High Resolution Magnetic Resonance Venography at 3 Tesla: Optimized Acquisition, Reconstruction, and Post-processing

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Introduction: Magnetic resonance venography (MRV) based on a susceptibility weighted imaging (SWI) approach [1,2] can be used for detailed study of intracranial venous vasculature. As many veins of interest, especially veins in the periventricular regions, are quite small (< 0.5 mm) it is essential to use high resolution imaging. The main problems with the existing high-resolution MRV techniques are the long scan time (20 minutes and longer) and poor visibility of small veins due to relatively large voxel volume (typically 0.25 mm^3 or larger). Such long scan times make MRV studies very vulnerable to motion artifacts. Further reduction in voxel volume is required to improve sub-millimeter vein visibility. To achieve this goal and keep scan time relatively short, all components of MRV (data acquisition, reconstruction, and post-processing) should be tuned. A composite method for MRV consisting of optimized data acquisition, image reconstruction, and post-processing has been developed. The method allows acquisition of high quality intracranial venograms with high small vessel visibility in clinically acceptable scan time (< 10 minutes).

Theory and Methods: SWI exploits differences in tissue magnetic susceptibilities to enhance image contrast. In the case of MRV, paramagnetic deoxyhemoglobin causes a local magnetic field inhomogeneity resulting in reduced T2* relaxation time of venous blood and phase difference between veins and surrounding tissues [1,2]. Scan parameters for MRV should be chosen such that both magnitude and phase contrast between veins and background tissues are maximized.

Imaging Protocol: More than 20 MRV studies of intracranial venous vasculature were performed on a 3 Tesla MRI scanner (Siemens Medical Solutions, Erlangen, Germany) using an eight-channel head coil (MRI Devices, Waukesha, WI) and a T2*-weighted gradient echo pulse sequence with flow compensation. The optimal scan parameters found from these studies were: TR/TE=38.5/22 ms, flip angle=12°, FOV= 205x153.8 mm, 80 slices with 0.6 mm thickness, imaging matrix=512x384x104, bandwidth=40 Hz/pixel, and scan time=9:15 minutes. Such a short scan time was achieved using partial Fourier sampling (80%) in both phase-encoding directions and parallel imaging with reduction factor R=2 and 32 reference k-space lines. Voxel size for the described protocol is 0.4x0.4x0.6 mm resulting in 0.096 mm^3 voxel volume. In cases, when larger volume coverage was required, parallel imaging with R=3 and 36 reference lines was applied and/or slice thickness was increased to 0.8 mm. The flip angle value used in this protocol was chosen such that CSF intensity was comparable to that of brain tissues to guarantee high contrast between veins and the surrounding tissues. About 70% of the imaging studies were done approximately 10 minutes after injection of 0.1 mmol/kg of Gd-DTPA extracellular contrast agent (Omniscan, Amersham Health, Princeton, NJ).

Reconstruction: Measurement (raw) data were saved and processed offline using custom software developed in MATLAB (MathWorks, Natick, MA). First, a complete k-space dataset for each coil element was reconstructed from the undersampled data using the GARSE algorithm [3]. GARSE was used instead of the standard GRAPPA algorithm [4] due to the better quality of the GARSE reconstructed images (higher SNR and complete de-aliasing), especially for the studies with R=3. Because MRV exploits phase information to enhance contrast between veins and brain tissues, a composite complex image should be constructed from individual coil images. To achieve this goal low frequency phase variations were removed from individual coil images using the method described in [1,2]. This operation removes coil specific phase factors but preserves phase differences between veins and surrounding tissues. In our implementation, this operation also compensates for undersampling of high frequencies in the phase-encoding directions. The composite complex image was constructed by combining the processed coil images as:

$$I(r) = \frac{\sum_i I_i(r)|I_i(r)|/\sigma_i^2}{\sqrt{\sum_j I_j^2(r)/\sigma_j^2}}$$

where σ_i is the standard deviation of the noise in i-th coil image. This parameter was estimated for each coil from a signal free image region. Finally, zero-filled interpolation was utilized to achieve isotropic voxel dimensions (0.2 mm) and to the reduce partial volume effect resulting in better small vein visibility.

