquired on a 1.5 T Siemens® Avanto and 3 T Siemens® Trio human scanners running Syngo® housed at the hospital. The Syngo SWI sequence was used with the following parameters: matrix size = 224 × 256, repetition time (TR) = 200 ms, echo time (TE) = 20 ms, flip angle = 15°, bandwidth = 120 Hz per pixel, in-plane resolution = 1 mm × 1 mm, slice thickness and separation = 2 mm and number of slices = 72.

3. Theory/calculation

3.1. Phase filtering and ex vivo mouse brain image reconstruction

SWI and HCSF SWI reconstructed images were obtained by implementing algorithms in MATLAB® version R2013a running on a 12 core 64 GB 64-bit Windows 7® Dell Precision T7500 workstation. Specifically, the original MRI data are complex signals containing magnitude and phase information, which can be written as (Wang et al., 2000):

$$I(x, y, z) = M(x, y, z)\, e^{i\theta(x,y,z)}, \tag{1}$$

where I is the reconstructed complex signal for a voxel with coordinates (x, y, z), and M and θ are the magnitude and phase signals, respectively. In the *ex vivo* mouse brain study magnitude images were generated by taking the absolute value of Fourier-transformed k-space data. The voxel intensities were then normalized to the range [0, 1] for visualization purposes.

The application of low spatial frequency filtering of phase signals serves the purpose of removing unwanted background effects due to gradients and instrumentation. The amount of information retained or removed changes the extent to which phase signals due to sample susceptibility are retained. Hence, the filter cut-off affects the result. When the established method of homodyne filtering is used to remove low spatial frequency phase differences, the amount of high frequency information contained in the phase mask is a function of Fourier filter size. Previous results suggest that a Fourier filter size of at least 1/8th or larger of the image matrix size should be applied to remove background variations and unwanted phase information effectively (Barnes and Haacke, 2012; Rauscher et al., 2003; Wang et al., 2000). Several methods including polynomial fitting and projection onto dipole fields have been proposed for filtering of raw phases (Duyn et al., 2007a; Lee et al., 2013; Liu et al., 2011). These methods have been shown to improve the high-pass filtered phase result at air-material interfaces but not within the brain. As our work deals with the perception of image information within the brain, we limited our study to the application of homodyne filtering of raw signal phases.

To briefly explain homodyne filtering, we begin by noting that the Fourier filter is a symmetric low pass filter that removes high-frequency k-space content. The filtered complex signals (I_F) can be written as:

$$I_F = M_F e^{i\theta_F} \tag{2}$$

where M_F and θ_F are the low pass filtered magnitude and phase maps, respectively. The high frequency component of the phase is obtained by the division of the original image by the filtered complex image. This division of two complex images is termed homodyne