

Improved visualization of the subthalamic nuclei by reducing susceptibility induced signal losses in T2* weighted multi-gradient-echo images

S. Volz¹, E. Hattingen², C. Preibisch^{1,3}, T. Gasser⁴, and R. Deichmann¹

¹Brain Imaging Center, University Frankfurt, Frankfurt, Hessen, Germany, ²Department of Neuroradiology, University Hospital, Frankfurt, Hessen, Germany, ³Abteilung für Neuroradiologie, Klinikum rechts der Isar der TU München, München, Germany, ⁴Department of Neurosurgery, University Hospital, Frankfurt, Hessen, Germany

Introduction T2*-weighted gradient echo (GE) images show a good contrast in iron-rich structures like the subthalamic nuclei (STN) due to microscopic susceptibility induced field gradients with the additional advantage of low specific absorption rate (SAR) exposure [1]. They are therefore useful in Parkinson disease treatment, providing landmarks for exact placement of stimulation electrodes [2, 3]. However, T2*-weighted images are also sensitive to macroscopic field inhomogeneities, resulting in signal losses e.g. in orbitofrontal and temporal brain areas [1, 4, 5] which reduce the anatomical information content. Therefore, on one hand long echo times (TE) are required for achieving a good T2* contrast, on the other hand the extent of signal losses increases with TE. In this work we present an image correction method for multi-echo GE data, consisting of two steps: (1) signal loss correction by evaluation of the phase information and (2) weighted image combination for optimal T2*-contrast and signal-to-noise ratio (SNR) in the deep brain as well as low image degradation in areas affected by macroscopic field inhomogeneities.

Materials and Methods Measurements were performed on a 3T whole body MR scanner using a receive-only 8-channel array head coil and the whole body transmit coil. A multi-echo GE sequence (8 echoes, TE = 10/16/22/28/34/40/46/52ms, TR = 900ms, 15 slices, bandwidth = 300Hz/Px, imaging matrix 256*256, in plane resolution 1mm and slice thickness 2mm, total measurement time = 3min 30s) with modulus and phase image output was used. For excitation, an exponentially shaped RF pulse with the time profile $A(t) = \exp(-4 \cdot \text{abs}(t)/(P/2))$ with the pulse duration $P = 2\text{ms}$ and $-P/2 \leq t \leq P/2$ was used [6].

Macroscopic gradient maps G_{susc} were calculated from the phase difference between the two echoes with the shortest TE and smoothed via convolution with a 5*5 voxel kernel for noise reduction. Processing of magnitude images was performed in 2 steps (Fig. 1):

1.: Images were intensity corrected by pixelwise division by $A(G_{\text{susc}}/G_s \cdot \text{TE})$, $A(t)$ being the RF-pulse shape and G_s the slice selection gradient [6].

2.: A combined image was then created as a weighted sum of the intensity corrected images acquired at different TE where the weighting factors were calculated pixelwise for each TE in dependence on G_{susc} , in general preferring long TE for achieving good T2* weighting, but short TE in the presence of field inhomogeneities to reduce signal losses.

The SNR was measured in STN, white matter (WM) and red nucleus (RN), both in the original, uncorrected images and in the combined images.

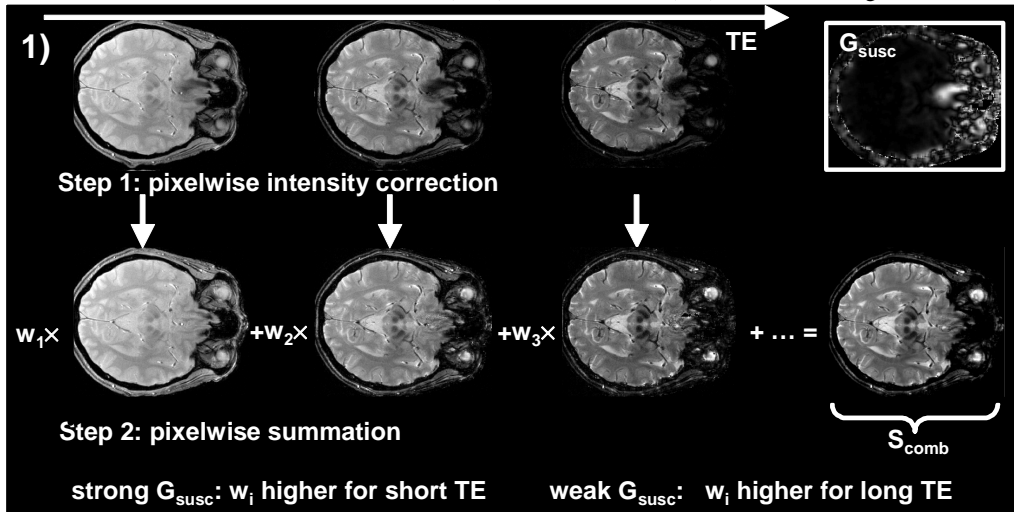


Fig 1: Data processing is performed in two steps: First, a pixelwise intensity correction for each TE is performed using the gradient maps G_{susc} calculated from the phase images. Second, the corrected images are combined by calculating a weighted sum.

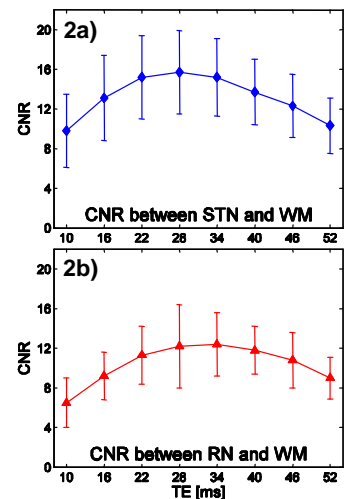


Fig 2: CNR between STN and surrounding WM (a) and between RN and surrounding WM (b) in dependence on TE for the original, unprocessed images. The results represent average values and standard deviations across 8 subjects.

Results The intensity correction improves the image quality in areas affected by macroscopic field inhomogeneities (Fig. 1, bottom) in contrast to the original images (Fig. 1, top). The weighted combination of intensity-corrected data yields good T2* contrast in the deep brain and does not suffer from signal losses in critical brain areas (Fig. 1, bottom, right). A quantitative ROI analysis of the contrast-to-noise ratio (CNR) between STN/RN and the surrounding WM yields a maximum CNR of 15.7 ± 4.2 at TE = 28 ms (STN-WM) and 12.4 ± 3.2 at TE = 34 ms (RN-WM) (Fig. 2) for the original, uncorrected images. For the combined images the respective CNR values were 37.1 ± 9.5 (STN-WM) and 30.0 ± 6.7 (RN-WM) corresponding to a CNR increase by a factor of about 2.4 as compared to the original images with the highest CNR.

Discussion It could be shown that the presented postprocessing method allows the calculation of a single intensity corrected, combined image data set from 8 individual GE images with increasing TE, providing good T2*-contrast in areas with iron containing structures like the STN in the midbrain and compensating for signal losses in areas with strong macroscopic field gradients. Given the image resolution of $1 \times 1 \times 2 \text{mm}^3$ the resulting image data set is therefore suitable both for targeting the STN and for fusion with other data. The measurement time of 3min 30s for 15 slices is acceptable in a clinical environment. One drawback is the lack of homogeneous T2* contrast across the whole image volume, because the weighting depends on the susceptibility induced field gradient strength. However, this does not affect regions in the midbrain where local susceptibility gradients are relatively weak. Only in areas with very strong susceptibility gradients like orbitofrontal cortex and temporal lobes T2* contrast is reduced for preserving structural information.

References

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