A Novel Approach for Separation of Background Phase in SWI Phase Data Utilizing the Harmonic Function Mean Value Property

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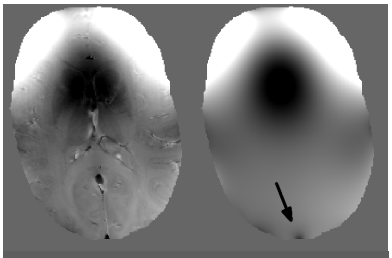
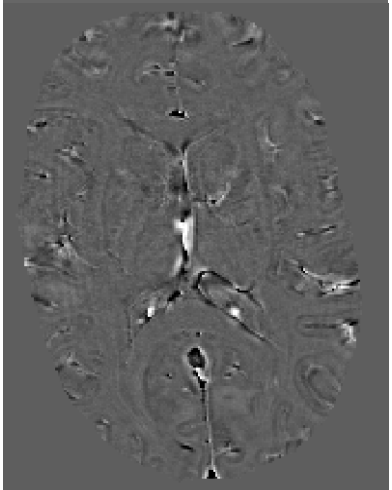
INTRODUCTION – Magnetic resonance phase data directly reflects local magnetic field changes and is used for shimming procedures, interpretation or correction of artifacts, and as anatomical contrast complementary to magnitude images [1]. Most of these applications require separation of local phase and background phase contributions that result from air-tissue and bone-tissue interfaces, eddy currents, temperature fluctuations or gradient deviations. Although several heuristic pre-processing methods [1, 2] have been established for this non-trivial task, it has been shown that these methods exhibit significant concomitant degradation of local phase information [3]. In this contribution, we present, for the first time, a non-heuristic, parameter-free approach for high-precision separation of local phase and background phase contributions for *in vivo* SWI-data.

THEORY – Static magnetic fields and corresponding phase maps are harmonic functions in regions of homogeneous susceptibility, i.e., they satisfy Laplace's equation $\Delta u = 0$ [4]. In regions with heterogeneous susceptibility the field is non-harmonic (discontinuous). Thus, phase data may be decomposed in harmonic (h) and non-harmonic (nh) components $\varphi = \varphi_{\mathrm{h}} + \varphi_{\mathrm{nh}}$, based on the principle of linearity. Background fields are supposed to be harmonic since their sources are located outside of the region of measurable phase. The harmonic component has the property that its mean value over a (normalized) spherical shell S is equal to the value φ_{h0} at the center of the shell: $\varphi_{\mathrm{h0}} = \langle\varphi_{\mathrm{h}}\rangle_S = (S \otimes \varphi_{\mathrm{h}})_0$ [5], where $\otimes$ denotes the convolution operator. Harmonic components may therefore be eliminated from phase data by subtracting the spherical mean value (SMV): $\varphi\prime = \varphi - S \otimes \varphi = \varphi_{\mathrm{nh}} - S \otimes \varphi_{\mathrm{nh}}$ [6]. The second term $S \otimes \varphi_{\mathrm{nh}}$ generally does not vanish in heterogeneous media and may degrade $\varphi\prime$ significantly. However, reconstruction of the non-harmonic component from the filtered data may be performed by deconvolution using the kernel $\delta - S$, where δ is a Dirac at the center of the sphere: $\varphi_{\mathrm{nh}} = (\delta - S) \otimes^{-1} \varphi\prime$.





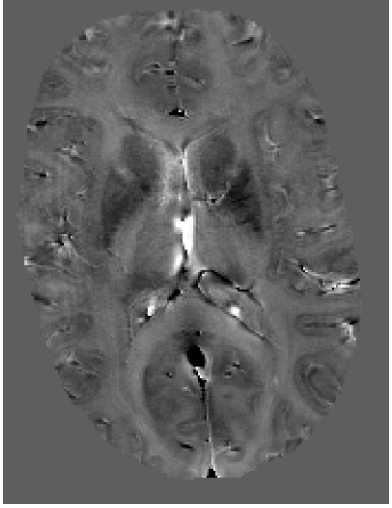


FIG. 2. Representative slice of *in vivo* data shows unwrapped MR phase (top-left), SMV-subtracted phase (middle), and deconvolved phase (bottom). Subtracting (bottom) from (top-left) reveals the background field shown in (top-right). Figures (top) and (middle,bottom) are windowed as [-2,3] and [-0.7,1.2] rad from black to white, respectively.

