Evaluating The Accuracy of Susceptibility Maps Calculated from Single-Echo versus Multi-Echo Gradient-Echo Acquisitions

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Synopsis

For Susceptibility Mapping (SM), Laplacian-based methods (LBMs) can be used on single- or multi-echo gradient echo phase data. Previous studies have shown the advantage of using multi-echo versus single-echo data for noise reduction in susceptibility-weighted images and simulated data. Here, using simulated and acquired images, we compared the performance of two SM pipelines that used multi- or single-echo phase data and LBMs. We showed that the pipeline that fits the multi-echo data over time first and then applies LBMs gives more accurate local fields and \( \chi \) maps than the pipelines that apply LBMs to single-echo phase data.

Introduction

For Susceptibility Mapping (SM), Laplacian-based methods (LBMs) can be used to perform unwrapping or background field removal of single- or multi-echo gradient echo (GRE) phase data. In previous studies on SM-LBM pipelines applied to multi-echo phase data, we have shown that fitting the multi-echo GRE phase over echo times (TEs) before applying LBMs gives more accurate local field (\( \Delta B_{loc} \)) and magnetic susceptibility (\( \chi \)) estimates than applying LBMs first and averaging over TEs. Previous susceptibility weighted imaging (SWI) studies on healthy volunteers and SM studies using simulations have shown that using multiple TEs reduces the propagation of phase noise into the SWI and \( \chi \) images compared to using single-echo acquisitions. Here, we aimed to investigate the effect of using LBMs on single-echo phase data compared to multi-echo fitting on both \( \Delta B_{loc} \) and \( \chi \) accuracy. We tested two LBM-SM pipelines on simulated complex data and, unlike previous SM studies, images of healthy subjects.

Methods

Multi-echo 3D GRE brain images of four healthy volunteers were acquired on a Philips Achieva 3T system (Best, NL) using a 32-channel head coil and 5 echoes, TE/ΔTE=3/5.4 ms, 1-mm isotropic resolution, TR=29 ms, FoV=240x180x144 mm³, SENSE acceleration factor in the first/second phase-encoding direction=2/1.5 and 20º flip angle.

Multi-echo (5 echoes, TE/ΔTE=3/5.4 ms) complex images were simulated from ground-truth \( \chi^{true} \) (Zubal head phantom6), \( M_0 \) and \( T_2^* \) distributions (Fig. 1), using a Fourier-based forward model7 of the total field perturbation \( \Delta B. M(t) = M_0 \exp(-t/T_2^*) \) and a constant phase offset \( \phi_0 = \pi/4 \). Random Gaussian noise (mean = 0, standard deviation (SD) = 0.03) was added to the real and imaginary parts of the complex images to give a realistic signal-to-noise ratio. Ground-truth local field perturbation \( \Delta B_{loc}^{true} \) was calculated using the reference scan method9.

\( \Delta B_{loc} \) was calculated using two distinct pipelines on the phantom and volunteers’ images: 1) Multi-echo (ME): non-linear fit of the complex signal over TEs; \( \Delta B_{loc}^{ME} \) calculation using SHARP10 (\( \sigma = 0.05 \), BET11,12 brain mask with 2- (phantom) or 4- (volunteer) voxel erosions); 2) Single-echo (SE): at each \( T_E \), \( \Delta B_{loc}^{SE} \) calculation using SHARP10 (\( \sigma = 0.05 \), BET11,12 brain mask calculated from the i-th echo and eroded as in ME) on \( \phi(T_E) \), and dividing by \( T_E \), and \( \gamma \), the gyromagnetic ratio. \( \chi^{ME} \) and \( \chi^{SE} \) (for \( i = 1 \) to 5) were calculated using TKD13 (\( \delta = 2/3 \) and correction for \( \chi \) underestimation19).

In the phantom, the accuracy of \( \chi \) was assessed by calculating means and SDs in the regions in Fig. 1 and Root Mean Squared Errors (RMSEs) in the brain relative to \( \chi^{true} \). In the volunteers, \( \chi \) means and SDs were calculated in the regions in Fig. 1. All regions except VN were segmented based on the Eve \( \chi \) atlas14, which was aligned to the fifth-echo magnitude image (TE=24.6/24 ms) using a combination of rigid, affine and non-affine transformations15,16. VN was segmented using the Multiscale Vessel Filtering method17 (scales=4, probability threshold for vein segmentation = 0.5).

Results and Discussion

The effect of noise on the SE images decreased with increasing TE (Figs. 2c-g, 3b-f, 4b-f and SDs in Fig. 5), in line with the known relationship of the phase contrast-to-noise ratio with time: contrast-to-noise is maximised at \( T_E = T_2^* \). In the phantom, mean \( \chi^{SE}_{1,4} \) were similar to mean \( \chi^{ME} \), but suffered from the greater noise at short TEs and had larger RMSEs than \( \chi^{ME} \) (Fig. 2).

In the phantom, ME gave the most accurate \( \Delta B_{loc} \) and \( \chi \) estimates (Figs. 2 and 5a). High-\( \chi \) structures, e.g. the SSS, showed the largest susceptibility errors in SE images, and were visible in the difference images, even at longer TEs (Figs. 2c-g). Susceptibility errors were also most prominent in the volunteers’ VN (Figs. 3b-f and 4b-f).

In the volunteers, \( \chi^{ME} \) and \( \chi^{SE}_{1,4} \) were approximately the same in the GP and PU (Fig. 5). However, in the other regions, only \( \chi^{ME} \) always gave average values consistent with the literature15. In particular, \( \chi^{ME} \) was always negative in the PCR, which is expected to be about 0.02 ppm more diamagnetic than water13,18. Furthermore, in the VN, \( \chi^{ME} \) was always the closest to \( \chi = 0.46 \) ppm, which is the expected value of \( \chi \) in veins (at 70% oxygenation)19.

Conclusions

Fitting the multi-echo phase over TEs before applying LBMs gives more accurate \( \Delta B_{loc} \) and \( \chi \) maps than applying LBMs to single-echo phase images, particularly for high-\( \chi \) structures.

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References


Figures

Fig. 2 In the phantom, the ground-truth $\Delta B_{loc}$ (top row, column a) and $\chi$ images (middle row, column a) were subtracted from the same slice of the local field and susceptibility map respectively calculated by the ME (b), SE1 (c), SE2 (d) SE3 (e), SE4 (f) and SE5 (g) pipelines. The arrows in the images point at the SSS. The bottom row shows the whole-brain RMSEs of $\chi$.

Fig. 3 In each healthy subject, $\Delta B_{loc}$ calculated by the ME pipeline (column a) was subtracted from the same slice of the local field calculated by the SE1 (b), SE2 (c), SE3 (d) and SE4 (e) pipelines. The arrows in the images point at large veins in the VN.

Fig. 4 In each healthy subject, $\chi$ calculated by the ME pipeline (column a) was subtracted from the same slice of the susceptibility map calculated by the SE1 (b), SE2 (c), SE3 (d) and SE4 (e) pipelines. The arrows in the images point at large veins in the VN.

Fig. 5 Mean and SDs of $\chi$ calculated using all the echoes (ME pipeline) or single echoes (SE pipeline) in the phantom (a) and the healthy subjects (b-e). In the phantom (a), the estimated $\chi$ values are compared to the ground-truth.