Removal of Background Fields with Spatially Variable Kernel Radii Guided by the Frequency-Offset-Gradient (FOG) Magnitude

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Target Audience: Researchers who are interested QSM, Phase, SHARP, Background Field Removal, Phase Unwrapping.

Purpose In the last decade MR-imaging based quantitative mapping of tissue-magnetic susceptibility, Quantitative Susceptibility Mapping (QSM), has evolved as a novel diagnostically useful imaging method. It derives susceptibility maps from MR phase data by 3 basic steps of phase unwrapping, background-field removal, and dipolar inversion. A recently developed variant of the algorithm, SDI[1], combined the first two steps and decreased processing times into ranges that are practical in clinical settings. It involves a Laplacian-based phase unwrapping[2] that requires a computationally efficient convolution and deconvolution only, and exploits the harmonic properties of the magnetic field via the spherical-mean value (SMV) theorem. Multiplication of the data with an eroded tissue mask before deconvolution is an important and critical step. Stronger erosion might lead to better QSM images at the cost of lacking information near brain borders, which in turn is defined by the mask segmentation. The segmentation is typically particularly problematic in regions with large field inhomogeneity, particularly in studies involving inhalation of paramagnetic hyperoxic gas, or with low-SNR.

Here we investigate the possibility of keeping the erosion of the binary mask at a minimum and recovering as much brain tissue in the susceptibility maps as possible, while maintaining sufficient image quality by a regional adaptation of the radius of the Laplacian convolution kernel. For the first time, we use the magnitude of the Frequency Offset Gradient[3] (FOG) as a measure of susceptibility-induced interference to guide the selection of the kernel radius – increasing the kernel size in regions of larger FOG

Materials and Methods **MRI**: Real and imaginary images of six consenting volunteers who inhaled 100 % oxygen were acquired on a 3T MR system (Ingenia, Philips Healthcare, Best, The Netherlands), with the following parameters: 3D Multi-Gradient-Echo (5TE, ΔTE=16ms, TE_1 = 4ms, TR = 71, FA = 50, 1.8 mm isotropic voxels).

Frequency Offset Gradient (FOG): Inter-voxel frequency gradients were calculated along x, y, and z-directions, using the equation: FOG = Δθ / 2πΔrTE and FOG magnitudes were calculated as mean square of the sums. Mean (FOG_{mean}) and standard deviation (FOG_{std}) of whole brain FOG values were calculated and two binary image masks were generated by thresholding: FOG-MASK[1] = FOG > (FOG_{mean} + 3*FOG_{std}) and FOG-MASK[2] = FOG > (FOG_{mean} + 5*FOG_{std}). Apart from the brain borders, the regions included by the FOG-masks were mainly in the vicinity of air-cavities, where rapid T2* decay occurs. The peripheral regions close to the brain borders, other than frontal cavities, were removed via multiplying with a highly eroded version of the mask.

Data Processing: Multi-echo complex data from the scanner were used to reconstruct combined phase images for all experiments according to $WPI_w = angle\,(\frac{1}{N-1}\sum_{n=1}^{N-1} S_n^* S_{n+1})$. Combined phase data were unwrapped and subject to background-field removal by the Laplacian-based SHARP[1] (threshold = 0.1) method, using the relation: $K(WPI) = \cos(WPI_w).K(\sin(WPI_w)) - sin(WPI_w).K(cos(WPI_w))^2$. Here, "*K*" represents the Laplacian kernel, whose size was regionally varied, depending on the FOG magnitude. The default kernel radius was 5 voxels, whereas in areas above the specific FOG threshold a radius of 7 voxels was chosen. Background noise and convolution artifacts were reduced by element-wise multiplication with the eroded binary whole-brain mask (BET[4] (default threshold = 0.5), eroded by 5 voxels). SHARP images were obtained by convolution with the inverted respective kernels. Quantitative susceptibility, ΔX, maps were obtained from the SHARP images by inversion, according to $\Delta X = FT^{-1}(\frac{FT(-SHARP/(\gamma B_0 \Delta TE))}{g})$, $g = \frac{1}{3} - \frac{k_z^2}{k^2}$, $k^2 = k_x^2 + k_y^2 + k_z^2$, where FT = Fourier Transform, γ = gyromagnetic ratio, B_0 = field strength, ΔTE = echo-time increment. The values around the magic angle of dipole kernel, g, was thresholded and regularized in order to avoid division by zeroes[1, 5]. The resulting susceptibility maps were compared to those obtained with a traditional Laplacian-based SHARP approach, using the same Laplacian-kernel with a radius of 1 voxel throughout the imaged volume.

