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Physiological noise compensation in gradient-echo myelin water imaging

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ABSTRACT

In MRI, physiological noise which originates from cardiac and respiratory functions can induce substantial errors in detecting small signals in the brain. In this work, we explored the effects of the physiological noise and their compensation methods in gradient-echo myelin water imaging (GRE-MWI). To reduce the cardiac function induced inflow noise, flow saturation RF pulses were applied to the inferior portion of the head, saturating inflow blood signals. For the respiratory function induced B_0 fluctuation compensation, a navigator echo was acquired, and respiration induced phase errors were corrected during reconstruction. After the compensations, the resulting myelin water images show substantially improved image quality and reproducibility. These improvements confirm the importance and usefulness of the physiological noise compensations in GRE-MWI.

Introduction

In conventional myelin water imaging (MWI), multi-echo spin echo (SE) data are acquired to measure T_2 decay (Mackay et al., 1994). In white matter of the brain, this decay has been demonstrated to contain multiple T_2 signals that originate from myelin water, axonal water and extracellular water. The measured data are fitted to a large number of multi-exponential decay functions using a nonnegative least squares method (Mackay et al, 1994; Whittall et al., 1997) or a few characteristic functions (Lancaster et al., 2003; Raj et al., 2014). Then the short T_2 myelin water signal fraction is estimated to generate a myelin water fraction as the ratio of myelin water signal fraction to total water signal.

In recent studies, signal decay characteristics of gradient echo (GRE) in white matter of the brain have been explored (Chen et al., 2013; Du et al., 2007; Hwang et al., 2010; Sati et al., 2013; Sukstanskii and Yablonskiy, 2014; van Gelderen et al., 2012; Wharton and Bowtell, 2012). These studies have demonstrated that T_2^* decay is also composed of multiple T_2^* components that originate from myelin water, axonal water and extracellular water. More interestingly, the three components have shown to possess distinct frequency offsets that change based on the relative orientation of axons to the B_0 field (Sati et al., 2013; Wharton and Bowtell, 2012). The origin of this signal change has been attributed to myelin (Lee et al., 2012; Liu et al., 2011) which has complex effects on the frequency offsets (He and Yablonskiy, 2009; Lee et al., 2010; Sati et al., 2013; Wharton and Bowtell, 2012). Based on these observations, the GRE signal decay has been fitted to a multi-component complex model that includes the frequency offsets to estimate the signal fraction of each component (Nam et al., 2015; Sati et al., 2013; van Gelderen et al., 2012; Wharton and Bowtell, 2012, 2013). The resulting signal parameters were used to generate MWI (Du et al., 2007; Hwang et al., 2010; Nam et al., 2015; Sati et al., 2013; van Gelderen et al., 2012). This approach of using GRE data to generate MWI is referred to as GRE-MWI in order to distinguish it from other MWI methods (Deoni et al., 2008; Kim et al., 2014; Labadie et al., 2014; Mackay et al., 1994; Oh et al., 2013). When compared to previous approaches of fitting magnitude decay data to a magnitude signal model (Du et al., 2007), the inclusion of frequency offset information and the use of a complex model have improved the reliability of the myelin water image estimation (Nam et al., 2015).

Despite the progresses in the signal modeling, however, the resulting myelin water images still suffer from artifacts due to an ill-conditioned fitting process and noises from various sources such as thermal and physiological origins. In particular, the effects of physiological noise (Glover and Lee, 1995; Noll and Schneider, 1994; Wen et al., 2014) have been observed (Du et al., 2007; Hwang et al., 2010; Oh et al., 2013) but have not yet been explored nor compensated in MWI.

In this study, we developed new methods to compensate for the physiological noise that originates from cardiac and respiratory functions in GRE-MWI. Artifacts from fast flowing blood were reduced by saturating inflow signals using saturation RF pulses. Respiration induced B_0 fluctuation artifacts were compensated by correcting phase

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errors using navigator echoes (Hu and Kim, 1994). With these compensation techniques, we demonstrate improved image quality and reproducible of GRE-MWI.

Materials and methods

Compensation for cardiac function induced inflow artifacts

Inflow of pulsatile blood can induce signal errors around vessels in the direction of spatial encoding (Nishimura et al., 1991). When multi-echo data are acquired, the signals from a vessel can be shifted over TE (yellow arrows in Figs. 2a, b, e and f), leading to alignment errors in multi-echo data fitting. In order to reduce the effects of such inflow artifacts on GRE-MWI, we propose to apply flow-saturation RF pulses at the inferior region of the imaging slab (Fig. 1). These RF pulses attenuate signal intensity of inflowing blood, reducing image artifacts (Figs. 2 and 3).

Compensation for respiratory function induced B_0 fluctuation artifacts

It has been shown that respiration induces B_0 field fluctuation in the brain (Noll and Schneider, 1994). This fluctuation has been suggested to produce artifacts in GRE images (Fig. 2a) (de Moortele et al., 2002; Lee et al., 2006; van Gelderen et al., 2007). In GRE-MWI, such artifacts may generate substantial errors in myelin water fraction estimation. In this study, a navigator echo was acquired at every TR to estimate a mean B_0 field fluctuation of an imaging volume (Fig. 1). The measured fluctuation was corrected during image reconstruction to reduce the artifacts (see below for details).

