High-Quality Susceptibility Weighted Venography from Multi-echo MR dataset using Linear Phase Model

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Introduction

Susceptibility weighted imaging-based magnetic resonance venography (SWI-MRv) is important for diagnosis of venous diseases because it can noninvasively provide anatomical image of veins without ionizing radiation unlike x-ray or computed tomography venography. The SWI-MRv is generally obtained from gradient echo (GRE) sequence data [1]. However, single-echo venography cannot depict all veins because optimal TE is different depending on veins' thickness or alignment with the static field. Therefore, several techniques of SWI-MRv with multi-echo dataset have been researched for the optimal observation of all veins [2,3]. The authors also studied an effective denoising method for multi-echo gradient echo (MGRE) sequence's magnitude data without introducing blurring effects or any artificial appearance in order to acquire high-quality MRv for the veins not parallel to the static field of 3T at much longer TE than 28ms [4]. However, not only reduction of noise of magnitude data but also reduction of noise of phase data should be performed in order to obtain high-quality SWI-MRv using a multi-echo dataset because noisy phase mask can be made at relatively longer TE, which results in the degradation of SWI-MRv's quality although the phase data has higher signal to noise ratio (SNR) compared with the corresponding magnitude data [2]. The aim of this study is to acquire high-quality SWI-MRv at longer TE by reducing the noise of phase data on the temporal domain of 3D multi-echo datasets for obtaining high-SNR phase mask.

Methods

A 3D multi-echo GRE sequence data was acquired using a 3T siemens MRI system (Erlangen, Germany) with a 4 channel head coil for data acquisition. The field of view was 215×215×51.2 mm^3 with a matrix size of 512×512×32 voxels, and slice thickness was 1.6 mm. The repetition time=95 ms, flip angle=30˚, bandwidth=444 Hz/Px, First echo=5.67 ms, echo spacing=5.51 ms, and sixteen echoes were acquired. Total acquisition time was 25.95min.

Since the phase variation rate of the T2* weighted MR signal on temporal domain at each location is caused by field inhomogeneity or susceptibility, ΔB and this relation is represented by: $\Delta\emptyset = \gamma\Delta B TE$, where $\Delta\emptyset$ is phase. γ is gyromagnetic ratio and TE is echo time [1]. This equation means that the phase is linearly increased or decreased by the echo time. Therefore, a modeling for the phase can be used to reduce the noise on the temporal domain. The phase could be modeled with a first order linear curve function, and least squares algorithm which compares the linear function with the original signal iteratively and then chooses solution with least squares error was used as modeling method. We generated a negative phase mask at TE=38.73 ms to confirm denoising performance of the proposed method using a 128×128 Hamming filter and threshold=-0.2π [5]. The denoising effect of phase mask was visually compared and SWI-MRv was obtained from the denoised multi-echo dataset.

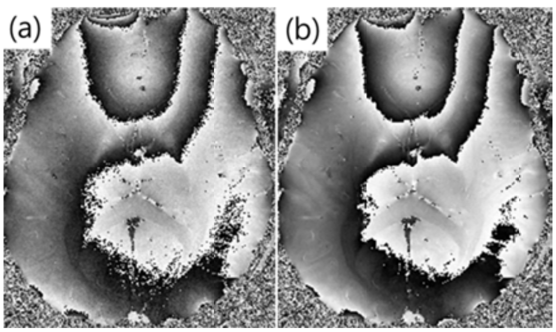

Fig1. Comparison of phase images of specific slice at TE=38.73ms (a) without denoising process and (b) with model-based denoising method.

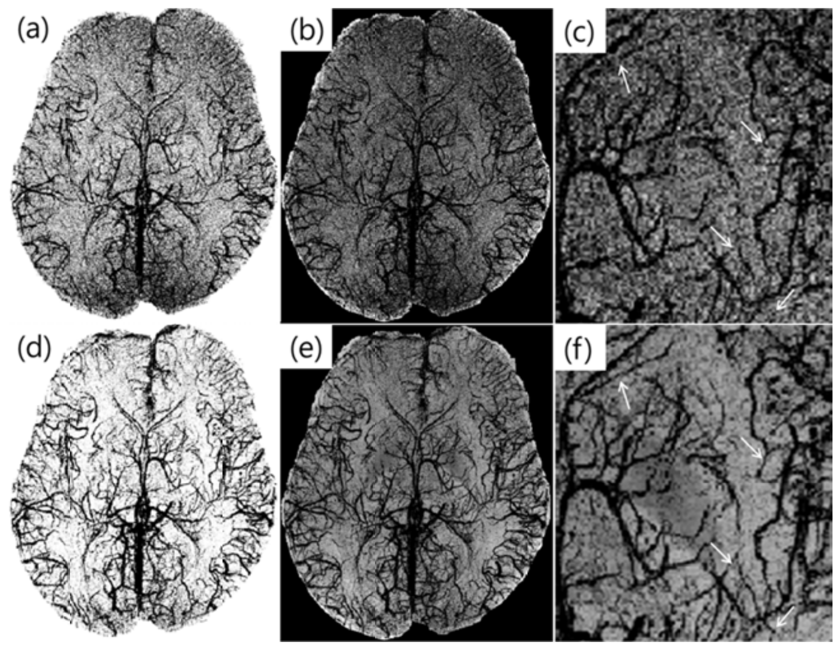

Fig2. The mIPs of the phase mask at TE=38.73ms (a) without denoising process and (d) with model-based denoising method. SWI-MRv (b) using Fig2-(a) and (e) using Fig2-(d) is showed and (c, f) its enlarged images, respectively.

Results

Fig1-(a) shows the original phase image of specific slice at TE=38.73 ms, which has low contrast of structural information such as veins and boundary of white-gray matters due to low SNR. Fig1-b is the denoised phase image of Fig1-(a) with model-based method. In this Figure, we can find that noise was effectively reduced and image's contrast was improved especially in the posterior.

Fig2-(a) is the minimum intensity projection (mIP) of original phase mask at TE=38.73 ms, which is very noisy in the area of normal tissue. Fig2-(d) is the mIP of the denoised phase mask with the proposed method at the same TE. Fig2-(b, c) and (e, f) are the SWI-MRv using Fig2-(a) and Fig2-(b), respectively. In Fig2-(f), the enlarged image of Fig2-(e), significant removal of noise was observed at normal tissue. The veins indicated by white arrows in Fig2-(f), which were barely seen in Fig2-(c), became distinct with clean contours due to the noise reduction.

Conclusion and Discussion

This study demonstrates that the linear model-based denoising method on the temporal domain can effectively reduce the noise on the images without significant spatial artifacts. High-quality SWI-MRv at 3T can be obtained by using the proposed denoising method.

Reference

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