Results An exemplary axial FOG image is shown n Fig. 1 (left), with the corresponding threshold-dependent binary FOG-masks (right). Shortest-TE magnitude images are shown in Fig. 2a and b, along with susceptibility maps calculated with the traditionally constant Laplacian kernel (Fig. 2c and d), and with a regionally adapted kernel radius (Fig. 2e and f). A histogram of the QSM values in the region of interest (ROI) marked with a red rectangle in Fig. 2d) is plotted in Fig. 3 (a: constant kernel, b: regionally adapted kernel).

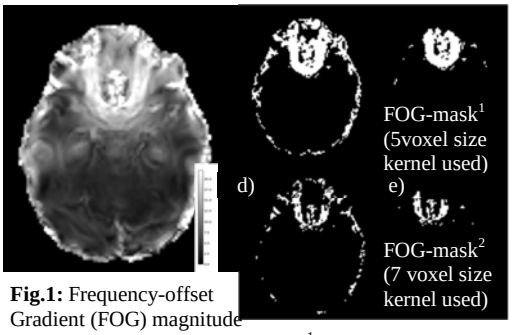

Fig.1: Frequency-offset Gradient (FOG) magnitude image (a), preliminary FOG-mask[1]obtained by thresholding at FOG_{mean} + 3*FOG_{std} (b) and FOG-mask[1] (c) that multiplied with highly eroded mask, (d) preliminary FOG-mask[2], thresholded with FOG_{mean} + 5*FOG_{std} and FOG-mask[2](e).

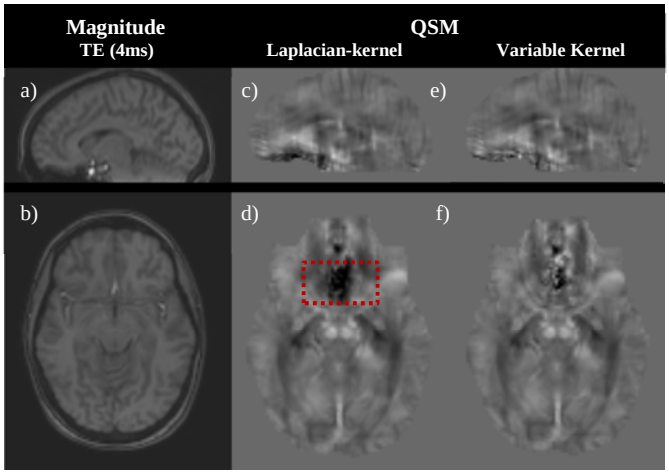

Fig.2: Magnitude images (a, b) (TE = 4ms) (left column), QSM images (c-d: Laplacian kernel used, e-f: variable Kernel used)

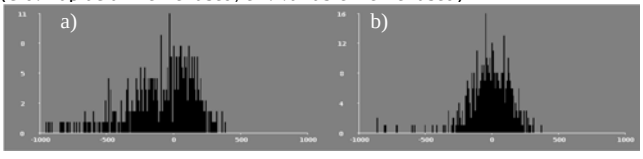

Fig.3: Histogram QSM values, within the red rectangle shown in Fig. 2(d)) as obtained with a) a constant Laplacian kernel b) a regionally adapted kernel.

Discussion & Conclusion The SHARP[7] algorithm in a modified version[1] is currently widely used but – as is obvious from comparison of Fig. 2a) with Fig. 2d) – can suffer from substantial artifacts associated with insufficient removal of non-local field contributions in areas with strong magnetic susceptibility variations.

In this work, we adapted the Laplacian kernel radius depending on the magnitude of the frequency-offset gradient, which is large in regions with strong susceptibility artifacts (Fig. 1a).

The corresponding susceptibility maps, upon visual inspection, show less obviously wrong susceptibility values in the corresponding tissue regions. The histogram analysis indicates more homogeneous susceptibility values these regions, in agreement with information available from anatomical images (Fig. 2 a) and b)). Local kernel-radius adaptation guided by the FOG magnitude could be of special interest for clinical cases involving tumor characterization in frontal regions, in low-SNR studies, and/or in studies involving inhalation of paramagnetic hyperoxic gases.

References: [1] Schweser et al., MRM (2013) [2] Zhu and Schofield, OPTICS LETTER (2003) [3] Özbay et. al, #261 ESMRMB, Lissabon (2012) [4] Smith et al., HBM (2002) [5] Özbay et. al, #3171 Proc. ISMRM 2014