

Reducing Artifacts in SWI based MR Venography - Post Processing Technique to Compensate for the Signal Loss

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Introduction: In susceptibility weighted imaging (SWI) based MR venography, veins experience much more signal losses than other surrounding tissues due to intrinsic magnetic susceptibility differences between the tissues and the veins [1]. This feature allows the veins to be clearly visible through minimum intensity projection (mIP). However, the susceptibility difference at air-tissue interface creates undesired macroscopic field inhomogeneity causing additional signal degradations [2], which eventually produces artifacts for longer TE. These artifacts are especially noticeable around the orbito-frontal cortex where higher field inhomogeneity is expected (mainly due to the hollow space of the nostril). Field inhomogeneity along the longitudinal direction (hereinafter called the z direction) is mainly responsible for the additional signal degradation since the voxel is asymmetrically longer in z direction [3] and thus shall be only concerned. In this study, backtracking method is chosen to reduce the artifacts: the amount of field inhomogeneity is calculated with the acquired signals and then each signal is compensated afterwards according to this field inhomogeneity value.

Theory: Magnetic gradient field along the longitudinal direction is calculated according to the following equation [4]:

$$G_z = \frac{\angle([S_a(t) \cdot S_a(t + \Delta t)^*] \cdot [S_b(t) \cdot S_b(t + \Delta t)^*])}{2 \cdot \Delta t \cdot \gamma \cdot z_0}$$

This equation accounts for the fact that the local field gradient (LFG) along z direction causes additional phase shift over time to the signals that are exposed to higher fields, where S_a and S_b denotes the signal at location a and b respectively,

with spatial distance z_0 . Δt equals the time deviation. If there were no magnetic field gradient along z direction, signals' phase at position a and b would remain identical during Δt . However, in reality, due to the field inhomogeneity along z direction, relative phase is not same.

It is assumed that the dephasing caused by the field gradient (G_z) contributes to an additional exponential decaying of the signal, $M(z, TE) = M_0(z) \exp(-i \gamma G_z \Delta t)$. In 2D spin warp imaging, the modified NMR signal is given by: $S(K_z=0, TE) = m \cdot \exp(-R_2^* TE) \cdot z_0 \cdot \text{sinc}[\gamma / 2 G_z TE z_0]$, where exponential term accounts for the attenuation of the signal by T_2^* relaxation [5]. With background gradient, sinc term is now present, which additionally attenuates the signal therefore disrupting signals' exponentially decaying manner.

Methods: Multi-echo SWI data were acquired on a 3T scanner with a matrix size of 256x256x50, a FOV of 20.4cmx20.4cm, and a slice thickness of 1.6mm. 5 sets data of different echo time (3.3ms, 6.31ms, 9.32ms, 12.33ms and 15.34ms) are used to create the local field gradient map. Total 10 different maps are created with 10 different pairs of data (3.3ms-6.31ms, 3.3ms-9.32ms, 3.3ms-12.33ms, 3.3ms-15.34ms, 6.31ms-9.32ms, 6.31ms-12.33ms, 6.31ms-15.34ms, 9.32ms-12.33ms, 9.32ms-15.34ms, 12.33ms-15.34ms). The echo time for the image that is reconstructed equals 60.49ms. Long TE was used for high vein contrast. With the LFG value previously achieved, raw signals are corrected by removing the sinc term thereby canceling out the effect of the gradient field.

Results & Discussion: Figure 1 shows the mIP of the slice number from 15 to 40 for echo time 60.49ms, which contains the susceptibility artifacts around the orbito-frontal cortex. Figure 2 shows the MIP of average local field gradient map (LFG). Mean value of 10 different LFG maps is derived in order to remove out the effect of the noise term for further processing. The orbito-frontal region reflects the gradient, which also coincides with the fact that this is the region with severest artifact. Figure 3 shows the mIP of corrected images. For each voxel, signal degradation is compensated according to each local gradient value. The restored image not only maintains the sufficient information about the veins but also the artifacts are significantly reduced as well. The structural shape of the brain is also restored.

Conclusion: With the local gradient map generated after the signal acquisition and by correcting each signal term according to the gradient value, the image significantly showed reduced artifacts. This method provides much enhanced image quality without any cost of additional acquisition time.

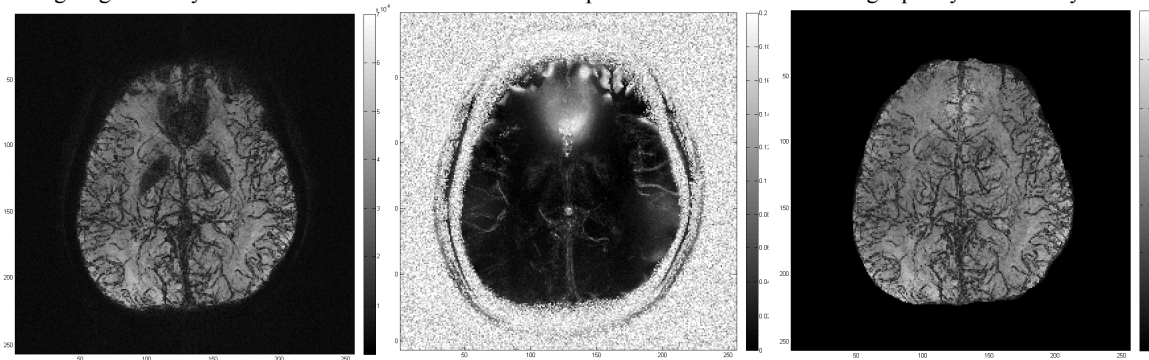


Figure 1 mIP of TE=60.49ms

Figure 2 MIP of Average LFG map

Figure 3 mIP of corrected image at TE=60.49ms

Acknowledgments: Fund by Basic Research Program of the Korea Science and Engineering Foundation (R01-2008-000-20270-0)

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