Resolution and Coverage for Accurate Susceptibility Maps: Comparing Brain Images with Simulations

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Synopsis

Magnetic Susceptibility Mapping is moving closer to clinical application. To reduce scan time, clinical images are often acquired with reduced resolution and coverage in the through-slice dimension. The effect of these factors has been studied using only balloon phantoms and downsampled brain images. Here, we used MR images acquired at low resolution or low coverage and compared these with images simulated in volunteers and a realistic numerical phantom. Simulated susceptibility maps were very similar to maps from acquired images. Our results show that low resolution and very low coverage both lead to loss of contrast and errors in susceptibility maps.

Introduction

Magnetic Susceptibility Mapping (SM) is a recently developed MRI technique with emerging clinical applications[1-2]. Clinical images are often acquired with large slice thickness and reduced coverage in the through-slice dimension[3-5] to shorten scans and increase patient comfort and throughput. Recently, the effect of these factors on SM has been studied using a balloon phantom[6] and downsampled brain images[7-10]. Here, we used MR images acquired at low resolution or low coverage and compared these with images simulated in volunteers and an anthropomorphic Zubal head-and-neck, numerical phantom[11].

Methods

Brain images were acquired in four healthy volunteers at 3-Tesla (Philips, Achieva) using a 3D GRE sequence with parameters shown in Figure 1a and slice thicknesses 1, 2, 4 and 6 mm. The same volunteers were also scanned using a 2D GRE sequence with parameters shown in Figure 1b and a through-slice field of view (FOV) of 144, 111, 78 and 44 mm. Two of the volunteers were also scanned with a 20 mm FOV.

The Zubal head phantom[11] was modified: i) to include the neck via co-registration with the torso phantom, ii) by interpolation to achieve 1 mm isotropic resolution, and iii) the oropharyngeal air space was made more realistic using an ellipsoidal shape (Figure 2a). Realistic susceptibility, magnitude and T_2^* values were assigned to several brain regions[12] and a Fourier-based forward model[13] was applied to estimate a field map. Multi-echo complex images were simulated at 3-Tesla, TE_1 = 3 ms, ΔTE = 5.3 ms, 5 echoes (Figure 2c).

Low-resolution complex MRI images were simulated from the full-resolution 3D acquisitions (Figure 2d) and the multi-echo phantom images (Figure 2c) by averaging the complex data across each slab of m = 1 to 6 mm slices (Figure 2h-i). Low-coverage images were simulated from the full-coverage 2D acquisitions (Figure 2b) and the multi-echo phantom images (Figure 2c) by including only the central n = 100% to 14% slices (Figure 2f-g).

Susceptibility maps were calculated from all the 3D and low-resolution images (Figure 2l) using: 1. Non-linear fitting[14], 2. Laplacian phase unwrapping[15] ($\sigma = 10^{-10}$), 3. Projection onto Dipole Fields[16] (PDF) and 4. Truncated K-space Division[17] (TKD, $\delta = 2/3$, with correction for underestimation[15]). The same pipeline was applied to the 2D and low-coverage images with steps 2-3 replaced with joint 2D+3D phase processing [18] (Figure 2k). Brain masks were generated by combining (by intersection) the result of FSL BET[19] on the last-echo magnitude image, and a mask obtained by thresholding the inverse noise map output by the non-linear fit[20].

Brain ROIs were obtained via: i) non-rigid registration[21] of the Eve atlas[22] magnitude image and the full-resolution, full-coverage last-echo magnitude images, ii) the full-resolution, full-coverage susceptibility maps were co-registered[23] with all other susceptibility maps and these transformations were applied to the ROIs from the step i). Mean susceptibilities were calculated in several brain regions (referenced to the posterior limb of the internal capsule[24] and the internal capsule for the volunteer and phantom respectively).

Results and Discussion

Figures 3 and 4 show the mean susceptibility in several brain regions as a function of slice thickness and coverage respectively for the volunteers (Figures 3a & 4a display both acquisitions (x) and simulations (o)) and the phantom (Figures 3b & 4b). Simulated susceptibility maps in both the volunteers and the phantom were similar to maps from acquired images. The susceptibility map contrast decreased with increasing slice thickness (see Figure 3). Figure 4 suggests that susceptibility maps were not affected by decreased contrast until coverage was reduced below 30% of the FOV (44 mm). The reduced contrast at both increased slice thickness and very low coverage is probably caused by insufficient sampling of the dipolar fields induced by the susceptibility sources. The phantom simulations also show that insufficient sampling leads to incorrect susceptibility values (Figures 3b & 4b where the ground truth values are indicated by the horizontal lines). The mean susceptibilities calculated in the full-coverage, full-resolution images depend on the SM pipeline (see Figures 3b & 4b: The susceptibilities estimated at full resolution/coverage are different for the 3D and 2D pipelines). However, the observed trends suggest that all pipelines would also produce inaccurate results at low resolution and very low coverage.

