Multi-echo susceptibility-weighted imaging using an adaptive frequency mask

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Introduction: Compared to the single-echo SWI, multi-echo SWI results in improved SNR and enhanced contrast in small veins in SWI maps [1, 2]. While the frequency mask for multi-echo SWI can be generated using the same parameters as those used in conventional, single-echo SWI [1], independent frequency-mask (FM) generation from each echo results in improved results [3]. In a previous study [3], a frequency normalization method was proposed for generating a positive FM where the negative frequency values are set to unity and the positive frequency values are normalized by the highest frequency shift. However, because the highest or lowest frequency value in LFS maps may be very sensitive to noise or artefact, we developed a modified FM generation method where different values of the lowest frequency shift of interest (f_{th}) are calculated for each echo.

Adaptive Frequency Mask (FM) generation: The LFS - defined as the local Larmor frequency after subtraction of background field contributions - can be used to generate an element-combined, multi-echo SWI image. Because the conventional SWI mask used in single-echo SWI cannot be directly applied in the multi-echo case, we define an independent frequency mask for each echo. Specifically, the negative frequency mask at the j^{th} echo is defined as:

$$\begin{array}{ll} \text{if } f_\chi \geq 0 & FM = 1, \\ \text{elseif } f_\chi < 0 \text{ and } f_\chi \geq f_{th,j} & FM = (-f_{th,j} + f_\chi)/(-f_{th,j}), \\ \text{else} & FM = 0, \end{array} \qquad (1)$$

where f_χ is the frequency value of a pixel; $f_{th,j}$ (<0) is the minimum frequency shift of interest and is equal to $(1/(2\times TE_j))$ in order to satisfy a condition $2\pi\,|f_{th,j}|\times TE_j = \pi$. Note that Eq. (1) will be same as the frequency normalization method when $f_{th,j}$ is equal to the lowest frequency value.

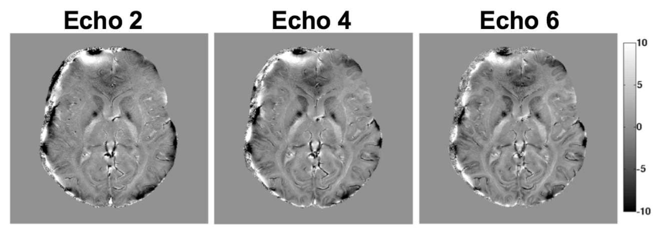

Fig. 1 Channel-combined LFS maps (Hz) of the central axial slice from three different echoes.

Methods: Data for phase image post-processing was acquired on a 7 T MR scanner with 16 independent RF transmit and receive channels. Imaging was performed using a 2D FLASH sequence (TR = 2 s., TE_1 = 4.56 ms, ESP = 4.41 ms, GRAPPA factor = 2, flip angle = 50^o, 6 echoes, 2 mm slice thickness, 40 slices, 0.5 mm in-plane resolution and 100 KHz readout bandwidth).

To generate local frequency maps, the complex images of individual channels at each echo were unwrapped by using the PUROR algorithm [4]. The resulting data was high-pass filtered to remove background fields. Specifically, a 2D Gaussian high-pass filter with full width at half maximum (FWHM) of 9.4 mm was applied to the Fourier transform of the unwrapped phase data to remove background fields [5]. The channel-combined LFS maps were then calculated using a trimming and weighting strategy from all channels at individual echoes.

Adaptive frequency masks were calculated using Eq. (1), where $f_{th,j}$ was set to -56, -37, -28, -23, -19 Hz for j varying from 2 to 6, respectively. Note that the first echo was excluded from the calculation due to its extremely low SWI contrast. When the standard SWI FM approach was implemented, $f_{th,j}$ was equal to the lowest frequency value in each frequency image for each echo. The number of frequency mask multiplications was four. Echo-combined SW images were generated from the mean of SW images from five echoes (echoes 2 – 6). Minimum intensity projections (mIPs) for each FM calculation technique were computed across 5 axial slices. Data processing was performed off-line using MATLAB.

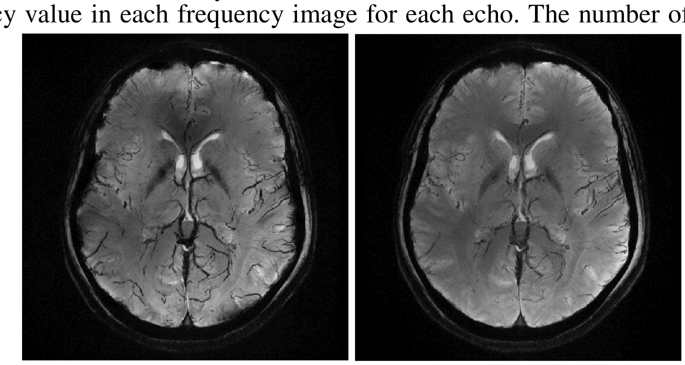

Fig. 2 mIP SW images (mIP over 5 axial slices) using the proposed method (left) and the standard frequency normalization method (right).

Results and Discussion: Figure 1 shows LFS maps from the central axial slice at three different echo times. The contrast in the LFS map at the longest echo is highest. When the frequency-normalization method was used, the mean ± standard deviation of the lowest frequency values from the 40 axial slices was -114 ± 34, -90 ± 37, -75 ± 31, -65 ± 27 and -65 ± 28 Hz for the 2^{nd} to 6^{th} echo, respectively. The absolute values of f_{th} used by the frequency normalization method are 2 to 3 times larger than the values used by the proposed adaptive method (values defined in the Method section). Additionally, the large standard deviations (~ 30 Hz) in the frequency cutoff value obtained from the frequency normalization method suggest that FM generation using this method is susceptible to noise or artifact existing in the LFS maps. FMs generated using this method may not accurately delineate small venous structures whose frequency shift is only slightly larger than that of structures in the brain parenchyma. Figure 2 displays the mIP SW images (mIP over 5 axial slices) generated by using the proposed method (left, Fig. 2) as well as the frequency normalization method from [3] (right, Fig. 2). The improved contrast between the veins and the surrounding brain parenchyma is apparent in the mIP images generated using the adaptive FM method.

Conclusion: The adaptive FM mask generation approach from Eq. (1) produces a more robust frequency mask than using the standard, single-echo frequency normalization method resulting in increased SWI contrast.

References: [1] Haacke et al, MRM 52: 612-618, 2004. [2] Wu et al, Neuroimage 59:297-302, 2012. [3] Luo, et al., Neuroimage 60: 1073-1082, 2012. [4] Liu and Drangova, MRM 68: 1303-1316, 2012. [5] Rauscher et al, MRI 26: 1145-51, 2008.