MATERIALS AND METHODS

Data Acquisition and Pre-Processing: High-resolution volunteer data of the whole brain were acquired with a 3D fully flow compensated gradient-echo sequence (TE/TR/FA/BW=20ms/30ms/15°/80 Hz/px, voxel size 0.6×0.6×0.6 mm³, 75% PF in phase and slice encoding direction) on a 3T MR-scanner (Tim Trio, Siemens Medical Solutions) using a 12-channel head-matrix coil. Multi-channel phase images were combined using uniform sensitivity reconstruction [7] and 3D phase unwrapping [8] was applied to resolve phase aliasing.

Numerical Model: A detailed numerical anatomical brain model was created from T1-weighted volunteer data (1x1x1 mm³) via automatic segmentation (FreeSurfer, Athinoula A. Martinos Center for Biomedical Imaging, Massachusetts General Hospital, Charlestown, MA, USA) with manual segmentation of venous vessels. To include field contributions from the torso the brain model was embedded into the skull of a human whole-body model of The Virtual Family (Duke, 1x1x1 mm³, [9]). Reasonable magnetic susceptibilities were assigned to each of the 55 anatomical regions and the field perturbation was computed by fast forward-field calculation [10] and uniformly distributed noise was added. A reference model without background distortions was generated from brain-tissue compartments of the numerical model (without skull and bone) embedded in parenchyma. Based on the simulated fields phase accumulation was calculated for B_0·TE=60 ms·T.

Processing: The SMV (spherical shell; radius/thickness=3/1 voxels) was calculated for each voxel and subtracted from the phase data. Reconstruction of the non-harmonic components from reliable phase data was facilitated by a Krylov subspace method [11] in spatial domain.

RESULTS – Figure 1 and 2 reveal representative slices of intermediate and final results for the numerical model and the *in vivo* data, respectively. After subtraction of the SMV the data suffers from significant degradation of local phase information similar to high-pass filtering of the phase (Fig. 1-top-right and 2-middle). Via deconvolution local phase information was reconstructed (Fig. 1-bottom-left and 2-bottom). The difference between deconvolved phase and input phase reveals the background phase (Fig 2-top-right) which is totally smooth and without fine-scale features. The difference between the reconstructed model data and the reference field is almost flat (Fig 1-bottom-right) indicating perfect estimation of the background phase.

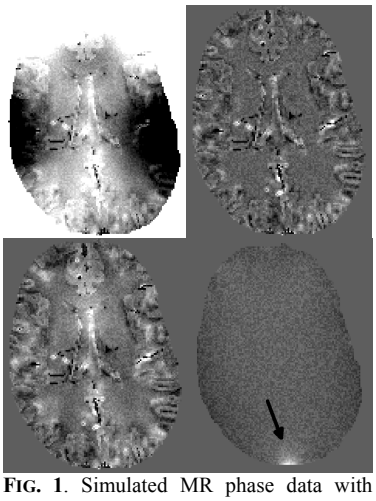


FIG. 1. Simulated MR phase data with background fields (top-left) was post-processed by subtracting SMV (top-right) and deconvolved (bottom-left). The difference between the reference model (not shown) and (bottom-left) is depicted in (bottom-right). Figures (top-left) and (top-right, bottom) are windowed as [-4,-2] and [-0.7,1.2] rad from black to white, respectively.

DISCUSSION AND CONCLUSIONS – A novel technique is suggested for suppression of background field inhomogeneities and maintenance of susceptibility-related local tissue-information. Small deviations in the model data may arise from the definition of the reference phase and cropping of the demagnetization field of the sinus sagittalis (Fig. 1-bottom-right and 2-top-right, arrow). Homogeneity of the calculated background phase indicates that even non-susceptibility related tissue-phase effects, e.g., due to local electronic screening or microscopic anisotropy [12], were maintained. Accepting slightly reduced quality, the presented technique may be implemented very efficiently employing direct inversion in Fourier domain. The presented approach for background field compensation is cutting edge for high-precession quantitative phase analysis (e.g., mapping of magnetic susceptibility [13] or analysis of absolute phase values [2]).

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