Data acquisition and processing

Data from eight healthy volunteers (seven male and one female, mean age = 28.5 years, and age range = 22–40 years; IRB approved) were collected at a 3 Tesla clinical scanner (Siemens Tim Trio, Erlangen, Germany) using a 32-channel phased-array receiver coil. For MWI, 3D multi-echo GRE data were acquired using following parameters: TR = 84 ms, number of echoes = 17, first TE = 1.6 ms, echo spacing = 2.0 ms, bandwidth per pixel = 1502 Hz, voxel size = $2 \times 2 \times 2$ mm^3, matrix size = $128 \times 128 \times 32$, and scan time = 5 min and 45 s. The flip angle (=60°; Ernst angle for T_1 = 120 ms and TR = 84 ms) was optimized for the short T_1 value of myelin water (Labadie et al., 2014) to generate the largest SNR for myelin water. Before the last echo, the phase encoding gradient was rewound to generate a navigator echo (last echo; TE = 33.6 ms; Fig. 1). For flow saturation, a regional saturation pulse (duration = 3.8 ms, off-resonance = 2.65 kHz, Hanning windowed sinc shape, TBW = 8, and flip angle = 90°) followed by all three directional spoiling gradients was placed in 20 ms before the excitation pulse at every TR. To acquire a rapidly decaying myelin water signal, the first TE was minimized and, therefore, no flow compensation gradients were applied. To evaluate the reproducibility of MWI, each subject was scanned four times using the GRE sequence: two scans without flow-saturation RF (FlowSAT OFF) and the other two scans with flow-saturation RF (FlowSAT ON). The scan was performed in a single session without repositioning the subject. To verify the effects of the respiration compensation, data were reconstructed with and without the B_0 navigator correction (B_0NAV ON and B_0NAV OFF). Hence, a total of four conditions (FlowSAT OFF & B_0NAV OFF, FlowSAT OFF & B_0NAV ON, FlowSAT ON & B_0NAV OFF, and FlowSAT ON & B_0NAV ON) were compared.

After collecting the GRE data, an MPRAGE image (TR = 1900 ms, TI = 900 ms, matrix size = $128 \times 128 \times 32$, GRAPPA acceleration factor = 2 and scan time = 2 min and 20 s) was acquired to generate a white matter mask for the reproducibility test. The white matter mask was segmented using FSL (FSL 5.0.7; University of Oxford, UK, http://www.fmrib.ox.ac.uk/fsl).

The total scan time including localization, manual shimming, four GREs, and MPRAGE was approximately 30 min. To minimize head motion during scans, sponge forms were placed at the space between the subject's head and the coil.

For compensation of respiration induced B_0 fluctuation, the following steps were applied during the reconstruction: First, the k-space navigator echo data were Fourier transformed along the readout direction, and the resulting complex data was ordered by their acquisition order. Then the data were low-pass filtered (0 to 2 Hz) along time in order to remove high-frequency fluctuations that were not related to respiration. After filtering, phase was calculated from the complex data, and the phase difference was calculated in each TR as the difference between the phase data of the TR and those of the first TR. The resulting phase difference data were averaged over the readout, generating an averaged phase offset of each navigator echo. Finally, the averaged phase offset was removed from non-navigator echo complex image data (1st to 16th echoes) that were acquired at the same TR.

To generate MWI, the multi-channel GRE k-space data were processed to create channel combined magnitude (root sum-of-squares of all channels after noise normalization) and phase (mean of complex images after phase offset correction) images (see Hammond et al. (2008) for more details). To reduce ringing artifacts, a Tukey window (parameter = 0.5) was applied in k-space. The resulting multi-echo complex images were fitted to a three-pool complex model (Nam et al., 2015), which contained frequency offset terms (Sati et al., 2013; van Gelderen et al., 2012; Wharton and Bowtell, 2012). An iterative non-linear curve-fitting algorithm (lsqnonlin function in MATLAB, tolx = 1e−5, tolfun = 1e−5) was used to estimate the model parameters. The myelin water fraction (MWF) was calculated as the ratio of the shortest T_2^* magnitude value to the sum of all three magnitude values.

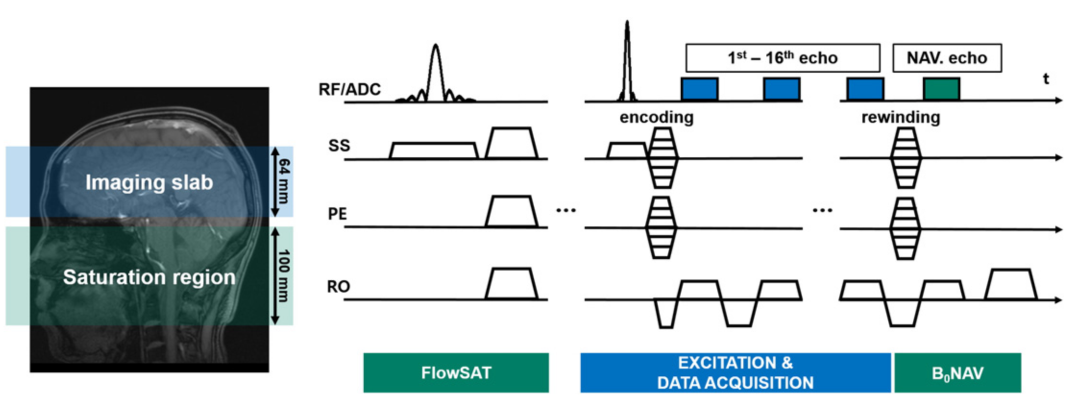

Fig. 1. Physiological noise compensation schemes for 3D GRE-MWI. Flow saturation RF pulses were applied at an inferior head area to saturate inflow signal reducing cardiac function induced inflow artifacts. Navigator echoes were used to compensate for respiratory function induced artifacts.