Post-processing: A filter utilizing the phase of the combine image was constructed in the following form:

$$F(r) = \begin{cases} (1+\phi(r)/\pi)^{\alpha}, & \phi(r) < 0 \\ (1-\phi(r)/\pi)^{\beta}, & \phi(r) > 0 \end{cases}$$

with α=4 and β=1. The filter was applied to the magnitude of the combined image to enhance contrast between venous structures and the surrounding tissues [1,2]. The resulting image volume was subjected to an additional post-processing step based on a median filter [5] to improve vessel visibility in the re-projection images constructed by the minimum intensity projection algorithm (MinIP).

Results: Typical results from the application of our method are shown in Figs. 1 and 2. Figure 1 demonstrates the utility of both post-processing steps. The phase filtering improves small veins visibility by enhancing contrast between veins and the surrounding tissues (Fig. 1b). However, the filter can have a detrimental effect in areas of fast phase variations and low signal intensity as is shown in this given example. Median filtering can be used to resolve these effects and to allow construction of MinIP reprojection images from 3D image volume without prior segmentation of tissue free image regions. Figure 2 illustrates the ability of our composite technique for imaging sub-millimeter veins. Venous vasculature in the periventricular space is depicted with exclusive details.

Discussion: The quality of MRV images depends not only on the data acquisition protocol but also on subsequent image reconstruction and post-processing steps. To achieve high quality, high-resolution MRV, all of these components should be optimized accordingly. If phase filtering is to be used in post-processing, the data acquisition protocol should be adjusted to optimize the phase contrast between veins and background tissues, and image reconstruction should be implemented such that this phase information is preserved and enhanced. A time efficient data acquisition protocol with optimized image reconstruction and post-processing has been developed for high-resolution intracranial MRV at 3 Tesla. The preliminary results indicate feasibility of the proposed method for clinical studies.

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References: [1] Reichenbach JR, et al. JCAT 2000;24:949-57. [2] Haacke EM, et al. MRM 2004;52:612-8. [3] Kholmovski EG, et al. ISMRM 2005, p. 2672. [4] Griswold MA, et al, MRM 2002;47:1202-10. [5] Alexander AL et al. MRM 2000;43:310-3.

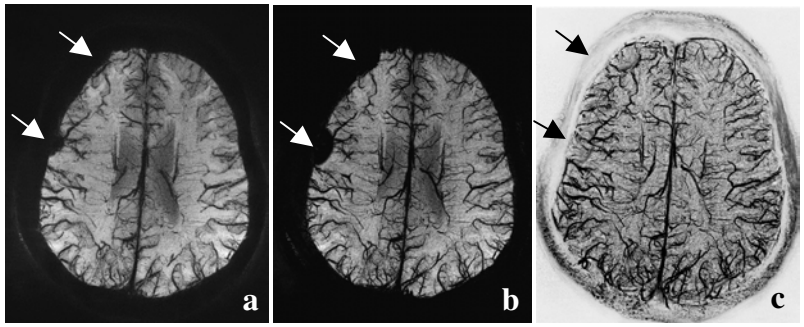

Figure 1. Venograms constructed from 2.4 cm thick image volume: (**a**) without post-processing, (**b**) after phase filtering, (**c**) after phase and median filtering. The regions of fast phase variations are indicated by arrows.

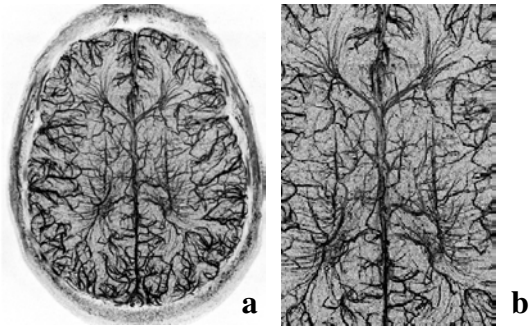

Figure 2. High-resolution venogram constructed from 1.6 cm thick image volume: (**a**) the complete view, (**b**) the magnified view of the central part of the brain.