Conclusions

Large slice thickness and very low coverage lead to loss of contrast and errors in susceptibility maps. Resolution and field of view need to be chosen to adequately sample the fields generated by the structures of interest. Simulated susceptibility maps were very similar to maps from acquired images. As calculated susceptibilities depend on the SM pipeline, future work may involve investigating different pipelines.

Acknowledgements

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Figures

	a) Slice thickness	b) Coverage
Sequence	3D GRE	2D GRE
Matrix size	240×240×144	240×240 in-plane
SENSE factor	1×2×1.5	1×2
Resolution	1 mm in-plane	1 mm isotropic
TE_1	3 ms	4.9 ms
ΔTE	5.4 ms	5.3 ms
Number of echoes	5	5
TR	29 ms	4549 ms
Flip angle	20°	90°

Figure 1: MRI acquisition parameters used to investigate the effect of large slice thickness (a) and reduced coverage (b) on susceptibility maps. Brain images were acquired with the parameters in column a) at slice thicknesses 1, 2, 4 and 6 mm, and with the parameters in column b) with through-slice FOVs of 144, 111, 78, 44 and 20 mm. All acquisitions used a 32-channel head coil.
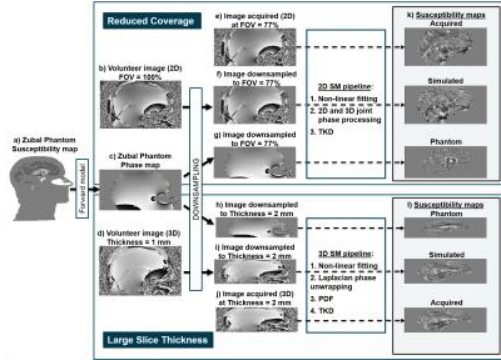

Figure 2: Methods overview. Multi-echo brain image volumes (c) were simulated from a modified Zubal phantom (a) and acquired in a healthy volunteer (b, d). The effect of a reduced FOV (f, g) and resolution (h, i) were investigated by downsampling the images as well as acquiring low-coverage (e) and low-resolution (j) brain images. Susceptibility maps (k, l) were calculated and compared in each case.
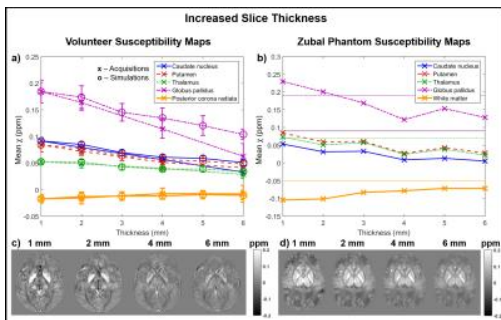

Figure 3: Mean susceptibility in five ROIs as a function of slice thickness averaged across four volunteers (a, x – acquisitions, o – simulations) and in the numerical phantom (b, horizontal lines indicate the corresponding ground truth susceptibility values). Axial slices of susceptibility maps calculated from images acquired at different slice thicknesses are also shown in a representative volunteer (c) and the phantom (d).
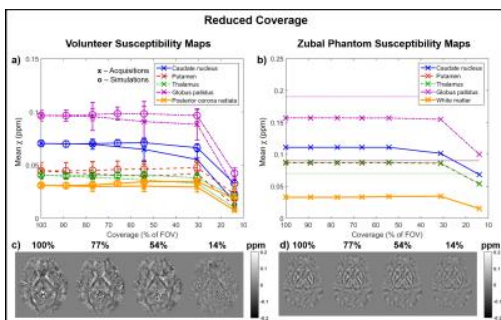

Figure 4: Mean susceptibility in five ROIs as a function of coverage averaged across four volunteers (a, x – acquisitions, o – simulations) and in the numerical phantom (b, horizontal lines indicate the corresponding ground truth susceptibility values except that of white matter where χ = -0.05 ppm). Axial slices of susceptibility maps calculated from images acquired at different FOVs are also shown in a representative volunteer (c) and the phantom (d).